A Fast and Robust Method for Quantifying Magnetic Susceptibility of Arbitrarily Shaped Objects Using MR

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Introduction: Quantifying magnetic susceptibility of biological tissue using MRI has important clinical implications such as for differentiating a hemorrhagic lesion as acute or chronic, in determining its size, identifying calcifications, quantifying iron deposition in sub-cortical structures and measuring oxygen saturation in blood. MR susceptometry of substances in standard geometries (sphere or cylinder) has been done in the past [1,2] and methods for dealing with arbitrarily shaped objects have recently been presented [3,4]. We present here a, fast, novel method for quantifying magnetic susceptibility of an arbitrarily shaped object, from its field distribution measured using MR. This method differs from earlier methods presented for arbitrarily shaped objects in that it uses a Fourier transform based approach [5,6] and is considerably robust and fast in convergence. Previously we made an important theoretical improvement of the Fourier based field calculation method [6] which we use here in our susceptibility quantification algorithm.

Theory, Materials & Methods: The induced magnetic field deviation distribution B(r), within and around an object, can be calculated through the Fourier transformation of the geometry and the tissue susceptibility $\chi(r)$ using $B(r) = B_o \cdot FT^{-1}$**[** $FT[\chi(r)] \,.\, FT[G(r)]$ **]**, where B_o is the main magnetic field, FT stands for Fourier transform, and G(r) is the Green's function (discrete Green's function given in [6]). The parameter $\chi(r)$ is spatial distribution of the susceptibility within the object. For an object with uniform susceptibility, the χ of the object can be separated from its geometry function and $FT[\chi(r)] = \chi.FT[geometry]$. Hence for any arbitrary object with uniform χ value, by obtaining the geometry from a fast gradient echo, high bandwidth, short echo time dataset, and using a measured field map from the phase images, we can estimate the susceptibility of the object. This can be done by fitting its 3D field distribution estimate, from Eq. 1 assuming χ=1, to the measured field through the least squares fit method. That is, minmizing f in Eq 2: $f = \Sigma^{n}_{i=1} SNR_i^2[(\varphi_i - (\varphi_o + \chi.g_i).\gamma.TE.B_o]$, where, 'i' denotes the voxel under consideration, SNR_i is the magnitude signal to noise ratio within the voxel, φ_i is the measured, unaliased phase, φ_0 is a constant phase shift due to rf pulse and frequency adjustment done by the spectrometer (to be determined), g_i matrix is FT^{-1}**[** FT[geometry] . FT[G(r) **]**, χ is the susceptibility value to be determined, and γ and TE denote the proton gyromagnetic ratio and echo time, respectively. Noise in the measured field map (from phase) can influence such a least squares approach considerably. Hence we developed an iterative thresholding algorithm to exclude voxels with noise and those with no considerable phase information and through this, eventually arrive at the right value of χ.

Phase, being $\varphi_i = \gamma.TE.\Delta B$, the amount of phase information and consequently the measurable ΔB information is echo time dependent. The standard deviation of phase is $\sigma_{phase} = 1/SNR_{magnitude}$, is determined by the SNR in the magnitude image, and is only a measure of the expected deviation of phase values from the 'supposed to be' phase value within a given voxel. So, thresholding in phase, to exclude noise voxels using σ_{phase} requires the knowledge of an estimate of the expected φ_i in the voxel. Our algorithm involves the following steps: **(a)** obtain the field map from unaliased phase images; **(b)** for first iteration only: estimate $\varphi_{o,}$ and χ by least squares fitting ($\varphi_{o,initial}$, $\chi_{initial}$), considering all the voxels within the object; Or use a physically reasonable estimate for χ and 0 for φ_o as initial values; **c)** find the voxel set which satisfy $|\varphi_i| > \pm 3\, \sigma_{phase}$; **(d)** find the voxel set which satisfy $\{\gamma.TE.B_o.(\varphi_{o\,(m-1)} + \chi_{(m-1)}.g_i) - 3\sigma_{phase}\} > \varphi_i > \{\gamma.TE.B_o.(\varphi_{o,(m-1)} + \chi_{(m-1)}.g_i) + 3\sigma_{phase}\}$ where $\chi_{(m-1)}$ and $\varphi_{o,\,(m-1)}$ is the χ and φ_o values estimated in the previous iteration; **(e)** find the intersection set of the voxel sets found in (c) and (d) and use them for the least squares fitting and get new values of $\varphi_{o,\,m}$ and χ_m and iterate through steps (d) and (e) until the % change between χ_{m-1} and χ_m is less than 0.1%. We use the above algorithm for quantifying χ of distilled water (with respect to air) contained in a phantom with a complicated geometry ($\Delta\chi_{water/air}$=9.4ppm). Furthermore, we evaluate the robustness of this method by (i)varying the initial χ (about the expected 9.4ppm); (ii) by only using arbitrary sub-sections the phantom for the least squares fitting; and calculate the error in the quantified χ w.r.t the expected 9.4ppm.

A cylindrical polypropylene container (wall thickness 1mm) with a small hollow cylindrical cavity inside, was filled with distilled water and imaged at 1.5T (Siemens Sonata) using the SWI sequence (fully flow compensated) with TR 15ms, FA 6^o, voxel 0.78 x 0.78 x 0.78mm , matrix 256x256x192, BW 610 Hz/pixel (to avoid distortion), and at echo times 6.58ms and 9.58ms to get phase at TE=3ms from which the field map was calculated. Just before imaging the water phantom, shimming was performed using a spherical phantom (dia-170mm, containing $NiSO_4$), to within 6Hz FWHM frequency spread, and the shim current values were noted which were then used while imaging the cylindrical water phantom.

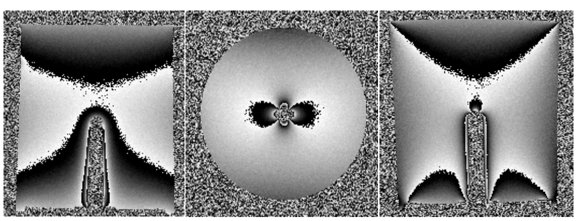
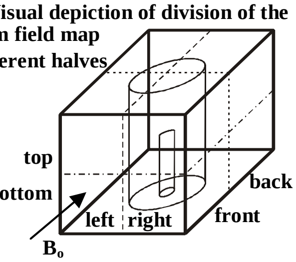


Fig 1: Axial (left), Coronal and Saggital (right) views of phase images of the phantom @ TE=9.58ms

Fig 3: Visual depiction of division of the phantom field map into different halves



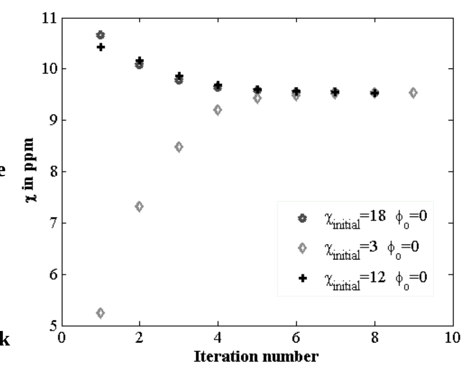


Fig 2: Plot showing convergence of χ value under different initial conditions.

Results: The iteration method, irrespective of the $\chi_{initial}$ value used, converges to the values of φ_o=0.76ppm and $\Delta\chi_{air/water}$ = 9.53 ppm; an error of only 1.4% w.r.t the expected theoretical value of $\Delta\chi_{air/water}$ = 9.4ppm. The convergence, as shown in Fig 2 and Table I, occurs within less than 15 iterations. Even with partial field information, i.e. considering field information from only half of phantom (different halves, Fig 3), the value converges quickly to within 5% of the theoretical value (Table I). The least squares estimation error in the parameters themselves is quite small; on the order of 0.5% of the parameter's value, which is due to the large number of points used in the fitting procedure.

	φo in ppm	χ in ppm	Least Squares Error in φo in ppm	Least squares Error in χ in ppm	% Error in χ w.r.t theory	Iteration number at convergence
All voxels	0.76	9.54	0.00015	0.0033	-1.48	8
front	0.77	9.78	0.00021	0.0045	-4.07	7
back	0.76	9.17	0.00022	0.0048	2.46	11
top	0.80	9.80	0.00020	0.0046	-4.22	5
bottom	0.67	9.50	0.00023	0.0045	-1.11	9
left	0.76	9.54	0.00021	0.0046	-1.50	7
right	0.76	9.53	0.00022	0.0047	-1.42	10

Table I (see figure 3 for orientation)

Discussion and Conclusion: The fact that even with partial field information from the object, this method converges quickly to within 5% of the actual value is quite impressive. And the fast convergence, independent of the initial estimates, shows the robustness of the method. A key requirement for this method is that the 3D geometry of the field perturbing object is to be well defined; else discretization error in the representation of the object could increase the error considerably. Since FT is linear in nature, field due to any complicated χ distribution can be represented as the sum of fields due to many sub-structures, each with uniform χ. Thus, it lends itself easily to be applied to even complicated distributions of χ. And, being based on FT, the implementation of this method is quite fast even for large matrices. In summary, we present here a fast, highly robust and easy to implement method for quantification of susceptibility of arbitrarily shaped objects using MR.

References: 1. Weisskoff RM, et al. Magn Reson Med 1992:375–383. 2.Holt RW et al. J Magn Reson Imaging. 1994: 809-18. 3. Li L. Magn Reson Med. 46(5):907-16. 4. Li L et al Magn Reson Med. 51(5):1077-82. 5. Marques et al. MR Eng. 25:65, 2005. 6 Neelavalli et al., ISMRM, 1016, 2007.