A model-free method for high-precision MR susceptometry

C. Barmet[1], N. De Zanche[1], L. S. Sasportas[1], and K. P. Pruessmann[1]
[1]Institute for Biomedical Engineering, University and ETH Zurich, Zurich, Switzerland

INTRODUCTION:
NMR and MRI of small samples frequently suffer from susceptibility broadening, which is induced by the variation of magnetic susceptibility at material interfaces. One generic remedy to this problem is susceptibility matching, e.g. by immersing the probe head in a suitable liquid [1]. To fully implement this approach, the magnetic susceptibility of both probe and matching materials must be controlled with great precision, often below the 0.1 ppm scale. To this end we describe a novel MRI-based method of measuring the magnetic susceptibility of liquid, solid, and gaseous samples. Unlike previous methods of MRI susceptometry [2-5], the new procedure does not rely on model assumptions and accommodates samples of arbitrary geometry. Moreover it works in inhomogeneous background fields and the samples do not need to yield NMR signal themselves. The new approach is illustrated by high-precision susceptibility calibration of paramagnetically doped liquid and polymer materials for miniature NMR field probes [6].

THEORY AND METHODS:
In conventional MRI-based susceptometry [2-5], model assumptions are required for computing the unknown magnetic susceptibility. Such assumptions include perfect knowledge of the sample geometry, a perfectly homogeneous B_0-field over the sample, a net magnetic field that points exclusively in the z direction (for the validity of the Laplace equation), the presence of regions of homogeneous magnetic susceptibility, smooth surfaces or the absence of partial volume effects. Additionally many methods are limited to measuring the susceptibility of NMR-active samples.
Such experiments are therefore often demanding to set up and limited in accuracy by residual violations of the model assumptions. The proposed method works without a model of the B_0 variation, relying instead on two reference samples of known susceptibility. The experimental setup is depicted in Fig. 1, consisting essentially of a material sample immersed in water doped with Gd-DOTA (Dotarem, Guerbet). The water serves as the source of MRI signal and was doped merely for SNR optimization. Liquid or gaseous samples are contained in a glass vial. A circular receive coil serves for MR signal detection. When imaging any plane or volume of this setup with a gradient-echo sequence, the image phase at TE reflects the magnitude of the local static magnetic field. When changing only the sample material the change in phase at any point in the image is proportional to the change in sample susceptibility; the effects of any kind of background fields created, e.g., by the magnet, the setup, or the sample vial, can be eliminated by phase subtraction (see Eq. 1). Hence, based on three scans, one with the target sample (S) and one each with two reference samples (R1, R2) of known susceptibility, the susceptibility of the target sample can be determined as:

$$\chi_S = \chi_{R1} + (\chi_{R2} - \chi_{R1}) \cdot avg\left[\frac{\varphi_S - \varphi_{R1}}{\varphi_{R2} - \varphi_{R1}}\right] \qquad (1)$$

where denotes image phase, χ denotes magnetic susceptibility and avg denotes the average over all image pixels depicting imaging liquid. Before evaluation of (1) it is crucial to a) unwrap the phase difference images and b) apply a mask that accounts for image regions corrupted by the chemical-shift of the sample or signal dephasing. The avg of the bracket in (1) is obtained by least squares fitting of the numerator and denominator maps, including a constant offset to account for the arbitrary choice of the unwrapping seed. The insertion of different samples can lead to slight variation in image warping due to local B_0changes. This confound can be countered by B_0 unwarping [7] or, in moderate cases, ed by high-bandwidth readouts, which proved sufficient in the present study. B_0 perturbations can also result in slightly varying slice selection, which also was negligible here but could be addressed in principle by 3D imaging with B_0 unwarping. Conductive samples can only be measured when they are broken into small pieces such that no significant eddy currents are induced.
The chosen reference substances are water (-9.032 ppm at 20°C [8,9]) and air (+0.358 ppm at 20°C, 10^5Pa and 50% relative humidity [10]). The selection of sample materials was inspired by the demands of miniature NMR field probes [6] (see Fig. 3) and included ethanol, D_2O, cyclohexane, solutions of $MnCl_2.4H_2O$ in water, enameled copper wire, and an epoxy polymer doped with varying amounts of $Dy(III)NO_3$. All imaging was performed on an Achieva 3T system (Philips Medical Systems, NL), using a Cartesian gradient-echo sequence with TE=40ms, TR=43ms, matrix=256x256, slice thickness = 3.6mm.

RESULTS:
For the $MnCl_2$ dilution series, Fig. 2 shows a strictly linear relation between the concentration and the susceptibility, which is expected in the examined regime of low $MnCl_2$ concentrations. A maximum deviation of the data points from the linear regression of 0.011 ppm indicates a precision of the method of around 0.01 ppm. For an ethanol sample the error bound was studied in more detail: ethanol images with 1 to 25 averages were used to evaluate equation (1); water and air data were acquired with 32 averages each. Already with relatively few averages (•15), a precision of below +/- 0.005 ppm can be attained. The susceptibility of ethanol was determined as -7.23 ppm, which deviates by 0.03 ppm from the literature value (of unknown accuracy) of -7.26 ppm [11]. To measure the susceptibility of copper wire for probe head construction the enameled wire (diameter 0.4mm) was cut into pieces of • 3mm length and poured into a vial. The vial was then filled with Fluorinert FC-72 (3M, USA). Using the volume ratio wire/FC-72 of 1.09 and the measured susceptibility of pure FC-72 (-7.97 ppm), the wire susceptibility was computed to -8.90 ppm. Susceptibility measurements on an epoxy polymer showed that it needs to be doped with 0.223mg of $Dy(III)NO_3.5H_2O$ per gram of epoxy. For the D_2O (-9.01 ppm) inside the glass capillary (Fig. 3) it was found that it needs to be doped with 7.5 mM of $MnCl_2.4H_2O$ to match cyclohexane (-7.68 ppm). The suitability of these matching values was confirmed in a field probing study that will be described in a separate submission.

DISCUSSION AND CONCLUSION:
The presented MRI-susceptometry method is straightforward and highly sensitive, enabling greater precision in the choice and design of susceptibility matching materials. It enables assessing a large range of practically relevant materials.

REFERENCES: [1] DL Olson et al. Science Vol. 270;5244:1967 (1995), [2] RM Weisskoff, MRM 24;375-383 (1992), [3] O Beuf, JMR 11;111-118 (1996), [4] ZJ Wang, JMR 140;477-481 (1999), [5] L Li, MRM 46;907-916 (2001), [6] N. de Zanche et al., ISMRM Seattle (2006), p. 781, [7] P Jezzard, MRM 34;65-73 (1995), [8] JF Schenck, Med. Phys. 23(6);815-850 (1996), [9] JS Philo, J.Chem.Phys. 72(8);4429-4433(1980), [10] RS Davis, Metrologia 35;49-55 (1998), [11] CRC Handbook of Chem. and Phys., 86th Ed.

Fig.1: Measurement setup. The sample is contained in a glass vial that is immersed in an imaging liquid. The setup must enable precise repositioning of the sample.

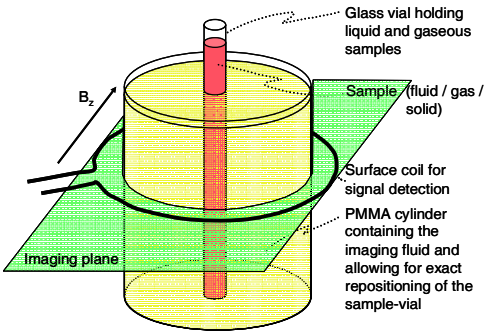

Fig.2: Measured susceptibility of a $MnCl_2$ dilution series confirming the expected linear relation between $MnCl_2$ concentration and magnetic susceptibility.

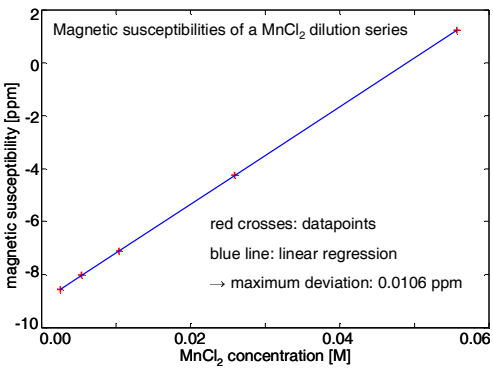

Fig.3: Schematic of an NMR probe head according to [6]. The region outside the glass capillary (copper, epoxy) as well as its inside (cyclohexane, D_2O) need to be of homogeneous susceptibility, respectively.