Simple and Robust Design for Susceptibility-Matched NMR Magnetic Field Monitoring Probes

P. Sipilä[1,2], D. Lange[3], S. Lechner[1,4], W. Loew[1], P. Gross[1], M. Baller[3], G. Wachutka[2], and F. Wiesinger[1]

[1]Imaging Technologies Lab, GE Global Research, Garching bei München, Bavaria, Germany, [2]Institute for Physics of Electro Technology, Technical University of Munich, Munich, Bavaria, Germany, [3]Sensor Technology Lab, GE Global Research, Garching bei München, Bavaria, Germany, [4]Institute for Nuclear Medicine, Technical University of Munich, Munich, Bavaria

Introduction:

Recently magnetic field monitoring (MFM) has been introduced as a novel concept, which permits simultaneous measurement of induced **k**-space signals together with the exact, corresponding encoding locations in the **k**-space (1). In this concept, conventional MRI data acquisition hardware is supplemented by appropriate signal-generating NMR probes (2). These probes permit extraction of different orders of global $B_0(\mathbf{r},t)$ information over the region spanned by them. In its simplest implementation with three probes and a 2D imaging experiment, MFM enables measurement of the spatially-constant $B_0(t)$ offset and the 2D **k**-space trajectory. It is expected that MFM significantly improves applications, which either require high gradient waveform fidelity (i.e. aggressive single-shot acquisition schemes) and/or suffer from eddy current distortions (i.e. diffusion and phase contrast imaging). In this work, we present a novel method to manufacture NMR-probes for magnetic field monitoring and demonstrate their usefulness for the case of spiral imaging.

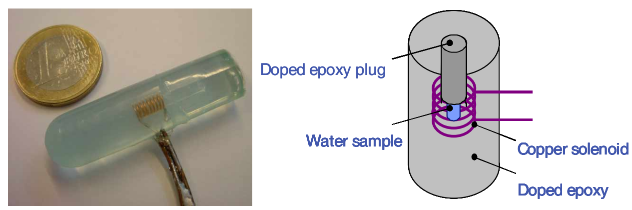

Fig. 1 Water filled NMR-probe cast inside susceptibility-matched epoxy.

Methods:

Using NMR probes with very small signal droplets (~1 μl) permits measuring time-resolved magnetic fields based on the detected signal phase $\varphi(t)$, according to:

$$\varphi(t) = \varphi_0 + \mathbf{k}(t)\mathbf{r} + \gamma\Delta B_0(t)t \qquad [1]$$

with φ_0 a constant phase offset, $\mathbf{k}(t)$ the **k**-space location at time t, **r** the spatial position of the signal droplet, γ the gyromagnetic ratio and $\Delta B_0(t)$ the time-dependent, spatially-constant main magnetic field offset. In order to successfully perform the required phase unwrapping throughout a whole readout period, the NMR probes have to provide both a high baseline signal-to-noise ratio (SNR) and long signal lifetimes. The high SNR is relatively easily achieved by using small diameter solenoid coils in combination with non-conductive samples (4). For achieving long signal life times (i.e. long T2* relaxation times), susceptibility-matching methods have been applied (2,3).

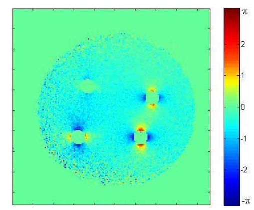

Fig. 2 B_0-field maps of four epoxy samples with different doping levels in a water container.

Compared with previously presented five-component NMR probe designs (2,3), our method drastically simplifies manufacturing the probes (Fig. 1). Embedding the signal water droplet and the solenoid coil into a cylindrical shaped long diamagnetic epoxy block is an effective way to improve the field homogeneity across the droplet. An advantage of our epoxy casting method is that, with a proper mould design, one is able to have great freedom over the geometry of the sample void, e.g. a sphere or a cylinder. In our implementation, we have chosen a cylindrically shaped droplet (diameter 1 mm, length 2 mm) placed in the constant sensitivity region of a solenoidal receiver coil. The 2 mm diameter coil was made by tightly winding eight turns of copper wire with the diameter of 400 μm (Fig. 1). To further increase the T2* relaxation time, the susceptibility of the diamagnetic epoxy was fine-tuned by doping with paramagnetic Er^{3+} ions to match the similar susceptibilities of water ($\chi_{H2O} \approx$ -9.1 ppm) and copper ($\chi_{Cu} \approx$ -9.6 ppm). Accordingly, with this three-component only NMR probe design no further susceptibility matching is required.

In order to avoid any additional B_0 homogeneity degradation, the matching and decoupling circuit was placed some centimetres away from the coil. This decreased the SNR by 30 %, but was a good compromise between the baseline SNR and the signal lifetime limited by the T2*. The probe was connected with a coaxial cable to a low noise preamplifier. Spiral imaging experiments were conducted by interfacing three NMR probes and a single circular surface coil (diameter 70 mm) to a 3 T GE Signa Excite HD system (GE Healthcare, Milwaukee, WI).

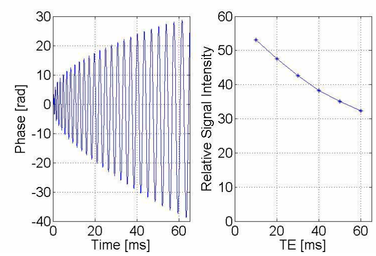

Fig. 3 Unwrapped phase obtained from a spiral acquisition (left) and a FID curve of a NMR-probe (right). The T2* time calculated from the FID is 100 ms.

Results and Discussion:

In order to find the proper Er^{3+} concentration for susceptibility matching, a concentration series of differently doped epoxy samples was evaluated (5). Fig. 2 shows acquired B_0 maps, which well illustrate the expected influence on the dipole-shaped B_0 variation. The Er^{3+} concentration to match the susceptibility of water was identified by the zero crossing of the signed B_0 dipole distribution (~1.3 mmol/l). Three susceptibility matched NMR probes were fabricated and characterized showing baseline SNRs above $3 \cdot 10^4$ and T2* times of up to 100 ms. These values comfortably meet the requirements for evaluating $\Delta B_0(t)$ and $\mathbf{k}(t)$ according to Eq. [1]. The probes can easily follow the phase throughout the 64 ms long read-out time of a single shot spiral scan (Fig. 3). Fig. 4 illustrates the benefit of MFM for multishot spiral imaging obtained with a purposely-degraded gradient system without (left) and with (right) MFM monitoring.

Our novel design of casting the droplet void into epoxy greatly simplifies the manufacturing of robust NMR probes still exhibiting the high performance required for magnetic field monitoring.

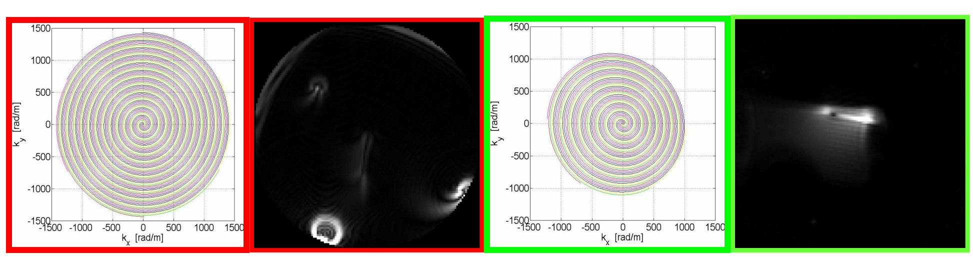

Fig. 4 Multi-shot spiral imaging example (6-arm spiral, BW 250 kHz, 4096 points) using a deliberately misadjusted gradient system. This example clearly shows the benefit of using magnetic field monitoring (right) as compared to the case of performing a reconstruction based on the assumed, ideal trajectory (left).

References: (1) K.P. Pruesmann et al., ISMRM 2005: p.681, (2) De Zanche et al. ISMRM 2006: p.781, (3) D. L. Olson et al., Science, 270: 1967-1970 (1995), (4) K.R. Minard et al., Conc. Magn. Reson. 13(5): 190-210 (2001), (5) C. Windischberger et al., JMR 20: 730-734 (2004).

Acknowledgements: Some aspects of this work received co-funding under the Bavarian research initiative "Leading Edge Medical Technology Programs"