

Continuous magnetic field mapping with pulsed ^1H NMR probes

P. Sipilä^{1,2}, J. Tropp³, S. Greding¹, G. Wachutka², and F. Wiesinger¹

¹GE Global Research, Munich, Bavaria, Germany, ²Institute for Physics of Electrotechnology, Munich University of Technology, Munich, Germany, ³Global Applied Sciences Laboratory, GE Healthcare, CA, United States

Abstract: Non-ideal magnetic field profiles can severely degrade image quality, especially with fast-imaging sequences, phase contrast and diffusion tensor to mention a few. Artifacts arise from imperfect gradient performance, non-homogenous B_0 field and/or from eddy-currents. Previously, a method to measure magnetic fields with dedicated sensors, simultaneously during image acquisition, and apply the information within image reconstruction, has been proven a promising solution to significantly reduce these imaging artifacts [1-5]. The most promising results have been obtained with susceptibility-matched transmit-receive NMR probes [6,7]. However, the susceptibility matching complicates the manufacturing process, increases costs and hinders the miniaturization. Respectively, strong gradient performance (for example in high resolution imaging) can significantly shorten the effective read-out time of MFM probes. This drawback has been partly dealt with by reducing the NMR sample size, at the expense of SNR. In this paper, we present a novel method for the operation of magnetic field monitoring (MFM) probes based on continuous, short-TR, broad-band excitations. Therefore, no susceptibility matching is required anymore, simplifying the manufacturing and improving the probe integration.

Materials and Methods: A short solenoid coil gives the best sensitivity profile compared to other designs with comparable dimensions [8]. A droplet of water is applied as the NMR sample. The T_1 relaxation time of the sample is adjusted with copper sulfate to increase SNR efficiency with short TR values. Respectively, the T_2 relaxation time is reduced and, thus, T_2^* value will be less dominated by field-inhomogeneities. Counter-windings in the solenoid are applied to reduce coupling with imaging elements tuned to the same frequency. Short hard-pulses for the excitation are generated by chopping the output of a frequency synthesizer with high-speed switches [7]. The timing and pulse RF widths are accurately controlled with a 16-bit microcontroller. The pulses are amplified to obtain sufficient RF excitation power levels using a separate 5W RF amplifier (Mini-Circuits, Brooklyn, NY). A passive duplexer-based transmit-receive switch protects the receiver preamplifier during excitations (Fig.1).

One of the challenges with pulsed NMR is the dead time after each RF excitation pulse. This is primarily caused by the ringing of the reactive elements within the NMR probe and the preamplifier (e.g. capacitors and inductors). Dependent on the components and circuit design chosen, the dead time can reach values beyond $\sim 10 \mu\text{s}$. In our work, we have modified a standard low input impedance preamplifier by removing the LC circuit before the input HEMT (high electron mobility transistor). Respectively, the coil matching should be modified to match the noise figure minima of the HEMT. With VHF frequencies, this commonly corresponds high impedance with some (negative) reactance. This simplifies the NMR probe matching and tuning, which is now well achievable with only one element, a parallel capacitor. As the gate of a HEMT is seen as almost an open circuit, with small capacitance in parallel, the damping of the complete circuit comes simple by applying a resistance in parallel to the coil and to the preamp. In comparison to active damping, we have selected a static resistance value: a decision which compromises with the circuit complexity, damping efficiency, probe noise figure, and probe matching to HEMT noise minima. A reduction in ringing to value below $2.5 \mu\text{s}$ after RF pulse is achieved. Respectively, the detuning effect of HEMT with resistance in parallel is calculated to be slightly more effective than the traditional preamp-decoupling scheme.

Magnetic field information is extracted based on the signal phase evolution according to the Larmor relation [2].

Corrupted points due to each RF pulse are eliminated, and the resulting phase jumps are numerically interpolated. To obtain the intrinsic phase decay of the samples, a calibration scan, or short measurement period (at the order of tens of microseconds) is performed. Obtained k-space data is then used for non-Cartesian image reconstruction. As no methods was implemented to obtain phase coherency between each excitation pulses and the NMR signals, noticeable loss in SNR can be expected. As ^1H NMR probes have proven high SNR, this is considered acceptable. The inhomogenous B_0 field within the sample can cause error in characterizing the inherent phase evolution, φ . This error can be approximated as $d\varphi_{\text{max}}/2\pi = (1/T_2 - 1/T_2^*) \cdot dt$, where dt is the differential time step. A value equaling φ_{max} would be the case when there is two spin isochromats within the sample. As the end of each repetition can be considered as the start of a new measurement period, the accumulated error can then be approximated to be $2\pi \cdot \text{TR} \cdot (1/T_2 - 1/T_2^*)$ at the worst. With exemplary values for TR, T_2 and T_2^* , of 100 μs , 8 ms and 5 ms, respectively, this would equal an error of ~ 0.23 rad or 0.29 nTs with proton in 3T background field. It is estimated that the MFM should have accuracy comparable to 1 nTs [4].

Results: Fig.2 shows the advantage of exciting the NMR probes multiple times. Field monitoring signal can be obtained infinitely long, apart from the time periods where the RF excitation corrupts the NMR signal. In this experiment, a water based NMR sample was used with the T_2 adjusted to ~ 8 ms. The probe was placed into GE Signa Excite 12M4 3T scanner (GE Healthcare, Milwaukee, WI). 800 ns RF pulses with intervals of 500 μs were applied, and points equaling the length 2.5 μs were removed from the data set during to each excitation. Next, four NMR probes were placed around ROI to characterize gradient performance during circular echo planar imaging sequence. The ^1H imaging coil was turned off during this procedure, as artifacts arising from the probe excitations would be too severe. A change in the NMR nuclei would allow us a continuous monitoring, also during the actual imaging sequence. Fig. 3 shows promising results of tracking the k-space trajectory with pulsed NMR probes.

References [1] G.F. Mason et al., Magn. Reson Med., vol.38, p.496-492,1997, [2] K.P. Pruessmann et al., ISMRM 2005:p.681, [3] Sipilä et al., Sens. Actuators, A, vol. 145-146, pp.139-146, (2008), [4] N. de Zanche et al., Magn. Reson. Med., vol. 60, pp. 176-186, 2008, [5] C. Barmet et al, Magn. Reson. Med., vol. 60, pp. 187-197, 2008, [6] Barmet et al. ISMRM 2008: p.1152, [7] P. Sipilä et al. ISMRM 2008, p.680 [8] K. R. Minard et al., Concepts Magn. Reson., vol. 12, pp. 190-210, 2001.

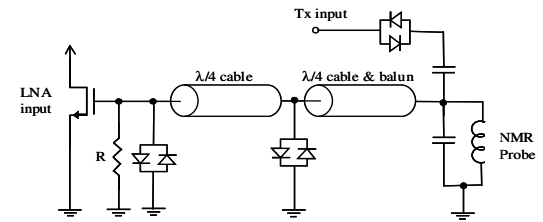


Fig. 1: Electrical schematic of the NMR probe, and how it is connected to low noise preamplifier (LNA) and transmit chain. Note the resistance R to damp the ringing of the NMR probe.

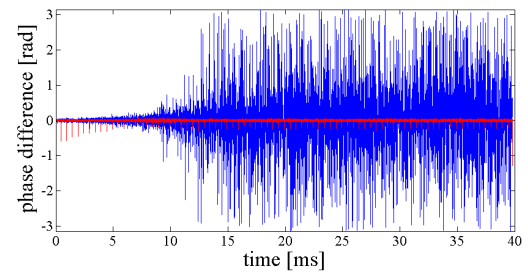


Fig. 2: Comparison between the normal single excitation, and multiple excitation phase profiles. As normal FID converges to noise, the multiple excitation profile can be applied for continuous field monitoring despite the short T_2 value of the NMR probe. Note the phase jumps due to multiple excitations.

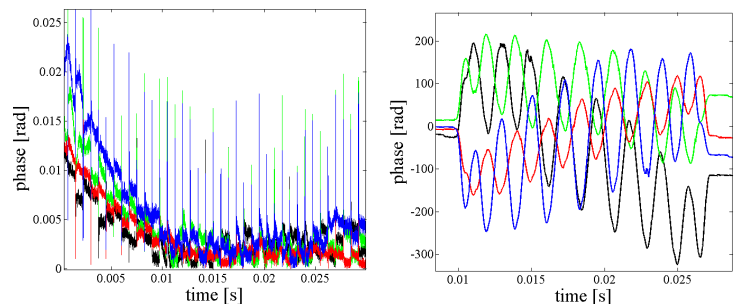


Fig.3: Monitored phase evolutions of four NMR probes during a circular echo planar imaging sequence (right). The signal absolute values shows clearly when the RF excitations occur (left)