Susceptibility-matched ^2H_2O NMR probes for magnetic field monitoring

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ABSTRACT: Measuring magnetic field profiles real-time during imaging acquisition promises reduced level of artifacts arising from B0 offsets, gradient imperfections and/or eddy-currents. Recently, efforts have been made to develop field sensors accurate enough for magnetic field monitoring (MFM) in MRI. The best results so far have been achieved by monitoring phase evolution of a signal from small NMR samples [1,2]. The work has been concentrated on developing ^{1}H based NMR probes as the probes can easily be connected to a standard multi-channel receiver system, without any additional adjustments with respect to the receiver hardware [3-5]. To gain independency from imaging sequences, the probes are preferred to be transmit-receive design. A broad-band pulse with appropriate power levels is applied to the probes to obtain optimal signal to noise ratio. However, as the NMR probes and imaging elements are operated at the same frequency, the effect of cross-talk, and signal coupling from NMR samples to imaging elements, and imaged object to the NMR probes, cannot be completely avoided. This gives rise to certain image artifacts and/or inaccuracy in the field monitoring data. Shielding the probes [6], using counter-wound solenoid geometry [7], compromises between coupling and sensing accuracy. To completely overcome these drawbacks, NMR probes based on another nucleus as the imaging object is preferred. In this abstract, we present the first ^{2}H based NMR probes for MFM. For increased flexibility those probes are operated by a separate transmit-receive, multi-channel, broad-band NMR spectrometer. An imaging experiment with accompanied field monitoring has been done, and the effects to improve image quality have been proven.

MATERIALS AND METHODS: A solenoid coil is wound around the NMR sample, as it has a proven high sensitivity with respect to coil dimensions. A material of deuterium oxide (D_2O) has been chosen for the NMR sample for the following reasons: 1) D_2O has same preferable high spin density as H_2O, 2) its liquid form gives a high intrinsic T2 relaxation time, which is required for long-readouts, 3) chemical and magnetic properties are close to H_2O, thus susceptibility matching and adjusting the T1 and T2 relaxation times can be performed with similar methods as with H_2O, 4) due to different Larmor precession frequencies coupling distortions to other nuclei (i.e. proton imaging) are completely eliminated, 5) respectively, one can perform high resolution imaging with other nucleic 6) because of the small Larmor frequency the probe to probe coupling is reduced, 7) phase unwrap algorithm has smaller chance of failure because of the lower signal bandwidth. However, loss in detected signal level is expected to accompany the smaller gyromagnetic ratio. The known formula for amplitude signal to noise ratio (SNR) gives us the basis to study this effect [7]:

$$\frac{A_{signal}}{A_{noise}} = \frac{\omega \Delta V \cdot \dfrac{2\mu_z^2 B_0 n \left| S^- \right|}{2k_b T}}{\sqrt{4k_b \Delta f R}}, \quad (1)$$

where ω denotes Larmor frequency (angular), ΔV sample volume, μ_z is magnetic moment, B_0 main magnetic field strength, n spin density, S^- coil sensitivity, k_b Boltzman constant, T temperature, Δf receiver bandwidth, and R total coil resistance. Thus, the electromotive force (emf) induced to coil is expected to reduce by factor of γ^3, as $\omega, \mu_z \propto \gamma$. However, if we consider signal dephasing,

$$\frac{1}{T2^*} = \frac{1}{T2} + \gamma(\Delta B_0 + \mathbf{G_{xyz}} \cdot \mathbf{\Delta d_{xyz}}), \quad (2)$$

where $\mathbf{G_{xyz}}$ is the gradient profile, and $\mathbf{\Delta d_{xyz}}$ is the sample dimensions, we found that the dimension can be increased to compensate this γ^3 loss in emf. In general the sensitivity of a solenoid coil $\propto 1/d_{coil}$, [7] where d_{coil} is the coil diameter, and therefore, loss in sensitivity can be expected. Respectively, higher level of noise will be observed due to increased conductor length (e.g. $R \propto d_{coil}$). This effect, however, is fully compensated by the reduced bandwidth of the same factor. To summarize, approximately a decrease of $1/\gamma$ in SNR should be expected when changing to nucleus with lower gyromagnetic ratio. Some optimization in coil geometry, like number of turns, wire thickness etc. should be done to improve the coil sensitivity.

A copper coil was wound around a 1.8 mm diameter glass capillary, which was filled with heavy water (^2H_2O) were cast into a cylindrical shaped epoxy. The epoxy was susceptibility-matched to the value of copper. The sharp sensitivity drop-off provides volume restriction in the longitudinal direction. The NMR probes were used in transmit/receive configuration [7]. For the excitation of the NMR probes, a separate RF source, chopped with a high-speed switch, provided the broadband pulses required for high flip angles despite the present k-space location of the probes. Passive duplexers were constructed to protect the preamplifiers during RF excitations. Additional gain and anti-alias filtering is implemented before the signal was sampled with a self-made multi-nuclear receiver. The rest of the standard heterodyne receiving scheme was implemented software-wise by using LabVIEW in a standard PC. The transmitter and receiver timing properties were configured by a 16-bit microcontroller. A triggering pulse and sharing the same master clock synchronized the MFM toolkit with the clinical MRI device.

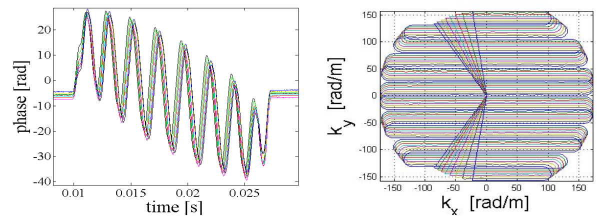

Fig.1 Monitored NRM Signal phase (left) and k-space trajectory (right) during circular Echo-Planar Imaging sequence with eight interleaves. The signal has good SNR throughout the whole acquisition window

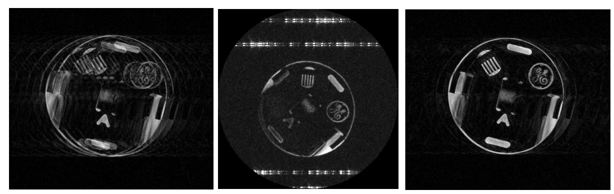

Fig.2 Circular Echo-Planar Image (8 interleaves) without any MFM correction applied (leftmost). Same sequence, but corrected with the field monitoring data from ^{1}H probes [7] (center). ^{2}H MFM probes provide improved image quality without any additional artifacts (rightmost).

Results and discussion:

The signal to noise ratio of the ^{2}H NMR probes was ~ 8×10^4 $Hz^{1/2}$, and T2* was ~100 ms, which were found to fulfill the requirements for MFM probes [7]. With the self-made receiver, the center frequency can be adjusted separately for each probe, which provides advantages with respect to the required phase-unwraping in the MFM post-processing.

A GE Signa 12M4 3T system (GE Healthcare, Milwaukee, WI) was accompanied with the MFM hardware based on four ^{2}H NMR probes. The probes were placed around a resolution phantom and circular EPI sequence was applied (Fig.1). The magnetic field profile was simultaneously monitored, and the information applied in the non-Cartesian k-space reconstruction algorithm [5]. The coupling between the imaging coil and the MFM probes was completely eliminated (Fig. 2). However, some ghosting artifacts were still observed despite the field correction data. It is expected that optimizing the SNR of the ^{2}H NMR probes and improving the reconstruction algorithm, the effectiveness of MFM can be even further increased.

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