# Coherent excitation scheme to operate pulsed NMR probes for real-time magnetic field monitoring

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**Introduction:** Utilizing magnetometers to monitor image-encoding gradients in MRI can improve image quality in novel acquisition schemes that are requiring high-fidelity gradient performance such as non-Cartesian imaging, phase-contrast, and fMRI [1-4]. Unlike calibration methods, the field monitoring does not lengthen image acquisitions, nor is it vulnerable to parameter drift between calibrations. NMR based magnetometers have been shown to fulfill the strict precision requirements for such a field monitoring system. The operating scheme is typically based on a single RF excitation, after which the following free-induction-decay curve is monitored [5-7]. In this method, the probes are usually optimized with susceptibility matching techniques for over 100 ms long monitoring periods. Such a scheme is not functioning optimally with (ultra) short TR acquisitions, and severe SNR loss is expected.

A method to excite probes multiple times during each imaging repetition has been presented to overcome this drawback [8]. The pulsed method simplifies the manufacturing of NMR probes as no susceptibility techniques are required. In addition, the probe sample sizes are not as strictly limited by the applied imaging resolution, i.e. by the gradient induced dephasing. Thus, high monitoring accuracies become achievable also with short TR, and high-resolution imaging. For further benefit of the scheme, the probe integration to magnet bore, imaging coils, and patient table becomes less challenging due to higher robustness against susceptibility artifacts or fringe fields. In this work, a novel feedback based excitation scheme is presented to preserve the phase coherency between the precessing spins and the consecutive RF pulses. This method offers higher SNR and more robust functionality for the pulsed NMR probes, which otherwise would have pseudo-random signal amplitudes distributed over a steady-state value [8].

**Materials and Methods:** A transmit-receive $^1$H NMR probe based on passive duplexer design is manufactured for proof-of-concept experiments. It is estimated that a sampling rate of 50 kHz is sufficient for monitoring the relevant gradient and eddy-current components. This limits the length of the RF pulses to <20 μs including the transients. The repetition rate of transmit pulses, $\Delta T$, is dependent on the chosen sample dimension, $\Delta r$. In order to avoid signal dephasing to zero it is required that

$$\Delta T < \frac{\pi}{\gamma G_{\max} \Delta r}, \qquad (1)$$

where $\gamma$ is the gyromagnetic ratio, and $G_{max}$ is the maximum applied gradient strength. To ensure sufficient longitudinal magnetization for each excitation, the NMR sample should be doped in a way that T1 ≈ $\Delta T$ ~ 100 μs. Consequently T2* ≈ T2, thus probe performance becomes less dependent on the background field inhomogeneities. Each excitation can introduce residual phase errors to the signal despite any applied smoothing algorithm, and therefore the number of pulses per time unit is desired to be kept minimal. This indicates medium power RF pulses, ~20 dBm.

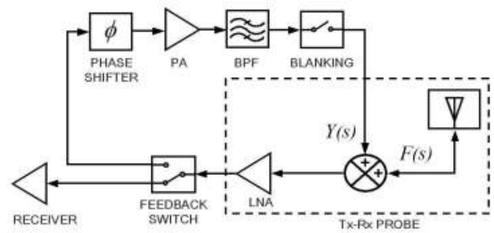


Fig. 1: Block diagram of a positive feedback based exciter to ensure phase coherency between RF pulses and the spin ensemble.

A block diagram of a circuitry ensuring the phase-coherent excitations is shown in Fig. 1. The design is based on a positive feedback that detects the received signal and amplifies it to power levels corresponding to the required flip-angle. If the band-pass filtering is omitted, one can simplify the closed loop gain of the system to

$$Y(s) = \frac{Ae^{-\tau s}}{1 - Ae^{-\tau s}} F(s), \qquad (2)$$

where $Y(s)$ is the Laplace transformed feedback signal, $F(s)$ the input signal, $A$ the gain, and $\tau$ the delay. The system is inherently unstable for $Ae^{-\tau s}>0$. If only the stimulating frequency component, i.e. $\omega_l$, is considered, the system is simulated (LT Spice, Linear Technologies, Milpitas, CA) to follow roughly a delayed exponential growth, thus

$$y(t) \approx \alpha e^{(\beta + i\omega_l)(t-\tau_0)}, \qquad (3)$$

where $\alpha$, $\beta$ are constants. The exponential growth is limited by the saturation in the gain block of the feedback loop. By adjusting the delay, one ensures that the feedback signal is in correct phase with the precessing spins. The total electrical delay of the feedback loop is restricted by dispersion, i.e. one needs to ensure that the phase response does not significantly vary over the signal bandwidth, e.g. $\Delta\varphi < \pi/10$. In practice, cable lengths as well as filter and amplifier phase responses are considered.

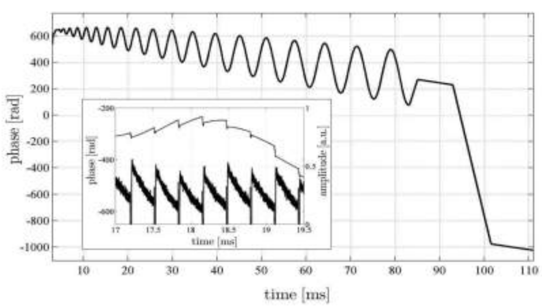


Fig.2: Monitored phase of a NMR probe during a spiral k-space encoding is shown. In the box, the unprocessed amplitude and phase signals are highlighted.

The feedback gain should be high enough in order to ensure that the NMR signal drives the system into saturation instead of any unstable poles of the feedback, to keep the transient time before the saturation short, and to overcome the coupling losses in the transmit-receive switch. Gain ripple over the passband should be avoided, as otherwise the system can be driven into saturation by the passband noise. In this work, a total open loop gain of ~65 dB is chosen with the 1 dB compression point of 26 dBm, defined by the output power amplifier (ZHL-3010+, Mini-Circuits, Brooklyn, NY). Surface acoustic wave filters offer narrow passband with tolerable dispersion, and the phase adjustment of the feedback loop is done with a transmission line of an appropriate length. No initiating RF exciter, e.g. signal source, is required, as the spins picks-up the correct Larmor frequency component from the noise amplified by the unstable feedback loop. After starting the monitoring sequence, the system is self-maintaining and in principle infinitively long acquisition periods become available.

**Results and Discussion:** The manufactured $^1$H probe (T2*~280 μs, 1.4 mm sample diameter, and 2.1 mm length), is placed inside a GE Signa Excite 12M4 3T scanner (GE Healthcare, Milwaukee, WI), approximately 20 cm off from the isocenter. A separate NMR receiver and control circuitry for the feedback loop is utilized for the monitoring [9]. The signal evolution during a spiral sequence is illustrated in Fig. 2 (256 kHz bandwidth, 4096 points, 1.2 mm resolution). The steady state is observable, and apart from the short 15 μs long excitation periods occurring every 250 μs, continuous monitoring of magnetic fields becomes available. Based on the average SNR of $2.1 \cdot 10^4$ Hz$^{1/2}$, monitoring accuracy of ~8.9 nT/Hz$^{1/2}$ is expected at 50 kHz sampling rates [3], which fulfills well the estimated requirement of ~150 nT/Hz$^{1/2}$.

The performed experiments proof the concept of the feedback based phase-coherent excitation scheme for operating NMR probes. The developed scheme makes the operation of NMR probes for magnetic field monitoring less dependent on applied imaging parameters, e.g. TR, or resolution. By chancing the bandpass filtering, one can apply the feedback scheme for operating non-proton NMR probes [8], and it is assumed that both technologies are required to make the real-time magnetic field monitoring feasible for clinical applications.

**References:** [1] G.F. Mason et al., Magn. Reson Med.,vol.38, p.496-492,1997, [2] K.P. Pruesmann et al., ISMRM 2005:p.681, [3] N. de Zanche et al., Magn. Reson. Med., vol. 60, pp. 176–186, 2008, [4] Sipilä et al., Sens. Actuators, A, vol. 145-146, pp.139-146, 2008, [5] Barmet al. ISMRM 2008: p.1152, [6] P. Sipilä et al. ISMRM 2008, p.680, [7] P. Sipilä et al. ISMRM 2009, p. 3076, [8] P. Sipilä et al. ISMRM 2009, p.782, [9] Sipilä et al. Concepts Magn. Reson. Part B, vol (in press), 2009.