An extremely broadband low-frequency MR system

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ABSTRACT

NMR, MRI and NQR explore the phenomenon of resonant absorption and emission of RF energy by nuclear spin systems. Conventional MR electronics also employs a front-end resonant circuit for efficient RF transmission and reception. Such resonant circuits are analog narrow-band devices that require frequency and impedance tuning to achieve efficiency and tend to make multi-frequency probes very complex, expensive and inflexible. We devise a broadband approach to the MR front-end electronics so that both the transmission and reception do not need resonant circuits. This approach has significantly simplified the probe circuit, electronics and operation, with efficient power transmission and reception as well as robust operation without tuning. This results in flexibility and simplicity for a wide range of low-frequency MR experiments, such as studies of porous media.

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1. Introduction

The fundamental signal of magnetic resonance (MR) comes from the resonant radiation and interaction of an RF magnetic field and the nuclear spin system. Metallic wire-wound coils are commonly used to excite nuclear spins by passing an RF current and generating an RF magnetic field, and also to detect the nuclear spin magnetization by receiving the electric current induced in the coil. Traditionally, experimental apparatus has embedded this coil within a tuned circuit to achieve efficient transmission of the RF power for spin manipulation and significant signal gain for high-fidelity reception [1]. This technology was borrowed from radio and radar electronics and the resonant circuit, typically consisting of an inductor (coil) and a capacitor forms the front-end of the NMR probe. The rest of the MR electronics has benefited from tremendous changes to the digital electronics during the past two decades. For example, rapid advances in the speed, reliability, and capabilities of digital electronics have resulted in the wide adoption of direct digital synthesis and high speed analog-to-digital conversion. However, the front-end of MR remains the resonant circuit – an analog device.

The tuned front-end circuit consists of the NMR coil (L) in parallel with a capacitor (C) and acts as an analog filter at the resonance frequency \( \omega_L = 1/\sqrt{LC} \). The bandwidth of this filter is \( \omega_L/Q \), where \( Q \approx \omega_L/L \) is the quality factor of the circuit and \( R \) is the series resistance of the coil. The value of \( Q \) is typically on the order of 100 for non-conducting samples. A second capacitor is often added in series with the tuned circuit to adjust its impedance to match that of the power amplifier and receiver (typically 50 \( \Omega \)). When the circuit resonance frequency \( \omega_L \) is adjusted to the Larmor frequency \( \omega_L \) of the nuclear spin, efficient power transmission and reception can be achieved near \( \omega_L \). Adjustment of the resonant frequency is typically carried out manually. Because of the narrow bandwidth, multi-frequency MR probes are built by combining several resonant circuits, each resonating around the Larmor frequency of one nuclear species. Such probes are complex and expensive due to the strong interaction between the resonant circuits, particularly at low frequencies where \( Q \) is typically lower.

Assuming a matched condition, one may derive the relationship between the input power (P) and the resulting current amplitude in the coil, \( I = \sqrt{2P/Q\omega_L} \). Similarly, for the receiver, the signal gain of the circuit close to resonance is \( \sqrt{Z_0Q/(\omega_L)} \) (assuming matching to an impedance \( Z_0 \)). As a result, the performance of the resonant circuit is highly sensitive to the power dissipation in the coil and the environment. In a field application of NMR, such as well-logging, the Q-factor is known to change significantly depending on temperature, pressure, conductivity of the sample and environment, which makes calibration of the NMR signal amplitude more difficult.

In this paper, we describe fully broadband MR front-end electronics that does not use a resonant circuit for NMR excitation or reception. We use conventional RF pulses to manipulate the spins, unlike non-resonant methods based on reversing the direction of the static magnetic field [2]. However, unlike in tuned probes, our system synthesizes the RF waveform inside the coil and thus controls the spin excitation current directly. It also directly senses the MR signal current induced in the coil so that the reception is self-calibrated. Our approach is fundamentally different from ultra-broadband MR systems constructed with delay line or transmission line probes [3–6]. Such probes are considerably less...
sensitive detectors than simple coils, such as solenoids. By con-
trast, we use a standard, highly sensitive solenoid coil as our de-
tector. The broadband nature of our electronics allows rapid switching
of the operating frequency with a range of over a decade. This al-
lows multi-frequency and multi-nuclear MR experiments with a
single coil. The final narrow detection bandwidth is defined digi-
tally in the later stages of the electronics.

2. Experimental

A simplified block diagram of our broadband MR system is
shown in Fig. 1. The key features include a single un-tuned trans-
mitt and receive coil, a simple switching transmitter using an H-
bridge circuit [7], low-noise, broadband gain of the received signal
using a transformer and a broadband FET switch-based duplexer
with isolated driver. A commercial NMR spectrometer (Kea 2, from
Magritek) is used to program the pulse sequence, synthesize low
level RF signals, digitize the receiver output and perform further
signal processing.

The entire system was integrated onto a printed circuit board
(PCB) with four wiring layers. The layout of this board is shown
in Fig. 2. The board has two independent RF input channels, labeled
CH1 and CH2. It also allows an external signal (labeled MOD) to
control the transmitter threshold voltage, thus making it possible
to generate shaped pulses via duty cycle modulation.

The sample coil was a solenoid with ID = 5.5 cm and
length = 8.8 cm, consisting of 27 turns of AWG 16 magnet wire
wound around a Teflon form with variable pitch (\( \frac{C}{2} \) mm at the
center, \( \frac{C}{6} \) mm at the edge). Its self-inductance was 15 \( \mu \)H after
assembly inside a tight shield box. The voltage gain of the receiver
was \( \sim 70 \) dB between 80 kHz and 4 MHz. Its measured input-re-
ferred noise with the present coil is in very good agreement with
simulations and is lower than the thermal noise of a 1 \( \Omega \) resistor
for frequencies between 60 kHz and 3 MHz, i.e., over a 1:50 fre-
quency range. As a result the sensitivity of our system over this en-
tire range is within \( \pm 3 \) dB of that obtained over narrow bandwidths
by a tuned probe. This statement was verified by measuring the
noise figure of the receiver as a function of frequency and also by
directly comparing the SNR obtained with the un-turned sample
coil with that obtained when the same coil was tuned and matched
at several discrete frequencies over this range.

3. Results

We have used our system for NMR experiments in the inhomo-
genous fringe field of a 2 T superconducting magnet. Relaxation
and diffusion measurements have been successfully performed
for Larmor frequencies between 98 kHz and 3.2 MHz with no hard-
ware tuning. The receiver recovery time was measured to be less
than 40 \( \mu \)s over this entire frequency range. Measurements of
long-\( T_2 \) samples such as benzene and distilled water match ex-
pected values, showing that the transmitted RF pulses are stable
in both amplitude and phase.

The key advantage of a broadband system is that the RF fre-
quency can be easily and rapidly changed, because the hardware
does not need to be modified or re-tuned in any way. In \textit{ex situ}
NMR, the excited sample volume is often power-limited by large
sample size and/or inhomogeneous magnetic field. We can easily
switch frequencies to move this region (often called the slice).
We refer to this process as depth profiling or 1D imaging. Using
the broadband system, switching to different nuclei also becomes
trivially simple without adjustment of the hardware system. For

![Fig. 1. Simplified block diagram of the broadband MR system.](image1)

![Fig. 2. Layout of the circuit board, measuring 25.1 cm x 10.9 cm in size.](image2)
example, we have measured depth profiles of a brine sample using both proton and sodium signals.

Our multi-frequency capability enables a multi-slice approach to many time consuming experiments, quite similar to the method used in MRI for 3D imaging. For example, in order to measure diffusion, several acquisitions have to be made with different diffusion weightings. In a conventional single-slice technique, substantial waiting time is necessary in between different scans in order for the spin system to recover its equilibrium. In the multi-slice mode, acquisition can be moved to a different slice without the waiting time, resulting in a significant acceleration of the measurement, as shown in Fig. 3. In this figure we show the results of applying a “diffusion editing” sequence [8] to multiple slices in a static field gradient. This sequence begins with a direct spin echo refocusing cycle with large echo spacing $T_0$. The encoding period $T_D$ is varied across the slices to get different amounts of diffusive attenuation, given by $e^{-g^2f_0^2T_D^2/12}$, where $g$ is the gradient and $D$ is the diffusion coefficient. It is followed by a train of refocusing cycles with short echo spacing $T_E$, where $T_E$ is kept short enough to eliminate further diffusive attenuation. The echoes produced by this train are added together to increase SNR.

A similar approach can be used to accelerate $T_1$ measurements. In general, any indirect dimension of a multi-dimensional experiment can be encoded in different slices. The multi-slice mode can also be used to simply increase the SNR per unit time by eliminating wait times between scans that are added together. The encoding period $T_D$ is kept short enough to maximize SNR, resulting in $S \approx 1 + T_W/(N_N T_2)$, where $T_W$ is the wait time that would normally be applied between scans to allow the longitudinal magnetization to recover, and $N_N T_2$ is the total length of the train of $N_N$ refocusing pulses used to acquire the signal. Typically we have $T_W = 3 \times T_1$ while $N_N T_2 = 1.26 \times T_2$ to maximize SNR, resulting in $S \approx 1 + 2.4(T_1/T_2)$. The resulting increase in SNR per unit time is therefore particularly significant for samples with large $T_1$ / $T_2$ ratios. We have even used very rapid frequency switching to excite and refocus multiple slices within the same CPMG sequence.

Our system also allows excitation of very broad spectra using frequency sweep pulses or composite pulses [9]. Such broadband pulses allow the excited volume and SNR to be increased in inhomogeneous fields without increasing the peak RF power level. Another application is measurement of the same sample at different Larmor frequencies. Such experiments are normally performed with expensive field-cycling electromagnets and are limited to $T_i$ measurements. By contrast, our system allows any NMR experiment to be performed as a function of Larmor frequency.

Our system is also well-suited for performing double-resonance experiments, such as measurements of heteronuclear $J$-coupling constants in inhomogeneous magnetic fields [10]. It is also useful for $^{14}$N NQR, in which typical resonant frequencies range between 0.1 and 5 MHz [11]. The latter has important applications in the detection of explosives and illegal drugs.

### 4. Discussion

In this paper we have described an extremely broadband system for low-frequency MR. Our design addresses the challenges of low-field NMR and NQR, such as low SNR, inhomogeneous magnetic fields and limited bandwidth of tuned probes. Some of these concepts may also be useful for high-field NMR. Our overall goal was to extend the numerous advantages of digital systems, such as programmability, flexibility and robustness, to the front-end electronics of MR, which have traditionally been constructed with resonant analog circuits. Another benefit of our approach is ease of use. By making the front-end electronics extremely broadband, we allow the user to focus on the resonances and dynamics of the spin system, rather than that of the electronics.

The performance of our current system is already good enough to perform a wide variety of MR experiments. In addition to improved circuit design, developments in semiconductor devices will also allow us to further improve performance in the future. For example, low-noise transistors, constructed from high-mobility compound semiconductors such as SiGe may make it possible to increase the bandwidth of the receiver and lower its noise figure. Similarly, improved high-power FETs constructed from materials such as GaN and SiC may allow us to increase the power-handling capability and bandwidth of the transmitter.

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### References