W. Overall[1], G. Scott[1], M. McConnell[2], J. Pauly[1]
[1]Electrical Engineering, Stanford University, Stanford, CA, United States, [2]Division of Cardiovascular Medicine, Stanford University Medical Center, Stanford, CA, United States

Optically Coupled Op Amp Transmitter for use with Interventional Microcoils

Introduction: **The use of micro-receiver coils in interventional MRI is relatively common, whether for active intravascular tracking or high-resolution imaging [1,2]. Microcoil transmission has also been demonstrated previously by placing an attenuator in series with the output of the power RF amplifier [3] or by using a dedicated power amplifier [4]. This work describes the design and implementation of an op-amp-based IC amplifier for small-volume excitation. This approach, used with a twin-lead or solenoid coil configuration, produces localized SAR well below accepted limits while providing localized B1 fields sufficient for extremely rapid excitation. Secondary heating of nearby guidewires and pacemaker leads is also not a concern because of the low total power deposition. To reduce the likelihood of device heating in the presence of body-coil RF excitation, the micro-transmit coil is optically coupled to the RF hardware, thereby significantly shortening its electrical length. Pulse-sequence gating of the op-amp transmitter enables real-time switching between body-coil and interventional transmission. Spins near the interventional coil may be selectively tagged or inverted to highlight specific anatomy.**

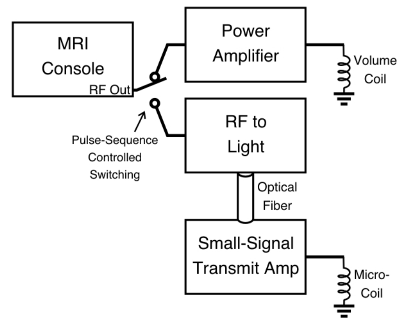

Figure 1: *Block diagram of transmitter circuit. The power RF amplifier input is optically coupled to an op-amp amplifier at the proximal catheter end.*

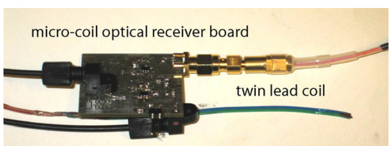

Figure 2: *Small-signal transmit amp, powered by a 9V battery (not shown). RF and gating fiberoptic connections attach from the left. Circuit board is approx. 3cm x 3cm.*

Methods: The transmitter hardware (Fig. 1) was connected to the small-signal RF output of a GE Signal 1.5T research system. This signal and a software-controlled gating logic signal were optically modulated using Analog Devices OPA843 op amps configured to drive two high-speed LEDs (Industrial Fiber Optics IF-E99). Optical fibers connect to a small-signal transmit amplifier (Fig. 2), located in the magnet bore, which demodulates and amplifies the optical signal to power the coil. The gating signal disables the transmission op-amp during reception so that output noise does not saturate the receiver (either the body coil or a 5-inch surface coil in these tests). Both circuits are powered with nonferrous 9V lithium-ion batteries. This amplifier design can source a maximum of 29 mW at 64 MHz, and is primarily limited by the slew-rate capabilities of the high-speed driver op amps (OPA847).

A twin-lead transmitter (shown in Fig. 2) and a 9-turn 7-French solenoid were tested. For the solenoid, SAR is roughly proportional to the square of coil radius; therefore, microcoil SAR deposition can be shown to be low even when transmitting continuously at the power limit of the amplifier. In addition, this low transmitted power prevents dangerous heating of other guidewires or pacemaker leads that may be present during an intervention.

Hard-pulse and adiabatic half-pulse excitations of varying lengths were combined with a 2DFT gradient-echo pulse sequence. Imaging time depended on the duration of the excitation pulse used, but was less than 6 seconds per image in all cases. Independent logic enable signals for the power transmitter and op-amp transmitter allowed the excitation coil to be rapidly changed from acquisition to acquisition.

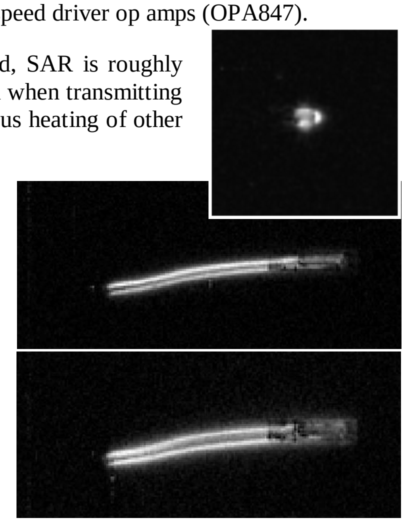

Figure 3: *Hard-pulse excitations. (top): Solenoid, 3.6-ms pulse. (middle): Twin-lead, 3.6-ms pulse. (bottom): Twin-lead, 10-ms pulse.*

Results: Images acquired using this apparatus (Fig. 3) demonstrate good SNR in the region of the coil. For the twin-lead coil, spin saturation near the coil occurred with a 3.6-ms hard RF pulse. The excited region could be pushed further from the coil by increasing the hard-pulse amplitude or duration (Fig. 4), while overtipping decreased signal adjacent to the coil. A 10-ms adiabatic excitation showed higher, more uniform signal near the coil, with a maximum SNR around 60 in the region near the coil. For a given pulse sequence, images exhibit similar peak SNR regardless of solenoid or twin-lead configuration.

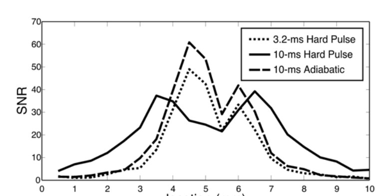

Figure 4: *SNR as a function of position perpendicular to twin-lead catheter. Longer hard pulses generate maximum signal further from the catheter.*

Discussion: High-quality images can be acquired with intravascular micro-coil excitation. Unlike receive-mode microcoils, where the imaged region is fixed by the coil sensitivity pattern, the transmit-mode region of sensitivity is moved away from the coil by increasing transmit power. Adiabatic excitations of sufficient strength and duration may increase signal away from the coil without sacrificing signal near the coil. This concept may be applied to intravascular imaging and navigation with virtually no risk of catheter heating. Because image SNR is primarily dictated by the receiver hardware, image quality is relatively unaffected by the configuration of the intravascular coil and RF amplifier as long as adequate tipping is achieved.

Conclusion: The feasibility of op-amp-driven interventional transmit coils has been demonstrated. Using such a coil instead of volumetric excitation decreases the risk of catheter heating, especially when combined with fiberoptic coupling between the MR scanner and catheter. Image SNR may be improved through the use of adiabatic excitation. When used in conjunction with a second RF transmitter for excitation over a larger volume, localized spin tagging and transmit-SENSE imaging is possible.

References:

1. Atalar E, *et al.* MRM 36(4): 596-605, 1996.
2. Konings MK, *et al.* MRM 13(1) 131-135.
3. Herlihy DJ, et al. JMRI 13(1): 127-130, 2001.
4. Yeung CJ, *et al.* JMRI 12(1): 86-91, 2000.