High Q Reactive Network for Automatic Impedance Matching

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Introduction: Magnetic resonance imaging and spectroscopy (MRI/S) of laboratory animals requires extensive infrastructure support, from protocol review, to housing, to life support monitoring during experiments. Monitoring equipment must fit into limited bore space already filled with coils and the animal, making access to variable capacitors for adjustment of impedance match very difficult (Figure 1). Most importantly, there is significant impedance shift from animal to animal in high frequency MR, requiring readjustment of matching components for every animal. To address these issues, we have developed a high-Q reactive network attached to an automatic impedance matching (AIM) system that enables quick and efficient coil optimization *in situ*, with the animal placed at the homogeneous center of the magnet, negating the need for suboptimal bore simulator tuning or long tuning wands attached to variable capacitors. Acquired images demonstrate a highly efficient automatic matching system when compared to a single tune coil with discrete capacitors.

Methods: A MRI coil in its simplest form can be modeled as an inductor in series with a small resistor, the resistor representing the combined losses of the inductor metal and the conductive load. The impedance of this circuit must be transformed to the system impedance (usually 50 Ω) by the use of an impedance matching network; characteristically shunt and series reactances attached to the inductor terminals. High Q ceramic capacitors or air/Teflon variable capacitors are typically used for impedance matching so that component losses are kept small compared to those attributed to the load. To achieve electronic control of the shunt and series matching components, we explored two designs; a multi-branch ceramic capacitor array with FET switches in every branch, and a shunt-series varactor network. The capacitor array was abandoned due to poor switch characteristics of the FET; i.e. either high resistive losses in the ON state or high parasitic capacitance in the OFF state. Alternatively, hyperabrupt and abrupt varactor networks were evaluated. We successfully implemented a varactor network, shown in Figure 2, which consists of nonmagnetic, high Q, abrupt junction devices in the GC1500 series by Microsemi. The GC1500 varactors range from 0.18 pF to 22 pF with Q values of 5,000 to 2,200, respectively, VHF-Ku bands. Comparatively, ATC100B ceramic capacitors in this range have Q values from 2,000 to 250 @500 MHz and 10,000 to 1,000 @150 MHz. Also evaluated, but rejected, were hyperabrupt varactors BB639 and BB833 (Infineon Technologies), which had broader capacitance ranges (2.6 pF-48 pF and 0.75 pF-13 pF, respectively) but much lower Q values (<50), consequently degrading the performance of the circuit.

The 2 cm diameter copper foil coil and varactor network were implemented in a receive-only, single tune configuration and placed into a transmit-only volume coil for operation at 470 MHz in an 11.1T, 40 cm clear bore Magnex magnet with Bruker Avance console. As such, a PIN diode decoupling circuit was added across the terminals of the receive-only coil to shift the resonance during transmission. The PIN diode was a GC4215 (Microsemi) with very low series resistance (<0.35Ω @60 mA) and very low reverse bias capacitance (<0.5 pF @ -30V). The low capacitance was important because it added to the shunt varactor capacitance, limiting the smallest capacitance attainable. DC blocking capacitors (ATC100B, 1000 pF) and low loss RF chokes (JW Miller 9210 series) were used to direct the DC voltages to the appropriate varactor and PIN diode. This varactor network was interfaced to the AIM system [1], which implemented a search algorithm to monitor reflected power as the varactor voltages were varied. Minimum reflected power indicated impedance match to 50 Ω and then the varactor voltages were held constant. The match point of the varactor network was adjusted near the minimum varactor capacitance (high DC control voltage) by a distributed capacitor in the coil. This was done to avoid RF signal modulation of the varactor capacitance due to the AIM system output of ~0 dBm. On the other hand, if the RF power output from AIM system was too low, the reflected signal was too small and the S11 would not be good (~ -10 dB). Consequently, pushing the match point to the higher voltage range avoided a linearity problem with the varactors and improved the search result. The last crucial part of the circuit was a single pole double throw (SPDT) switch that directed the signal path to either the AIM system or the MRI system. The SPDT switch was a 50 Ω, low loss, PIN diode device by Aeroflex-Metlics, MSW2000-200, with 0.15 dB insertion loss, transmit-receive isolation of -40 dB, and transmit-antenna isolation of -60 dB. The nonmagnetic MSW2000-200 was driven by Impellimax driver chip 9943 (low magnetic). CMOS switches were evaluated, but not used, because they had higher insertion losses (≥0.3 dB) than the PIN switch.

Control cables for the varactors and SPDT switch were fed into the shielded magnet room through the filter plate. The coils were loaded with a tissue equivalent phantom [2] that had electrical characteristics of average abdomen @ 470 MHz (ε_r=63, σ=1.3 S/m). Spin echo images (TR=500 ms, TE=15 ms, FOV=12 cm, matrix=512 x 512, slice thickness=0.5 cm) were acquired and signal-to-noise ratio (SNR) measured. The system was compared to an equivalent diameter, copper wire, single tune coil with discrete capacitors (ATC ceramic and Voltronics variable).

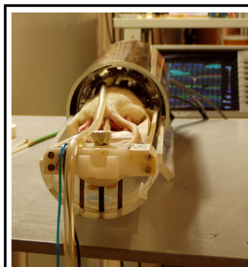

Figure 1.
Limited space of *in vivo* setup showing animal, animal support and monitoring leads, volume TX coil, surface RX coil, and bore simulator

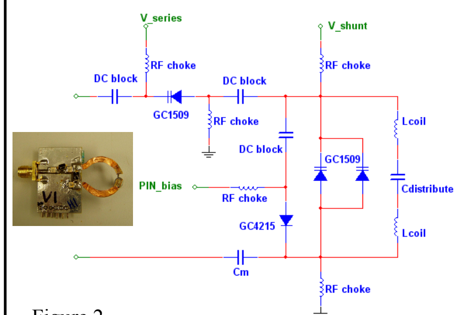

Figure 2.
Schematic and picture of varactor network

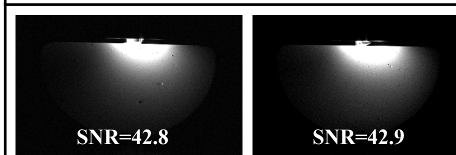

Figure 3.
Spin echo images of tissue equivalent phantom; automatic matching system left, and discrete capacitors right

Results: The high-Q reactive network and AIM system (with coil and sample at the center of the magnet bore) achieved a return loss of -25dB in less than 5 seconds. Spin echo images, using a coil with this automatic system and a coil with discrete capacitors, are shown in Figure 4. The SNR, measured with equivalent regions-of-interest (ROIs) in both images, was 43.8 for the automatic circuit and 43.9 for the discrete-capacitor circuit. The noise standard deviation was slightly higher with the automatic circuit, most likely due to the insertion loss in the switch, but the signal mean was higher.

Conclusions: MRI/S study of animal models requires large numbers of animals for statistical validity. There can be a significant shift in coil impedance at high frequency with different animals, requiring adjustment of impedance match for each. This is typically a time consuming and difficult activity involving adjustment of variable capacitors in limited bore space. We have developed a highly-efficient (high-Q) reactive network and automatic impedance matching system that allows rapid (in seconds) circuit optimization *in situ*, reducing stress on sedated animals and improving experimental efficiency.

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References: [1] Wu S, *et al.*, ISMRM 2010, #3920. [2] Beck BL, *et al.*, MRM, **51**:1103-1107 (2004).