A Wirelessly Programmable Implant Coil for Increased NMR Signal Sensitivity at Multiple Frequencies

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Introduction

Noninvasive monitoring of bio-artificial organs post-implantation is crucial for the development of tissue-constructs capable of physiological treatments to chronic diseases without the need for immunosuppressive medication (1). Monitoring the concentration of important metabolic nuclei using nuclear magnetic resonance (NMR) imaging and spectroscopy provides correlations between construct functionality and physiological effects in addition to providing early indicators of construct failure (2). However, deep-tissue implantation sites, small tissue-construct cell densities, and NMR coil design inherently limit the measurement signal sensitivity and penetration depth. A direct method for increasing NMR signal sensitivity is to implant an electrically resonant coil around the region of interest (ROI) to form an inductive link with the external NMR excitation/acquisition coil, increasing signal acquisition at the frequency of interest through the strongly coupled resonant (SCR) effect. Current techniques are unable to provide significant increase in signal sensitivity at multiple NMR frequencies due to the reliance on discrete components to achieve electrical resonance, thus limiting the acquired metabolic signature to a single nucleus as well as constraining NMR image acquisition to a single magnetic field strength. To overcome these limitations, a digitally controlled capacitor can be used to selectively resonate the implant coil across a frequency range spanning multiple NMR nuclei (3). This work describes the design and measurement results of a magnetic compliant device using an application specific integrated circuit (ASIC) to provide wireless control of the implant coil resonance for increased NMR signal sensitivity at multiple frequencies and magnetic field strengths.

Methods

The device in Fig. 1a consists of a single-turn NMR detection coil (L_{NMR}), ASIC, receiver coil (L_{RX}) with impedance matching capacitor, and discrete capacitors (C_S) for energy storage. Within the ASIC is an integrated digital capacitor (D-Cap) that is connected directly across the NMR coil terminals, forming a parallel LC tank whose resonance is controlled by programming the capacitance using a 10-bit digital word. Data packets are extracted from wireless radio-frequency (RF) transmissions coupled to L_{RX} using an integrated receiver front-end (RX), clock & data recovery (CDR), finite-state machine (FSM), and latch memory to store the digital capacitor value on-chip. A power management unit (PMU) enables untethered wireless operation by extracting energy from the RF transmissions to continuously power the ASIC throughout NMR acquisition. The RF powering scheme in Fig. 1b employs a transmit (TX) and SCR coil to leverage the uniform field distribution in the co-rotating frequency mode for increased wireless power transfer and decreased sensitivity to implant coil misalignments. Isolation between the RF powering and NMR coil pairs is achieved through orthogonal coil orientation in addition to RF carrier operation (80MHz) outside of the 4.7T/11.1T NMR band.

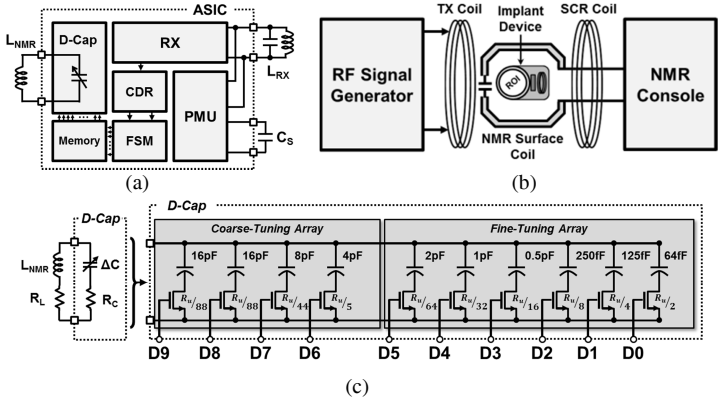

Figure 1. (a) Device system diagram, (b) NMR diagram, and (c) D-Cap schematic

The D-Cap in Fig. 1c is designed to resonate a 20nH coil across the 11.1T NMR frequencies of ^{31}P (190MHz), ^{19}F (442MHz), and ^{1}H (470MHz), including the 4.7T ^{1}H frequency (200MHz). Capacitive control is realized through metal-oxide-semiconductor field effect transistor (MOSFET) devices with minimum channel length for implementation as digital switches. Each branch shown is comprised of multiple sub-branches, consisting of unit sized capacitive elements and MOSFET devices, whose total capacitance equates to the necessary capacitive step size. While a digital approach allows for discrete control of the equivalent output capacitance, the effective series resistance (ESR) of the MOSFET channel degrades resonant quality factor (Q) in addition to MOSFET parasitic capacitance contributions limiting the highest attainable resonant frequency. To balance the tradeoff between resistive losses and capacitive tuning range, the D-Cap is separated into two capacitive banks (coarse & fine tuning) with differing unit sized capacitor values. A larger unit sized capacitor within the coarse-tuning array reduces the number of branches required to form the large capacitive step sizes, thus reducing overall parasitic capacitance contributions and area overhead. To counteract the inherent degradation in Q at higher frequencies, a smaller unit sized capacitance within the fine-tuning array decreases the branch time constant to maintain resonant Q across the total frequency range.

Results

The ASIC was fabricated in a 1.2V 0.13µm CMOS process with 1.5x0.7mm^2 die area, Fig. 2a. The assembled device is shown in Fig. 2b and measures 1.7x1.3cm^2. The coils are embedded within the FR-4 substrate using PCB traces to reduce discrete component count while providing increased mechanical strength. L_{NMR} consists of a single-turn copper trace with 1cm diameter and 30mil trace width (~18nH). The RX coil is constructed using parallel top and bottom layer PCB traces, connected with vias, to form a 5-turn solenoid coil (~150nH) imbedded orthogonally to L_{NMR} to ensure coil decoupling. The device is coated in polydimethylsiloxane (PDMS) using a multi-step coating process for biocompatibility. Through complete wireless control, the device can resonate L_{NMR} across a 152MHz–490MHz frequency span which includes the 11.1T NMR frequencies of 190MHz, 442MHz, and 470MHz. NMR spin-echo images were acquired on tissue equivalent gel phantoms with dielectric properties similar to a mouse abdomen (ε'= 63, σ=1.3S/m) to mimic tissue loading (4). Measurements were acquired in 4.7T (200MHz) and 11.1T (470MHz) magnetic field strengths using a surface coil coupled to the wirelessly programmed device at a 1cm depth. To quantify the increase in signal sensitivity provided by the implant device, NMR images were also acquired using the surface coil alone. Inclusion of the wireless device increased the acquired SNR within the ROI by 3.8dB and 2.6dB within the 4.7T and 11.1T magnetic field strengths respectively, Fig 2c and 2d.

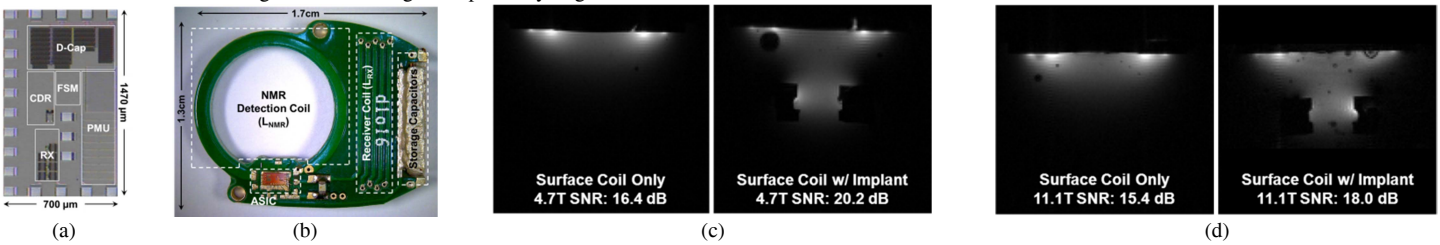

Figure. 2 (a) ASIC micrograph, (b) Assembled device, and NMR images on gel phantoms with and without the implant device in (c) 4.7T and (d) 11.1T field strengths.

Conclusion

These results validate the operation and performance of the proposed implantable device. Through complete wireless operation, the device can selectively resonate a NMR detection coil across a frequency range spanning multiple NMR nuclei. The acquired NMR measurements show that the device increases NMR image signal sensitivity by 3.8dB and 2.6dB within 4.7T and 11.1T magnetic field strengths respectively when coupled to an external NMR surface coil. The design can be further modified to achieve resonances at NMR frequencies within other magnetic field strengths.

References: 1) Lim F, *et al.,* Science 1980;210(4472):908-910. 2) Constantinidis I, *et al*., Annals of the New York Academy of Sciences 2001;994:93-95. 3) Turner WJ, *et al*., ISMRM 2010, #424. 4) Beck BL, *et al*., Concepts Magn. Reson. 2004;20B(1):30-33.

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