Design of Actively Decoupled Implant Coil System with Improved B1 Homogeneity in Rat Spinal Cord

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Introduction. An implanted coil can be used to increase SNR within internal sample regions due to its small size and limited reception volume [1]; however, B1 homogeneity is degraded. This limits its use in applications where good B1 homogeneity is required (e.g. quantitative T2 mapping in rat spinal cord [2]). Wirth et al. [3] improved homogeneity with an actively decoupled implanted coil in a cat, but this system is too bulky to implement in smaller animals since the diode switching circuit is built directly on the implant. Here we describe an active decoupling scheme that is similar to [4], but requires no decoupling circuitry on the implanted coil.

System Description. The coil system consists of the implant coil, a pickup surface coil with a diode-activated decoupling circuit, and a quadrature birdcage coil for transmission (Fig. 1). The implant and pickup are pretuned to the same "self-resonant" frequency ω_s in isolation and are then brought close together, so that two current peaks of equal amplitude are formed symmetrically above and below ω_s. This "overcoupling" occurs during reception, such that the low frequency peak is at the Larmor frequency ω_o [1], while the birdcage coil is decoupled by its own diode-activated circuit. During transmission, a diode-activated trap circuit detunes the pickup coil, removing its coupling to the implant and volume coil. The implant's resonant frequency returns to ω_s (between the two frequencies of the two overcoupled split peaks), which is sufficiently high enough to avoid interacting with the birdcage coil resonances at ω_o and thus achieve decoupling between the implant and birdcage coil.

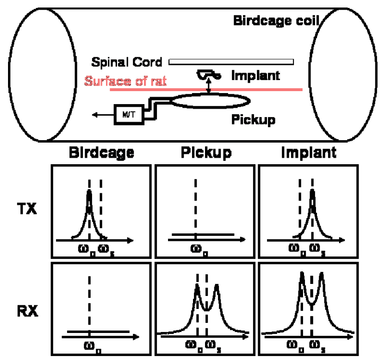

Fig. 1: System schematic

Methods. A rectangular copper implant coil (AWG 20, 11mm wide, 19mm long) with curved arches coated with polyolefin heatshrink and medical elastomer (Dow-Corning MDX-4210) was tuned with a chip capacitor to 347 MHz in air. The implant was surgically placed on the T9/T10 spinal level of a 250g female Sprague-Dawley rat; 3 days after implantation the resonant frequency ω_s dropped to 317 MHz. A 3.4cm diameter pickup surface coil (with diode-activated decoupling LC-tank circuit) was tuned to the same ω_s and brought close to the implant site, so that coupling produced resonances at 300 MHz and 332 MHz in the implant and pickup. A network analyzer was used to perform S21 measurements to determine the coupling between coil elements. Two imaging sessions on the isoflurane-anesthetized rat were performed on a 7T 30cm-bore Bruker MRI: first with the "non-decoupled" implant-pickup system alone used for both transmission and reception, and then with the "decoupled" implant-pickup system used for reception and a Bruker quadrature birdcage coil (7cm i.d.) used for transmission (the birdcage coil was rotated to maximize geometric decoupling). For each case, an *in vivo* flip angle correction map [5] was calculated in the transverse plane of the spinal cord, which describes the ratio between the actual flip angle and scanner-prescribed flip angle (data generated by 3D FLASH, nominal α=145°, 180°, 215°, TE/TR=2.7/400ms, FOV=5.12x2.56x2.4cm, matrix=36x128x16, zero-filled to voxel size of 0.2x0.2x1.5mm). Spin echo RARE images (TE/TR=11.8/4000ms, FOV=5.12x2.56cm, matrix=256x128, slice thickness=1.5mm, RARE factor=8) were acquired to compare SNR between the two cases, with transmitter power recalibrated (through knowledge of the flip angle correction factor) to ensure that the flip angles in the spinal cord were 90° and 180°. Absolute flip angle maps were generated by multiplying the correction factor maps by 90 (flip angles < 80° and > 110° have been set to the gre scale table's extremes to improve visualization). For the decoupled system, a RARE image was also acquired with the birdcage coil used for both transmit and receive, with the implant-pickup coil system remaining in the decoupled state.

Results. In transmit mode, S21 isolation between the decoupled pickup coil and birdcage coil is -17 dB (both channels). The birdcage coil TX/RX image (Figure 2) demonstrates that the isolation between the birdcage and implant coil is good but not entirely complete (birdcage SNR in spinal cord=28). In receive mode, S21 isolation between the active implant-pickup system and decoupled birdcage coil channels was -11.3 dB and -22.7 dB. The flip angle maps shown in Figure 3 are corrupted by low SNR and aliasing artifacts as well as in regions where the flip angle is far away from the prescribed value; however, the technique produces valid results within a limited region around the spinal cord The "decoupled" setup is more uniform than for the "non-decoupled" setup in the region of the spinal cord (standard deviation/mean of correction factor = 1.4% and 3.3%, respectively), and also produces less flip angle nutations around the coil wires. The RARE image using the actively decoupled system (SNR=79) is shown in Figure 2, and had comparable SNR to the transmit/receive image of the implant and pickup alone (SNR=81).

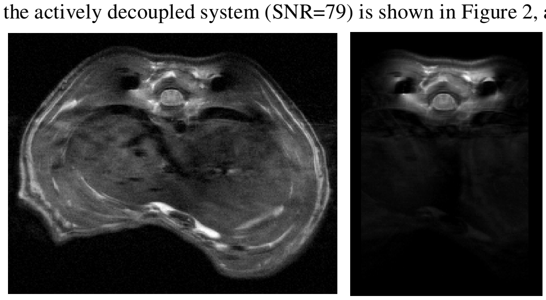

Figure 2: Spin echo image using birdcage TX/RX (left) and birdcage TX/implant-pickup RX (right)

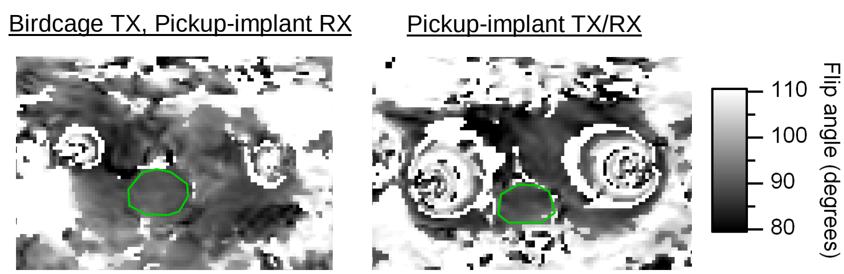

Figure 3: Axial flip angle maps for birdcage TX/pickup-implant RX (left) and pickup-implant TX/RX (right) in vicinity of spinal cord (outlined in green)

Discussion and Conclusions. The most critical design choice in this coil system was the choice of implant resonant frequency (317 MHz): the implant's self-resonant peak must be as far away from 300 MHz as possible to minimize coupling with the birdcage during transmit. However, the resonant frequency must be low enough so that the achievable coupling between pickup and implant can still produce a lower coupled resonance at 300 MHz during receive. It is evident that the coupling between implant and birdcage coil was not entirely removed; however, this is not surprising since even a nonresonant short-circuited coil will still couple to a resonant loop [6]. However, the local amplification of the transmit B1 field does not produce significant inhomogeneity within the region of interest below the coil, and may even help to reduce power requirements for sequences requiring short RF pulses (e.g. quantitative T2 mapping, as in [2]). In summary, we have demonstrated an implanted coil system which successfully improves B1 field homogeneity while achieving high SNR within the localized region around the rat spinal cord Such a system can be applied to applications where implanted coils are warranted.

Acknowledgments. This work has been supported by the Natural Sciences and Engineering Research Council of Canada and the Canadian Institutes of Health Research. **References**: [1] Silver et al., MRM 2001, 46:1216-22. [2] Yung et al, ISMRM 2004, 1537. [3] Wirth et al., MRM 1993, 30:626-33. [4] Smith et al., ISMRM 2005, p.323. [5] Dowell et al., MRM 2007, 58:622-30. [6] Mispelter et al., NMR Probeheads in Biophysical and Biomedical Experiments, 2006, p.237.