Dual-tuned strip-line loop array H1 / birdcage Na23 RF coil for 3T MRI

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Purpose: MRI-observable nuclei other than hydrogen are not or less abundant in living tissues, images of those nuclei suffer from a lower SNR. Imaging such nuclei in humans or animals with sufficient SNR requires an RF coil design that has high sensitivity. The birdcage coil with trap circuits, containing two quadrature outputs for hydrogen and two quadrature outputs for another nucleus, has widely been used[1]. However, this design generates an inhomogeneous $\mathbf{B_1}^-$ field for hydrogen MRI at 3T and suffers from low SNR due to the high resistance of air core inductors used in the trap circuit at the lower resonance frequency. To resolve these critical issues, the combination of a strip-line array (for hydrogen) and a birdcage coil (for other nuclei) was proposed for high field MRI. This RF coil for 7T showed minimum interaction between the strip-line array and the birdcage coil, as well as relatively low inter-element coupling of the strip-lines positioned between the legs of birdcage coil[2]. However, the strip-line coil array generated a less uniform $\mathbf{B_1}^-$ field than a rectangular-shaped loop array at 3T [3]. In this application, we propose a dual-tuned RF coil using and array of rectangular-shaped loops for H^1 imaging and a birdcage coil for Na^{23} imaging. The constructed dual-tuned coil has been evaluated by experimental measurements and MR imaging.

Method: A conventional high pass birdcage coil with eight legs was first constructed. It was a transmit/receive (T/R) coil driven in quadrature. Distributed capacitors on the end-rings were tuned at 33.81MHz. The optimum position for adding rectangular-shaped loop coils onto the birdcage coil for minimizing mutual coupling between adjacent coils was then investigated. The frequency variance for different locations of the loop coils was recorded and the optimized $\mathbf{B_1}^-$ field sensitivity was investigated through MR imaging with phantoms. To decouple neighboring coils, we investigated the decoupling loop method proposed by Ye Li et al[4], as well as an inductive decoupling method where the air inductors were cross-connected to each coil loop[5]. To construct a decoupling loop, a copper strip line (width: 2mm) was fixed adjacent to a leg of the birdcage coil. The strip line was soldered to one side of the leg and a tuning capacitor was soldered between the strip line and other side of the leg to create a loop. An array of eight receive-only loop coils was fixed onto the birdcage coil at the optimized location using the improved decoupling method (see results). The isolation between resonance frequencies of two different nuclei was also investigated. MR images were acquired on the 3T Philips Achieva system (Philips Medical Systems, Netherlands). A cylindrical shaped resolution phantom containing 80 mM CuSO4 and 0.5g/100ml NaCl was used for phantom H^1 imaging. For invivo imaging, hydrogen images were obtained using the following acquisition parameters: T1-weighted spin echo: TR/TE= 600ms/10ms, FOV=250mmx198mm, recon matrix = 560x560, slice thickness =4mm, NSA=2, scan time = 10:39. T2-weighted fast spin echo: TR/TE= 3000ms/80ms, FOV=250mmx201mm, recon matrix = 560x560, slice thickness =4mm, NSA=2, scan time = 07:24. Sodium imaging parameters were the following: TR/TE = 100ms/0.13ms, FOV =250mmx250mm, recon matrix = 128x128, flip angle = 60°, thickness = 5mm, NSA=12, scan time = 16:12.

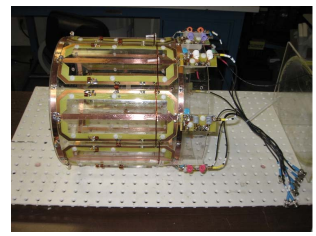

Figure 1. Prototype of the dual-tuned coil.

Results: The rectangular loops were first positioned within the windows of the birdcage coil = (fig. 2a-b). The resonance frequency of the loop coils increased 22.3MHz from the original tuning of 127 MHz, which caused a drop in the Q-factor (186 => 142). However, when the loop coils were positioned over the birdcage coil legs (fig. 2c), the drop in Q-factor was reduced (186 => 175). Both decoupling methods generated well-defined image profiles without mutual coupling. However, positioning of the coil loops over the birdcage coil legs using the inductive decoupling method produced a more sensitive $\mathbf{B_1}^-$ field seen in the MR phantom image than other configurations (fig. 2). Using the optimum configuration, the isolations were -22dB (at the hydrogen coil) and -16dB (at the sodium coil) (fig. 3a). We acquired in-vivo H^1/Na^{23} images from the surface coil array and birdcage coil successfully (figure 3b-d).

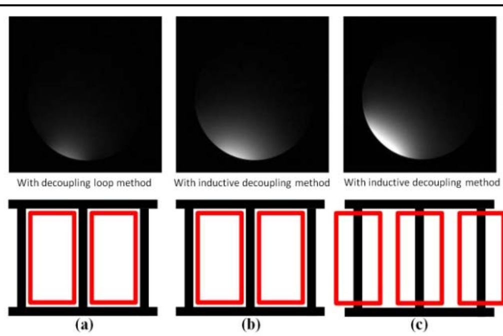

Figure 2. MR phantom images using the array of rectangular loop coils with (a) the coils positioned within the birdcage windows using the decoupling loop method, (b) the coils positioned within the birdcage windows using the inductive decoupling method, and (c) the coils positioned over the birdcage legs using the inductive decoupling method. All images were scaled using identical window levels.

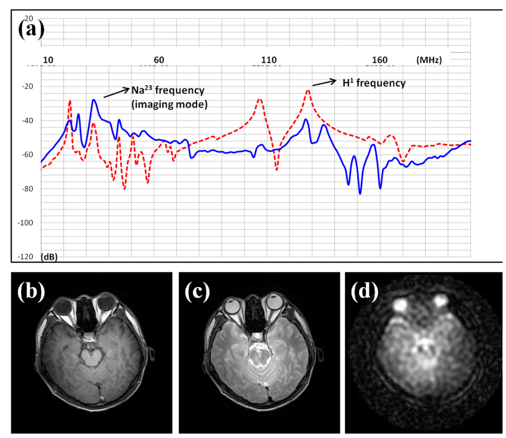

Figure 3 (a) Response spectrum from the separate coil outputs (red dot: one rectangular loop coil tuned to 127MHz, blue solid: birdcage coil tuned to 33.81MHz). (b) T1-weighted H^1 image. (c) T2-weighted H^1 image (d) Na^{23} image.

Discussion: We presented the design, construction, and testing of a new dual-tuned RF coil. The frequency of the loop coils when positioned over the birdcage coil legs was tuned efficiently without reducing the capacitance value and the inductive decoupling method, at the point of $\mathbf{B_1}^-$ field sensitivity generated from RF coil, was superior to adding decoupling loops. Although the isolation (-16dB) between the two nuclei components is not by itself sufficient, it is acceptable since the array of hydrogen coils are detuned by a bias current during sodium imaging. Our results demonstrate that the dual-tuned RF coil design of combined birdcage coil and array of rectangular-shaped loop coils generates a homogenous $\mathbf{B_1}^-$ field at both frequencies. Along with the dual-tuned array coil combining strip lines and a birdcage coil, our proposed design provides a feasible alternative to the standard trap-based method for dual-frequency imaging.

References: 1. Shen GX, *et al.* Magn. Reson. Med. 1999; **41** 268-75. 2. Wiggins GC, et al. Proc. ISMRM. 2010;pp 1500. 3. Alagappan V, *et al.* Proc. ISMRM. 2007;pp 165. 4. Ye L, *et al.* Med. Phys. 2011; **38**, 4086-93. 5. Nabeshima T, et al *U.S. patent* 1996; 489,847.

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