A Millimeter Scale Implanted Coil with an Integrated Wireless Amplifier for Imaging of the Rodent Kidney

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Motivation

It has long been a practice in the magnetic resonance community to use small internal radiofrequency coils for localized spectroscopy and imaging [1,2], and there continues to be active development of catheter coils for interventional MRI [3]. In order to detect the signals from a small isolated coil, it has been common to run a hard wired connection from the coil to the preamplifier sometimes over long distances leading to potential RF heating. Mutual inductive coupling between an internal coil and an external pick up loop has been utilized to eliminate the wired connection; however, sufficient sensitivity can be maintained only if the distance separation between the coupled coils is small. Recently, it was demonstrated that the detection sensitivity of remotely coupled coils can be enhanced when the locally detected MRI signal was amplified by an integrated wireless parametric amplifier before being inductively coupled to the external pick up loop [4]. In this work, a millimeter scale implantable coil integrated with a wireless parametric amplifier was constructed. This coil could be implanted on the kidney of a rodent to enable high resolution images to be obtained with excellent sensitivity. The implanted wireless coil enabled the identification of micro vascular structures of a rat's kidney *in vivo*.

Method

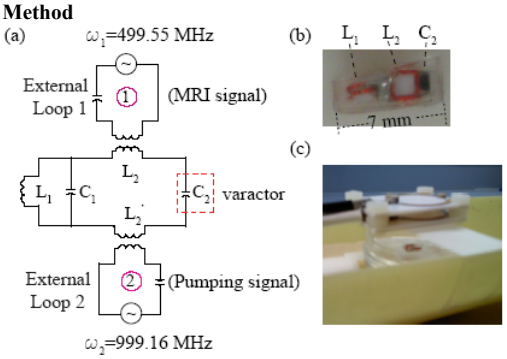

Fig. 1. (a) The circuit diagram of the implanted wireless parametric resonator (center) and the external coupling loops (bottom and top). (b) The enlarged picture of the parametric resonator with PDMS coating. (c) The layout diagram for testing the resonator on a gel phantom.

To build a compact implantable coil with an integrated parametric amplifier, a double frequency resonator was constructed as illustrated in Fig. 1a. Two *L-C* meshes were assembled with one resonance at 496 MHz (Q=99) and the other resonance at 1004 MHz (Q=110). The lower resonance is for MRI signal detection at 499.55 MHz (Avance 3, Bruker Inc, Billerica, MA) in an 11.73 T magnet (Magnex Inc, Oxford, UK). The higher resonance was for wireless reception of the pumping power at 999.16 MHz from outside the animal. The pumping frequency was chosen to be 60 kHz above twice the value of the MRI carrier frequency so that the up-converted outputs at the difference frequency have no overlap with the original signals centered around the carrier frequency, assuming an imaging bandwidth of 50 kHz. C_2 was a varactor (Infineon, Neubiberg, Germany) which performed frequency mixing with parametric gain. Fig. 1b is the enlarged view of the resonator that had a dimension of 7 x 3 x 3 mm^3. L_1 and L_2 were oriented perpendicular to each other so that the resonator could couple effectively to nuclear spins when placed inside the magnet at different orientations. There were two single resonance loops inductively coupled to the double frequency resonator at each frequency. Loop 1 resonated at 499.55 MHz to receive the amplified MRI signals; loop 2 resonated at 999.16 MHz to provide the pumping power. To test the resonator performance, the resonator was placed on the surface of an 1% (w/v) agarose gel, and the two coupling loops were placed at a 18 mm distance separation above the gel's surface, with the signal reception loop beneath the pumping loop (Fig. 1C). The required pumping power for oscillation was empirically determined by observing the onset of oscillation at one half of the pumping frequency. MRI experiments were subsequently performed at a power level that was 0.4 dB below the oscillation threshold.

The resonator was implanted on the surface of a rat kidney. The kidney was pulled out of the abdominal cavity through an incision before the resonator was glued to its medial lateral surface. The glue was made by mixing 10% (w/v) gelatin solution with 1% glutaraldehyde. The kidney was then returned to the abdominal cavity before the incision was closed with absorbable suture. After several days of recovery, the rat was anesthetized with 2% isoflurane for imaging. 2D FLASH images were acquired at TE = 5.2 ms, TR = 20 ms, FA = 8 deg, with retrospective gating for motion correction. A lower resolution image (156 x 156 x 1000 mm^3) was first acquired without parametric amplification followed by a higher resolution image (50 x 50 x 280 mm^3) obtained in the presence of parametric amplification.

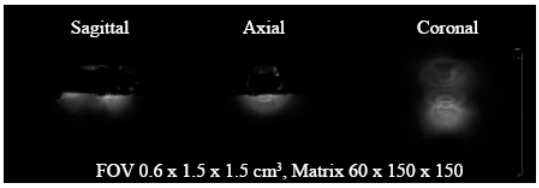

Fig. 2. A 3D FLASH image of the 1% agarose gel obtained by the parametric resonator with 100 μm isotropic resolution.

Result

Fig. 2 shows three orthogonal cross sections of a 3D FLASH image obtained with the parametric resonator located on the surface of the agarose gel. Due to the restricted size of the resonator, the images had sensitive regions close to the resonator's boundary. In the sagittal and axial images, an effective penetration depth of 2.4 mm was observed, which was approximately equal to diameter of the detection inductor L_2. In the coronal slice, the bottom region had higher intensity than the top region, because the bottom portion of image was obtained by L_2 that was placed parallel to the gel's surface while the top portion of image was obtained by L_1 that was placed perpendicular to the gel's surface. The parametric resonator retained 70% sensitivity compared to a direct connected coil with similar size, but it still had a 7 times better localized sensitivity compared to the large external surface coil that was able to detect the entire kidney.

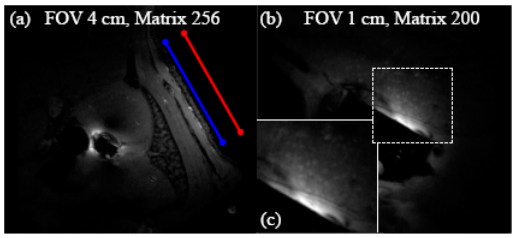

Fig. 3. (a) A low resolution FLASH image obtained without parametric amplification. The blue line represents a 22 mm diameter signal reception loop, and the red line represents a 22 mm diameter pumping loop that was placed 4 mm away. (b) A high resolution FLASH image obtained with parametric amplification. (c) The enlarged view of the image region enclosed by the dashed box.

Fig. 3 shows 2D FLASH images obtained with the implanted resonator on the kidney. Fig. 3a shows the lower resolution image (156 x 156 x 1000 mm^3) acquired without parametric amplification. Here the image is formed primarily from the larger loop external to the animal. The implanted resonator can be clearly located. Fig 3B shows the higher resolution image (50 x 50 x 280 mm^3) obtained in the presence of parametric amplification. Micro vascular structures show up as bright dots in the high resolution kidney image due to blood flow in these T_1 weighted images (see Fig. 3C for enlarged view).

Conclusion

A millimeter scale, integrated detection coil with parametric amplifier has been implanted inside the rodent body to enable high resolution imaging of the kidney. The increased sensitivity enabled ready detection of vascular structures in the kidney. This detection scheme is very suitable for implanted and catheter MRI coils where a wired connection is unfavorable or impossible. Future work includes the construction of parametric resonators with smaller dimension and better B_1 homogeneity. The use of chronically implanted, wireless amplified resonators should enable MRI at high resolution to follow a number of pathological processes.

Reference

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