An Array RF Coil for Rabbit Thoracic Arteries Imaging at 200MHz

R. P. Joensuu[1]

[1]AstraZeneca, Molndal, Sweden

Introduction

Parallel imaging is becoming standard in clinical systems but it still relatively uncommon in animal MRI studies. Most of the animal coil array designs have so focused on mouse and rat imaging applications, however larger animal models such as rabbits are also widely used in research, for example in vessel wall studies. The purpose of this work was to design a 4-channel array coil for rabbit's thoracic arteries imaging at 4.7T.

Methods

The coil elements were made of a semi-rigid coaxial cable (EZ141, Huber&Suhner, Switzerland) with a loop diameter of 45 mm. The loops were cut into two halves where in one end the outer shields were connected with a tunable capacitor (C_1) and on the other end with a DC-block capacitor (C_2) in series with a λ/4 –coaxial wire (EZ47, Huber&Suhner, Switzerland), the intrinsic capacitance of which (≈ 1pF/cm) constituted a part ($C_{\lambda/4}$) of the resonance circuit's total capacitance (Fig. 1). The λ/4 –cable also transformed the high impedance at the preamplifier's input to a low impedance at the coil end, thus lowering the coil's Q-value. Lowering the Q-value reduces the mutual resonance effects (resonance splitting and coherent noise) which are impossible to control by geometrical means alone if the number of elements exceeds 3. Sheet currents, induced both by the gradient and RF excitation fields, can couple to looping wires affecting the system's performance and potentially damaging its components. The cables can also scatter the incoming radiation disturbing the RF irradiation pattern at the site of excitation. Here this problem was tackled by feeding the Rx/Tx control current in a shielded manner throughout by using the center conductor of the coaxial loops and the λ/4 –coaxial cable as a transmission path. Thus no separate Rx/Tx control wires were used and the existing wires were minimally exposed to external fields. The outer shields of the λ/4 –cables were also covered by a resistive carbon coating which increases the cable's surface impedance and reduces the sheet current resistively, ultimately transforming it to heat.

Decoupling was performed by three PIN-diodes (D_1, D_2, and D_3) which were connected in parallel. In the reception mode the diodes are reverse-biased and the NMR current is flowing only on the outer shield of the coaxial loop (L_{out}) which together with the capacitors C_1, C_2 and $C_{\lambda/4}$ form the receiving resonant circuit. The NMR signal is not measured over any lumped capacitor but across the λ/4 –cable's center conductor and outer shield at the preamplifier's end (Fig. 2). During RF transmission the diodes are shorted. D_3 shorts the preamplifier's input to ground protecting it but it also raises the λ/4 –cable's impedance to high at the coil's end. This is counteracted by incorporating D_2 into the circuit which shorts $C_{\lambda/4}$ thus detuning the coil. Shorting the diode D_1 incorporates the center conductor into the circuit. It can be modeled as an additional inductance (L_{in}) coupled either in series or in parallel to the inductance formed by the outer conductor (L_{out}) (Fig. 2). The effective inductance will then change from L_{out} to: $L_{eff} = L_{out} + L_{in} + 2k\sqrt{L_{in} \cdot L_{out}}$ or $L_{eff} = 1/(1/L_{out} + 1/L_{in})$, respectively, which detunes the coil.

Receive and transmit mode behavior was tested by opening (Rx) and shorting (Tx) the diode D_1 and monitoring the resonant frequency. A 10pF test capacitor was used in C_1, C2 was shorted, and D_2 was replaced by a 20pF capacitor, and in the theoretical resonance calculations the measured inductance values were used (L_{out} = 115 nH, L_{in} = 86 nH). Finally, a four-element coil was constructed where the loop elements were overlapped geometrically to minimize the mutual inductance.

Results and Discussion

The coil element's unloaded Q-value was 330 without and 30 with the λ/4 –line, and the measured and simulated resonance frequencies are listed in Table 1. The measured Rx/Tx isolation was better than 60dB in all channels. No change was observed in the coils' noise level with any RF excitation or gradient power levels tested. Test images were acquired in a Bruker BioSpec 47/40 MR scanner by using a Bruker i.d. 197 mm quadrature birdcage coil for excitation (Fig. 3). No artefacts resulting from the system could be observed in the images. The resistive shielding in particular improved the coil's performance. This can be a useful tool when the number of channels is increased as scattering from multiple cables easily becomes a serious issue. As a conclusion, a multichannel coil has been constucted which is very well shielded from external disturbances.

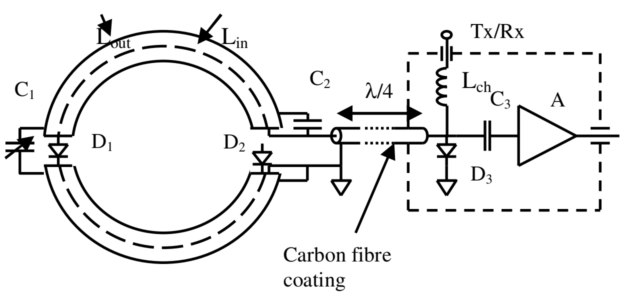

Fig. 1. The schematics of an individual coil element.

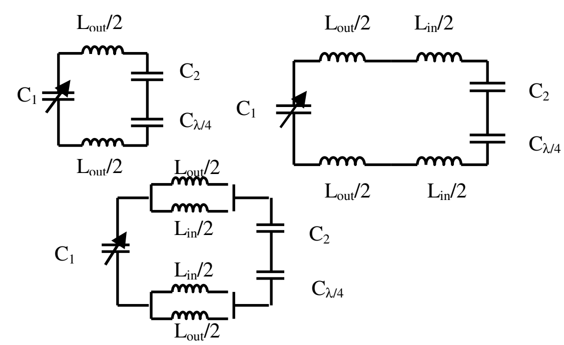

Fig. 2. The equivalent circuits of the coil element. Top left: Receive mode (D_1 open), only outer inductance is active. Top right: Transmit mode 1 (D_1 shorted), inner and outer inductances are in series. Bottom: Transmit mode 2 (D_1 shorted), inner and outer inductances are in parallel.

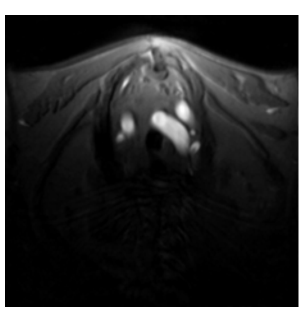

Fig. 3. An image of a rabbit's thoracic arteries acquired with the array coil. IG-Flash-9/4.5; FOV 79x79 (256x128); sl. 3.2mm; PPI Acc=2.

Table 1. Measured and simulated resonance frequencies.

D1	Mode	L_{out}, L_{in}	f_res [MHz]		Error
			Measured	Calculated	[%]
Open	Rx	L_{out} only	213.3	211.0	-1.1
Short	Tx	in parallel	97.7	97.7	0.0
Short	Tx	in series	280.0	278.1	-0.7