The design of dedicated RF coils for mouse MR

Dennis W. J. Klomp and Peter R. Luijten
University Medical Center Utrecht, the Netherlands

Introduction

MR imaging of mice in medical research has allowed for well controlled disease models for pharmokinetic and dynamic tests of new contrast and therapeutic agents. However, the volume of interest in mice is typically several orders of magnitude smaller than in humans, and concomitantly the number of spins that contribute to the signal in an MR exam will be substantially smaller, thereby potentially degrading SNR. Fortunately a large part of the SNR degradation can be regained using RF coils that are in close proximity to the anatomy of the organ of interest. Although some general purpose RF coils either come with the MR system or can be commercially purchased, many applications substantially benefit from the use of dedicated RF coils.

In this course, the basic steps will be presented required for the design, construction and use of such dedicated RF coils. The course will start with a brief theory in RF electronics and practical coil concepts in terms of transmission and reception. Additionally, receive only concept, multi array and multi nuclear applications will be discussed as well.

Theory

Coil Efficiency

The number one determinant in MR technology for optimal SNR is the RF coil that can be used to transmit RF for excitation and receives the MR signals. The SNR obtained with a receiver coil is proportional to the efficiency of the coil, which according to the principle of reciprocity [1] is equal to the *efficiency* of the coil as a transmitter. The efficiency of an RF coil is expressed as the magnetic field strength per unit of applied RF power as a quadratic relation (in T/√W).

Circularly polarized fields and B_1 field strength

A magnetic field can be generated by an electrical current flow (I) through a conductor. Per unit current the strength and orientation of this field in the object that is measured can be approximated using the law of Biot-Savart (as long as the wavelength of the RF is larger than the size of the object, which is the case for mouse MR at field strength up to 20T [2]):

$$\vec{B}(\vec{r}) = \frac{\mu_0 I}{4\pi} \int_{conductor} \frac{dl \times \hat{r}}{|\vec{r}|^2}, \qquad (1)$$

where dl is the length of a part of the conductor, r is a vector pointing from the location of the conductor part to the location of the point in space from where the magnetic field strength is calculated, and μ_0 is the magnetic permeability constant in vacuum (i.e. $4\pi \times 10^{-7}$). In order to create a B_1 field that rotates around the main magnetic field, a second conductor is required that generates an equal but orthogonal field with respect to the first conductor. When the electrical current through the orthogonal conductors oscillates with a frequency equal to the Larmor precession, with a 90º-phase difference, a rotating magnetic field (circularly polarized) is generated (Fig. 1). Compared to a single coil setup (linearly polarized field) this quadrature setup requires twice as less power for the same B_1, hence improves SNR by 41% (√2).

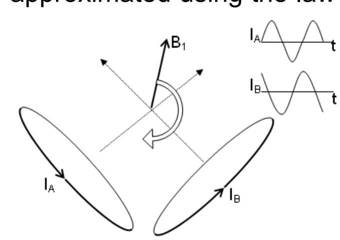

Figure 1: Graphical presentation of the creation of a circularly polarized magnetic field (B_1), using two (A and B) orthogonally positioned coils in which a sinusoidal current (I) flows with a 90° phase difference.

Tuning and Matching

An optimal transformation of RF power from an RF amplifier into current through the conductors is obtained when the conjugated impedance (Z*) of the amplifier matches the transformed impedance (Z) of the conductor. The impedance of the conductor can be split into a real part that absorbs power (i.e. resistance R) and an imaginary part that can temporarily store and release power (i.e. admittance X). The resistance of the conductor that generates a magnetic field in a sample consists of three parts: one part

(R_L) that depends on the conductivity, length and cross-section of the conductor, including skin effects at high operating frequencies; one part (R_R) is related to radiation losses, which relates to the operating frequency, size of the conductor and the electro magnetic properties of the surroundings; and finally a part (R_S) that is related to the relative power absorbed due to eddy-currents and electric fields in conductive tissue. The admittance of the conductor is linearly proportional to the frequency of operation and the inductance (L) of the conductor, where L can be calculated as:

$$L = \frac{\mu_0}{4\pi} \int_{surface} \int_{coil} \frac{dl \times \hat{r}}{|\vec{r}|^2} dA , \qquad (2)$$

where the conductor encloses a loop (coil) with a surface area A. In total, the impedance of the conductor is: $Z=R_L+R_R+R_S+j\omega L$, where $j=\sqrt{-1}$. This impedance needs to be transformed to the impedance of the RF amplifier (typically 50Ω), which can be realized by a simple capacitive network. A parallel capacitor (C_t) can be connected to both ends of the conductor, which creates a parallel impedance Z_p:

$$Z_p = \frac{R_L + R_R + R_S + j\varpi L}{j\varpi C(R_L + R_R + R_S + j\varpi L) + 1} . \qquad (3)$$

In case the real part of Z_p is tuned to 50Ω (using the appropriate value for C_t), the imaginary part of Z_p can be eliminated by adding a capacitor (C_m) in series with Z_p with admittance equal to the negative admittance of Z_p. In this case the impedance of the inductor parallel to C_t and in series with C_m matches the 50Ω of the RF amplifier (Fig. 2).

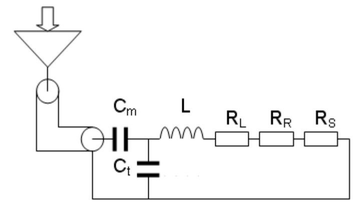

Figure 2: The conductors of the RF coil have a net inductance (L) and resistive losses (R), which are matched by capacitors (C) to the impedance of the RF amplifier.

As all RF power (P) from the amplifier is absorbed by the matched conductor (i.e. $R_R+R_L+R_S$), the current can be calculated using the formula:

$$I = \sqrt{\frac{P}{R_R + R_L + R_S}} . \qquad (4)$$

So optimization of efficiency can be realized by minimizing the resistance of the conductor and maximizing the magnetic field created by the conductor. In mouse MR imaging, radiation losses (R_R) are small, but can be further reduced using an RF screen. The conductor resistance (R_L) can be reduced by using better conductive materials, by cooling the conductor or even using a superconductive setup. Resistive losses in the sample (R_S) cannot be reduced for a given frequency and conductor geometry with respect to the living mouse. Nevertheless, in an RF coil that is smaller than the sample, the relative resistive sample losses are smaller than in a coil that covers the entire sample. Therefore if the smaller coil is positioned close to the region of interest in the mouse, the efficiency can be substantially improved compared to a larger coil. However, once the coil losses (R_L) dominate, reduction in coil sizes may no longer improve SNR.

Coil testing

Although the formulas presented here may be used to understand the electronics of RF coils, the fact that most values can actually be measured makes optimization of the quality of the coil construction a powerful method. The B_1 field can be measured using the MR system by adjusting the integral of an RF pulse up to a 90 or 180 degree pulse for the region of interest. For instance if a 50μs rectangular RF pulse leads to a 90 degree pulse of the ^1H spins, than the B_1 field would be $(4x50\mu s x42.6MHz/T)^{-1}$. The power (P) that was needed to create this RF pulse is given by the MR system either directly in Watts, or as an attenuation factor (att) in dB ($P = P_{max}x10^{att/10}$, where P_{max} is the maximum peak power of the RF amplifier) or in Voltage (U) ($P = U^2/50$). The values for the capacitors can be determined by adjusting the resonance frequency of the coil setup, which can be determined by measuring the reflected RF power as a function of frequency (reflection curve). Generally this can be done with the MR system during a coil-tuning procedure, using a network-analyzer or with a low cost handheld RF sweeper. Matching to the system impedance is obtained when the measured reflected RF power is zero at the Larmor frequency (f_0). The self inductance (L) of the coil can be approximated by $L = 1/C_t(2\pi f_0)^2$. The values for the resistances in the coil can be indirectly determined by measuring the quality (Q) factor of the coil. In a well matched condition, the Q value is inverse related to the bandwidth of the reflection curve ($Q = 2f_0/\Delta f_{-3dB}$), where Δf_{-3dB} is the difference between the two frequencies for which half of the RF power is reflected. Assuming R_R to be negligible, R_L+R_S can be determined as $R_L+R_S = 2\pi f_0 L/Q_{loaded}$, where Q_{loaded} is the determined Q factor of the coil loaded with the mouse in place. If the range of capacitances is sufficient to also tune and match the coil in an unloaded situation, R_L can be determined as $R_L = 2\pi f_0 L/Q_{unloaded}$.

Under the assumption that different types of coils are all matched to 50Ω, the detected noise (N) from this 50Ω point no longer depends on the type of RF coil as it is related to:

$$N = \sqrt{4kT50\Delta f}\ , \qquad (5)$$

where Δf is the receiver bandwidth. This means that the SNR obtainable by the coil is linearly related to the efficiency of the coil measured at this 50Ω point, enabling comparisons of different coil types with respect to SNR.

Practical coil setups

Transmit receive coils

The design indicated in figure 2 can be used for all sorts of RF coils. In case small surface coils are used, the inhomogeneous B_1 fields that come with these coils must be taken into consideration. For sequences that require homogeneous B_1 fields, volume resonators can be used. Although solenoid RF coils can create homogeneous transverse magnetic fields for some configurations (figure 3), mostly TEM resonators or birdcage coils [3] with many rods and distributed capacitances are used to create a homogeneous B_1 field. Practical software tools are available for free to calculate the values for the capacitances for any chosen geometry and frequency. However, in case the volume coil will be used for large differences in tissue loads, the Litze coil design may be more practical, since the tuning and matching of such coil is requires substantially less variable capacitors [4].

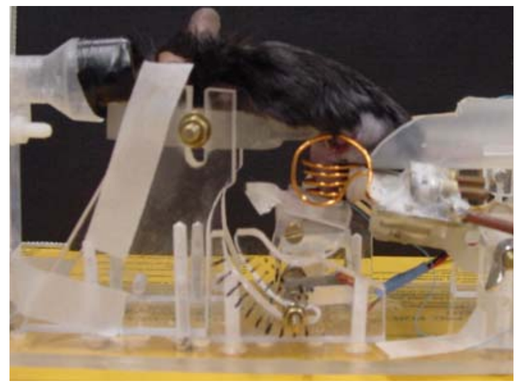

Figure 3: Coil setup of a solenoid coil orthogonally aligned to a surface coil for calf muscle of mice.

Receive only coils

A more elegant way of obtaining a uniform excitation field at high SNR, is to use a receiver (Rx) coil in combination with a volume transmit (Tx) coil. Additional circuitry is essential that prevents RF power coupling from the Tx coil to the Rx coil and visa versa. Electrical RF switches [5] can be used in series with the inductor that prevent current flow in the Rx coil during the transmit phase, or in the Tx coil during the Rx phase of the MR sequence. In general PIN diodes are used that act as a short when a bias DC current is set through the diode or as an open when a reversed DC voltage is applied to the diode. The PIN diode is used for is relatively slow switching time, thereby not changing state during the high RF voltages of the transmit phase. In addition, the capacitance is lower in comparison to PIN diodes which result in less RF leakage in the open state of the PIN diode. However, if the PIN diode is in the short state, the resistive losses degrade the Q factor of the coil. These losses can be prevented by using a circuitry as indicated in figure 4. The additional inductor (L_2) in series with a second tuning capacitor (C_2) is on resonance, which as L_2 is not loaded by tissue has a high Q value and therefore acts as a high impedance (open) when the PIN diode is DC biased (short). Large inductances can be used to guide the DC to the PIN diodes, which prevent RF to follow the DC track, and large capacitances can be used to block the DC. The DC lines are available from the MR system.

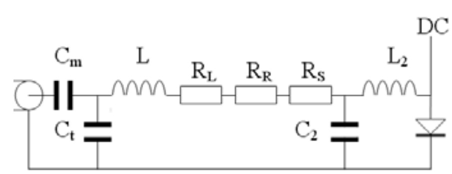

Figure 4: Detuning circuit (L_2, C_2) that opens the coil when a DC current is biased to the PIN diode.

Multi receiver coils

When coil losses dominate over tissue losses, which is the case in practically all mouse imaging (except for whole mouse imaging at ultra high fields), the use of multiple receiver coils will not improve the intrinsic SNR, but may be used to speed up image acquisition. Like in Tx only and Rx only coil setups, the multiple coils need to be decoupled from each other. However, RF switches can not be used for this application, as all multiple Rx coils need to be in operation at the same time. Coil overlapping can be used to minimize mutual inductance and thereby reduce RF coupling, however resulting in less effective G-factors in parallel imaging [6]. A more elegant way is to integrate low-impedance preamplifiers in the Rx coils. The difference between noise-matched (generally 50 Ω) and power matched impedance (almost a short) of these preamplifiers can be used to create a high impedance (open) in the coil [7] by carefully selecting the correct ratio of L_2 and C_2 to match the coil to 50 Ω and in the meanwhile creating a resonance at the Larmor frequency between these two components and the short of the preamplifier (figure 5).

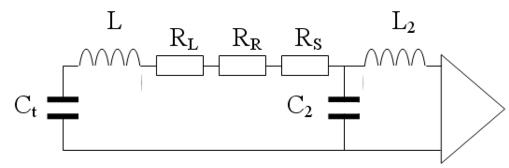

Figure 5: Low impedance preamplifier that opens the coil due to parallel resonant circuit (L_2, C_2).

Multi nuclear coils

In multi nuclear experiments, practically always a ^1H coil setup needs to be present to enable B_0 shimming using the high signal content of water and conventional MRI for localization. In addition, ^1H decoupling and polarization transfer methods may be employed which require coil setups for multiple nuclei. These techniques exclude the ability to use PIN diodes or preamplifiers to decouple the ^1H coils from the other nucleus coil, however, as the frequency of operation is substantially different, RF filters can be used to realize high impedances (open) in the other coil. When tissue losses are not dominant, such additional filters in series with the coil result in a reduction in SNR [8]. An alternative is to use an orthogonal coil arrangement without mutual RF coupling thereby excluding the need for RF filters inside the coil (like indicated in figure 3). In order to prevent the high RF power of spin decoupling during signal reception to saturate or damage the preamplifier, an additional RF filter needs to be inserted between the RF coil and its preamplifier. In addition, the RF noise of the power amplifier used for spin decoupling needs to be suppressed as well in the frequency range of signal reception, which can be realized using an RF filter between the RF power amplifier and the spin decoupling coil.

Conclusion

RF coil design for dedicated MR on mice remains in the near field regime of operation, therefore can be modelled relatively easy and can be validated using practical measurements. As coil losses contribute significantly in the sensitivity of the MR setup, dedicated RF coil solutions can improve the quality in mouse imaging substantially and therefore may be worth to invest effort into.

References

1. Hoult DI, Richards RE. Signal-to-noise ratio of nuclear magnetic-resonance experiment. J. Magn. Reson. 1976;24(1):71-85
2. Doty FD, Entzminger G, Kulkarni J, Pamarthy K and Staab JP. Radio frequency coil technology for small-animal MRI. NMR Biomed. 2007; 20: 304–325
3. Tropp J. The theory of the bird-cage resonator. J. Magn. Reson. 1989; 82: 51–62.
4. Doty FD, Entzminger G Jr, Hauck CD. Error-tolerant RF Litz coils for NMR/MRI. J. Magn. Reson. 1999; 140: 17–31.
5. Edelstein WA, Hardy CJ, Mueller OM. Electronic decoupling of surface-coil receivers for NMR imaging and spectroscopy. J Magn Reson 1986; 67: 156–161.
6. Pruessmann KP,Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. Magn Reson Med 1999; 42:952–962.
7. Reykowski A, Wright SM, Porter JR. Design of matching networks for low noise pre-amplifiers. Magn. Reson. Med. 1995; 33:848–852.
8. Fitzsimmons JR, Brooker HR, Beck B. A Comparison of double-tuned surface coils. Magn. Reson. Med. 1989; 10: 302–309.