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98: Practical design of RF transmit and receiver coil arrays

Dr. Dennis Klomp

Practical design of RF transmit and receive arrays

Dennis Klomp, Wouter Koning, Hans Hoogduin, Alexander Raaijmakers, Natalia Petridou, Nico van de Berg, Peter Luijten
Radiology and Radiotherapy, University Medical Center Utrecht, the Netherlands

Introduction

The Radiofrequent (RF) coil is one of the most important components in magnetic resonance. Its function is to manipulate the spins and detect their response. When the RF field of the coil (B1) is placed perpendicular to the main magnetic field (B0) and an oscillating magnetic field is created, spins are perturbed once the frequency of oscillation meets the Larmor frequency of the spins and when the rotation of the B1 field has a component that meets the rotation of the spins (B1+). By driving the RF coil with an RF amplifier, a B1+ field strength can be obtained that manipulate the spins sufficiently within an order of ms. However, the RF power required to generate the B1+ field will be absorbed by the RF coil and the tissue, which can lead to heating and may impose a safety concern that needs to be addressed carefully.

When using the RF coil as a receiver, the oscillating net magnetic flux originating from the relaxing spins can be captured by the coil, which will result in an induced voltage that after amplification (using a preamplifier) can be detected and digitized. The sensitivity of the detection is directly related to the efficiency of the receiver coil and can therefore be optimized for any point of interest in space taking the boundaries of the sample into consideration. For larger regions of interest, sensitivity may be compromised in case of reception with a single coil or remain optimized when an array of receiver coils is used. In fact, as the array of coils may each have a unique spatial dependent sensitivity, their sensitivity profiles can be used to spatially encode the MR signal (SENSE), hence accelerating MR imaging.

Over the last decade, developments in RF system design have contributed substantially to the advances in MRI. Using arrays of receivers made an extended field of view at an increased signal to noise level possible, but also scan accelerations by parallel acquisition brought MRI to a next level. Recently, the RF chain was expanded with multi transmit arrays as well. The increased freedom in the transmit chain opens the way for new features like RF shimming, SAR reduction, inner volume excitation, multi-dimensional RF at accelerated pulse timings and so forth. As a consequence, the number of RF coils in an MR system can be extensive, resulting in a potentially complex RF infrastructure with the coils coupling to the body and the electronics. In this course, focus will be on practical tools that simplify the design of an array into independent RF coils that can be combined without substantial destructive RF interferences.

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).

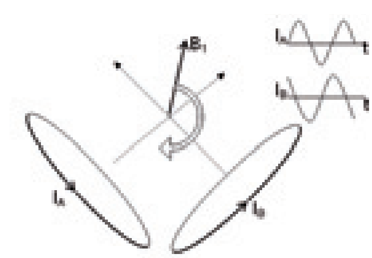

Figure 1: Graphical presentation of the creation of a circularly polarized magnetic field (B_1), using two (A and B) orthogonally positioned coils in which a sinusoidal current (I) flows with a 90° phase difference.

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], or human head MR up to 3T):

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

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 B1 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% ($\sqrt{2}$).

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:

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

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).

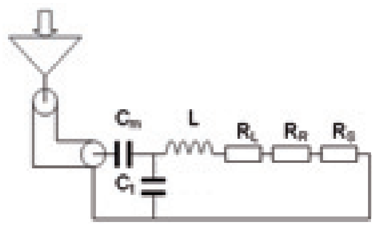

Figure 2: The conductors of the RF coil have a net inductance (L) and resistive losses (R), which are matched by capacitors (C) to the impedance of the RF amplifier.

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. At low frequencies, 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 subject *in vivo*. 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, 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 ^{1}H 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.

RF coil concepts

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 magnetic fields, its orthogonal alignment to the main magnetic field limits its accessibility, therefore mostly TEM resonators or birdcage coils [3] with many rods and distributed capacitances are used to create a homogeneous B_1 field. Practical freeware software tools are available 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 dominated tissue loads, the Litze coil design may be more practical, since the tuning and matching of such coil requires substantially less variable capacitors [4]. Figure 3 displays the most used volume resonators used in MRI.

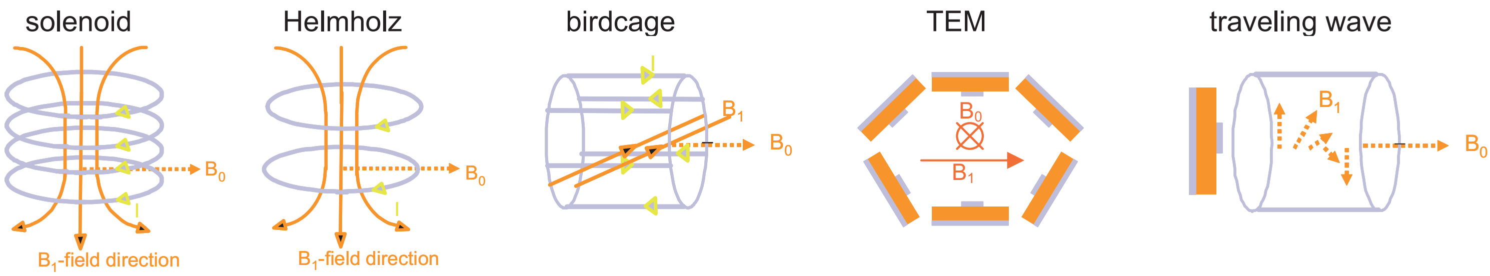

Figure 3: Graphical representation of generally used volume resonators in MRI, including the direction of its B1 field and the orientation with respect to the main magnetic field.

Receive only coils

A more elegant way of obtaining a uniform excitation field at high SNR, is to use a surface receiver (Rx) coil in combination with a volume transmit (Tx) coil. Mostly loop coils are used as receiver elements, but at higher frequencies, stripline resonator and radiative antennas are used as well (Fig 3 & 4).

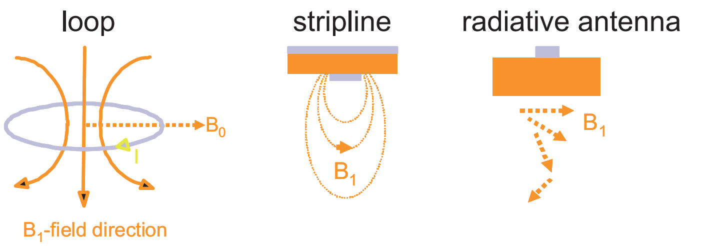

Figure 4: Graphical representation of generally used surface resonators in MRI, including the direction of its B1 field.

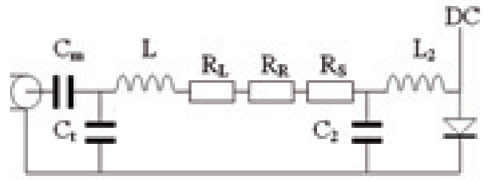

Figure 5: Detuning circuit (L_2, C_2) that opens the coil when a DC current is biased to the PIN diode.

When Rx coils are combined with Tx coils, 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 connection when a reversed DC voltage is applied to the diode. The PIN diode is used for its 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 PN 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 5. 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.

Array coils

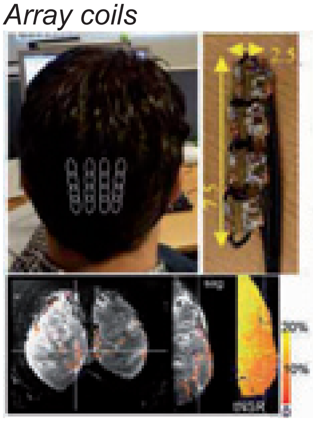

Fig 1: 16 channel surface array enables 0.55mm isotropic fMRI in the human brain at 7T.

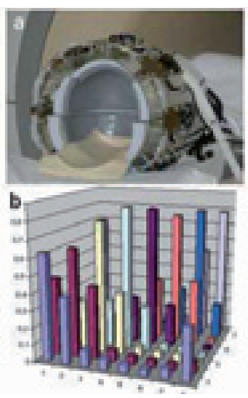

Fig 2: 8ch stripline transmit array with 16ch receiver array (a) with RF coupling detection (b).

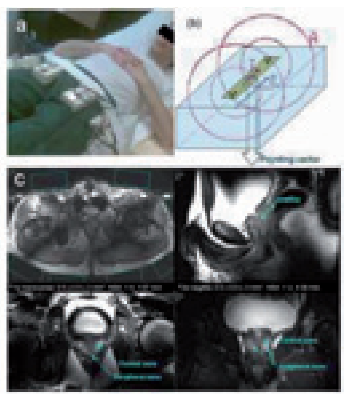

Fig 3: radiative antenna array (a) directs the Poynting vector (b) to the center of the body for efficient MRI of the prostate at 7T (c).

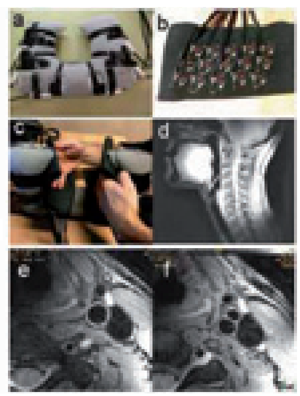

Fig 4: radiative 6ch antenna array (a) with a 30ch receiver array (b) combined (c) to obtain high resolution MRI of the neck (d-f).

RF signal interference between coils relates to the mutual inductance between the coils as well as their quality factor (Q, ratio between coil inductance and coil losses). Although the mutual inductance can be minimized by proper coil alignment (i.e. orthogonal, large distance, or overlap, all to minimize the net flux linkage), the Q factor, which is often dominated by tissue losses, can become low as well. It is demonstrated that loop coils as small as 2cm can couple strongly to the tissue at 7T, preventing substantial coupling between neighboring coils (Fig 6). Although preamplifier decoupling is generally used for receiver arrays, transmit arrays may be actively decoupled using proper power steering from the amplifiers. Control of these amplifiers can be gained by acquisition of pick-up probe signals, which is used for instance in a strongly coupled stripline array (Fig 7). When the wavelength drops below the size of the examined body, even different coil concepts may be considered for steering RF in or out of the body, like radiative antennas (Fig 8) or even combined with separate receiver arrays (Fig 9).

Conclusion

RF coils are key components in determining the quality of MRI. When using switchable volume coils with local switchable surface coils, uniform spin manipulation can be applied at the highest SNR. With a selection of scalable geometries of the coil that consider sample boundaries, any coil can be tuned and matched to the impedance of the MR systems at its operating frequency. Therefore dedicated RF coil setups can maximize the performance of MRI. Using independent transmitters and/or receivers, the design of coil arrays can be practical, opening up a wealth of flexibility in maximizing MRI performance.

References

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