Detection Coils for MRS

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The radio-frequency (RF) transmit and detection coils are transducers that excite and detect the magnetic resonance spectroscopy (MRS) signals. The signal-to-noise ratio, spatial and temporal resolution, spatial uniformity, and the efficiency with which the RF excitation power is used and lost to the measurement sample all depend critically on the RF coil's performance. Therefore, the success of the MRS measurement is determined in large part by the proper choice and application of the RF detection coil. The objective of this review is to provide an overview of the many coil design choices available, and how they can be used. A range of variants of the simple surface coil, concentric and phased-arrays of surface coils, solenoids, saddle coils, crossed-pairs of coils, 'birdcage' coils, transverse electromagnetic field (TEM), and transmission line coils are presented. These span the full range of frequencies and field strengths visited by in vivo MRS and offer options of linear, quadrature, and multinuclear operation. This should help guide the reader on the available options for their desired applications.

Keywords: radiofrequency, RF, radio-frequency coil, RF coil, NMR probe, NMR spectroscopy, MR spectroscopy, MRS coil, MR spectroscopy coil, NMR coil

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Introduction

Radio-frequency (RF) transmit and detection coils for magnetic resonance spectroscopy (MRS) are transducers that inductively couple RF energy to excite and then receive the NMR signal from a desired sample that is typically located adjacent to or within the volume of the coil(s).[1] Because both the excited and received RF signals are magnetic fields that share the same spatial direction and frequency, the transmit and receive coils can be one and the same, or alternatively, two separate coils that are decoupled from each other during transmission and reception. Both configurations are common in MRS. In any case, the coils are placed about a sample, or a sample is placed within the coil(s), both within the homogeneous center of a superconducting magnet whose main field, B_0, is directed along the cylindrical axis, as shown in Figure 1.

An electrical RF detection circuit, sometimes called a *tank circuit*, is connected to the RF coil, which stores the energy in an electromagnetic (EM) field, mostly within a wavelength (λ) of the coil confines. This energy oscillates at the RF between the electrical field in the capacitance of the circuit and the magnetic field in the inductance of the circuit and coil. It is the RF magnetic induction field of the coil that couples to the nuclear magnetic spins of the sample being measured. Because this occurs most efficiently at the Larmor frequency, $\boldsymbol{\nu}_0$, of the atomic nuclei under study, the inductive and capacitive reactance of the coil(s) are chosen to tune the frequency of the coil to match it. Ohmic and conductance losses in the circuit, the coil and the quasi-conducting sample and dielectrics result in noise and heat that degrade the signal-to-noise ratio (SNR) of the NMR signal, heating the circuit, and the sample.

Physically, in the vertical-bore magnet in Figure 1(a), the RF coil is mounted in a nonmagnetic cylindrical probe and inserted into the bore of the magnet. In a horizontal-bore magnet (Figure 1b) the RF coil may be mounted around the inside periphery of the magnet bore (e.g., to form a 'body coil') or be attached to the patient bed on a moving table that introduces the sample and coil into the magnet as shown in Figure 1(b) or even attached to the patient directly. The coil, which is often a cylindrical cage-type circuit structure, may be surrounded on its outside, by a Faraday shield as shown in Figure 2. This shield can be part of the probe, or it can be integrated into the bore of the magnet, between the body coil and magnet bore, for example. The Faraday cage is designed through the use of appropriate materials, circuit pattern, and/or geometry to efficiently shield the RF coil from losses induced by the presence of the gradient, shim, and magnet coils within the magnet bore, while allowing the lower frequency gradient magnetic field pulses necessary for spatial localization, to permeate the central bore and enter the sample under study.

This article is dedicated specifically to RF detection coils for MRS. While there can be as many RF coil designs as there are applications for them, 23 of the more common designs are presented. Background, description, circuit drawings, field plots, application notes, and references accompany each design. The intent is to provide the reader with enough information to select appropriate coils for their applications and to direct the coil user to additional literature on building, optimizing, and/or obtaining suitable coils for their desired application. For convenience of classification, RF detection coils for MRS might be divided into two design categories that originate with

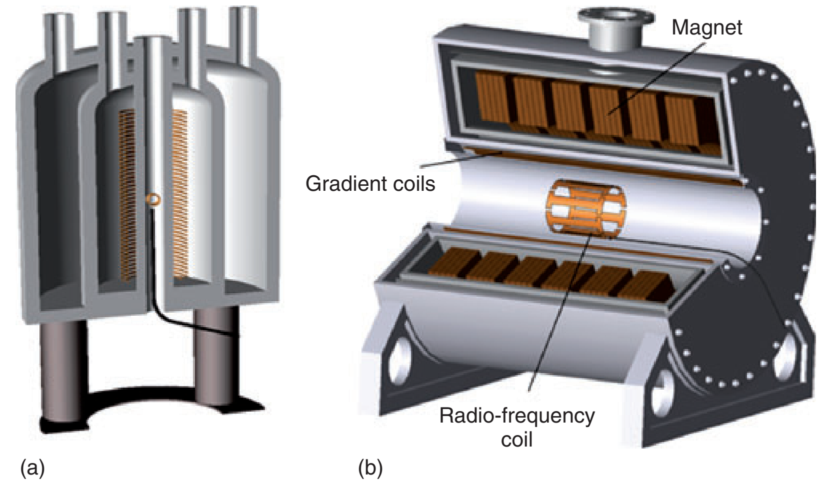


Figure 1. Cross sections of two typical superconducting MRS magnets. Vertical bore MRS magnets (a) are found in pharmaceutical and chemistry laboratories used for use with small animal models, excised tissue samples, and solutions. A small MRS detection coil is placed in its center. The coaxial cable connected to this coil, exits the magnet bore and interfaces with the MRS system's transceiver. Horizontal bore magnets (b) are used for preclinical and clinical MRS. While scaled to different sizes, the coils used in each system are of similar designs to those covered in this article.

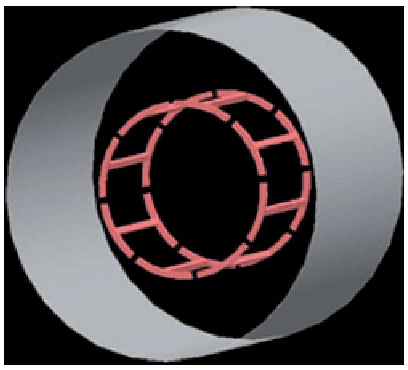


Figure 2. NMR detection coil (center, red) within a cylindrical Faraday shield (gray).

the two Nobel fathers of NMR, Felix Bloch's wire coils[2,3] and Ed Purcell's 'transmission line' coils.[4]

Wire Coils

'Wire' is a lay term for coils made of lumped elements or discrete capacitors in resonance with discrete wire or foil inductors. The simplest of these is exemplified by single or multiple looped wire-wound solenoids. These are easy to build and use and are often the most effective designs, especially in applications for which the coil circuit is shorter than $\lambda/10$ at $\boldsymbol{\nu}_0$, where ohmic and radiative losses are negligible and coil currents are uniform and well behaved. These wire designs include Felix Bloch's first crossed pairs,[3] single-loop solenoid 'surface coils',[5] Helmholtz pairs,[6] and loop arrays.[7]

'Birdcage' coils are another lumped element design typically consisting of a ladder network of rungs wrapped around a cylindrical or ellipsoidal volume former, with the side rails of the ladder joined to form 'end rings'.[8,9] In these designs, the lumped elements, while remaining discrete, are physically distributed around the coil's structure on the rungs and/or end rings. Unlike the simple lumped element wire coils, birdcage coils are most effective when they are electrically adjusted such that the RF completes a full wavelength as it propagates around the ladder network. Such birdcage coils can then be tapped at dual locations physically separated by $\lambda/4$ to provide truly circularly polarized RF sensitivity, known as *quadrature* excitation and detection. This is especially useful for providing uniform fields with maximum SNR over large sample volumes within the bore of cylindrical, superconducting electromagnets. Birdcage designs range from the Alderman–Grant resonator, birdcages, millipede coils, and litz designs.[8,10–13] These ubiquitous coils are readily available commercially and are relatively easy to build.

Transmission Line Coils

Transmission line (transverse electromagnetic, TEM) coils are the distributed circuit analogs of lumped element coils. Their inductance and capacitance, as well as their conductance and resistance, are uniformly distributed around the coil circuit. Transmission lines, consisting of two or more parallel conductors within close proximity, are often the best way to achieve this distributed circuit. Examples of transmission lines are cavities, waveguides, slotted tubes, coaxial lines, microstrips, and stripline structures.[14–19] Robert Pound and Ed Purcell's first coil was based on a re-entrant 'klystron' cavity.[4] Coils built by transmission line principles tend to have a lower inductance and are less radiative. They are therefore better suited to higher resonance frequencies or B_0 and larger sample volumes wherein the coil circuit including the sample cannot be tuned to a single wavelength or less, at $\boldsymbol{\nu}_0$.

Antennas

While coils can be thought of as storing magnetic energy in the near-field, antennas are designed for EM radiation in the 'far field', at distances $>\lambda$. This condition is rarely realized within high-resolution or small-bore horizontal MRS magnets today. However, MRS detection is conceivable in human bore-sized magnets that nowadays can be up to 10.5 T where far-field conditions would apply inside the body. Therefore, antennas are discussed here as well. The simple dipole antenna and dipole arrays are examples.[20]

Surface Coils

Single Loop Coils

The simplest coil circuit can be depicted as an 'RLC loop' as shown in Figure 3. Here, R is the electrical resistance (in Ω) of the wire, L is the inductance of the wire loop (in Henries, H), and C is the capacitance (in Farads, F). In this simple coil, L is typically determined by the diameter and number of turns in the loop. The coil diameter should be matched to the size of the sample volume of interest: single loop surface coils are sensitive to approximately a hemisphere of tissue extending from the plane of the coil on the sample surface to a depth of between about one coil radius and a coil diameter. The loop design may surround the sample or be placed against its surface, hence its common name, 'surface coil'. To afford the most efficient energy transfer (and to serve as a band-pass filter), the coil is tuned to resonate at $\boldsymbol{\nu_0}$ for the nucleus being studied by adjusting the value of C in the coil circuit such that $\boldsymbol{\nu_0} = 1/(2\pi\sqrt{[LC]})$. This coil can be used in transmit-only mode, receive-only mode, or with a transmit-receive (T/R) switch for both transmission and reception – a transceiver. Two central spatial planes to which the transverse RF magnetic (B_1) field of this coil inductively couples are depicted in Figure 4. The red color indicates higher intensity and the yellow and green, lower intensity in these log scale contour plots. An RF current loop like this is also considered to be an RF magnetic dipole because it generates an RF field directed from one 'pole' on one side of the coil through the coil center to another pole on the opposite side of the coil.

Construction of these single loop coils is quite simple, requiring typically less than a meter of, say, ~18 gauge transformer wire or 3-mm copper refrigerator tubing formed into a sample-sized loop and a nonmagnetic capacitor with a value typically in the range 10–200 pF (often with a parallel variable capacitor for trimming) soldered into the parallel position shown in Figure 3. This capacitor is used to tune the inductive loop resonance to $\boldsymbol{\nu_0}$. Figure 5 shows a typical, double-balanced, impedance matching circuit used to match the RF coil when it is loaded by the sample, to a 50-Ω impedance coaxial cable for connection to the MRS scanner. Shown are the coil tuning capacitor, C_t, and two variable impedance-matching capacitors, C_m of approximately equal range, soldered in series with each terminal of the coil. One of these capacitors is soldered to the shield of a coaxial cable and the other is soldered to the center conductor of the cable. When the coil is used as a detector and connected to the scanner's sensitive preamplifier receiver, the coil must be isolated from the transmit power using a transmit-receive (T/R) switch (see section titled 'Phased Arrays').[21]

To allow tuning to higher resonant frequencies with larger sized surface coils, the coil conductor can be split and the tuning capacitance distributed around the coil with discrete elements bridging the gaps as shown in Figure 6. This coil is depicted

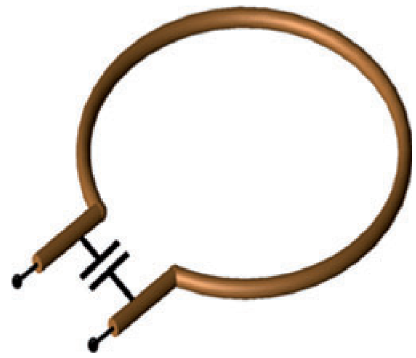


Figure 3. Single loop wire solenoid, 'surface coil' with a tuning capacitor connected across its terminals.

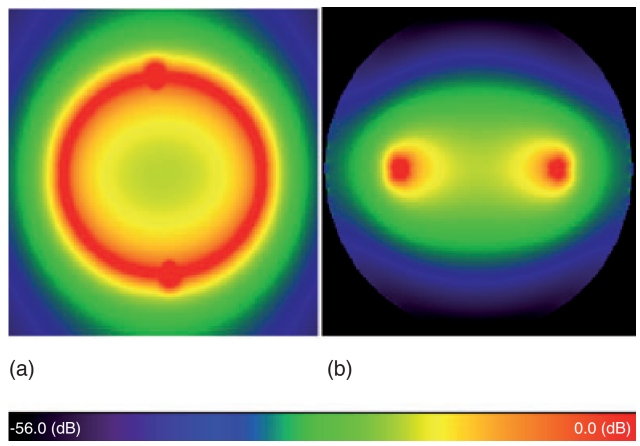


Figure 4. The B_1 field in (a) transaxial and (b) axial central planes for the single loop coil in Figure 3. Log contour scale is included for this coil.

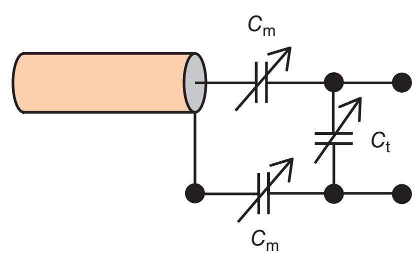


Figure 5. Coil-cable, double-balanced matching circuit. Capacitors C_m are adjusted to match the coil to the cable (typically 50 Ω) and should be approximately equal, while C_t is adjusted to tune the coil to resonance.

as being constructed of foil as well. The B_1 fields of this split coil (Figure 7) are similar to those of Figure 4. Adding lumped capacitance across multiple splits in the coil loop in this manner is a significant advance that reduces the effect of capacitive coupling and electric (E)-field losses between the coil and the sample by breaking up the voltage difference between the coil and the sample.

Surface coils may be constructed of wire or foil and may be shaped in three-dimensional (3-D) geometries and sizes to best fit and couple to a region of interest (ROI). Figure 8 shows a square 'saddle-shaped' surface coil conformed to a cylindrical

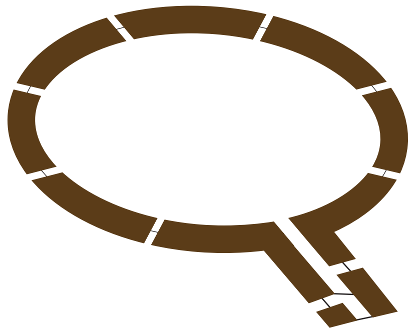


Figure 6. Single loop surface coil, capacitively split, made from foil solenoid. Distributed tuning capacitors (not shown) are used to bridge the gaps.

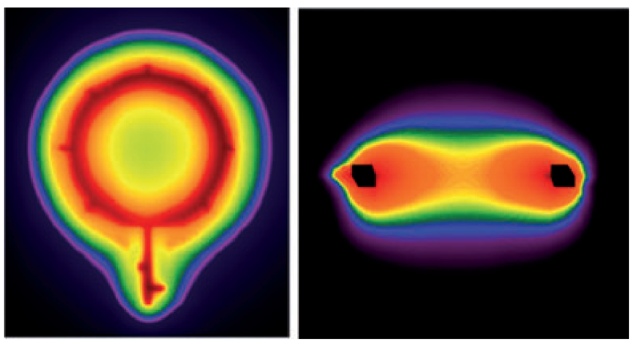


Figure 7. B_1 field in the (a) transaxial and (b) axial central planes for the single loop, split coil in Figure 6 (scale is similar to Figure 4).

surface. Cryogenically cooled and superconducting versions of a single loop coil that can significantly enhance SNR when sample noise is not dominant can be bought commercially.[22–24] One constraint to keep in mind is that these coils must be oriented such that the axis of the coil is generally perpendicular to the axis of the magnet bore to maximize transmit and/or receive efficiency. These loops can also be double tuned for observing two nuclei either serially or simultaneously in an NMR experiment, depending on the way the coil is interfaced to the MRS system and the pulse protocols used.[25]

Multiple Concentric Loops

Multiple loop elements may be configured in an array about a sample to accomplish various objectives. Concentric loops as shown in Figure 9 can be used for localizing a small ROI by field profiling, or when it is desired to provide a uniform excitation field from the larger coil, over the sensitive volume of the smaller one, e.g., for MRS quantification. In this case, the coils are separately connected to the transmitter and receiver but must be electronically decoupled from each other as they are coaxial. Alternatively, the concentric coils can be used for multinuclear MRS wherein each loop is tuned to a different NMR frequency.[26] For multinuclear applications, the larger of the two loops might be used for proton (^{1}H) scout imaging, shimming, decoupling, and/or Overhauser enhancement

Figure 8. Rectangular, sample-conforming saddle-shaped, foil solenoid surface coil. The B_1 field in this case is detected primarily horizontal in this diagram.

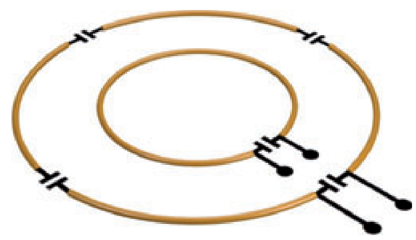


Figure 9. Concentric loops surface coil. The outer coil is split and fitted with distributed tuning capacitors.

(nuclear Overhauser effect, nOe; see ***The Basics***), while the smaller loop is used to acquire spectra from a second nuclear species. In this case, both coils are used in transceiver mode, with each is connected to its tuned matching network, T/R switch, filter, preamplifier, and power amplifier.

Phased Arrays

Multiple loops can also be configured into phased-arrays of nonconcentric coils as shown in Figure 10.[7] Phased arrays, used as surface or volume coils, extend the performance of an electrically small and efficient loop to an arbitrarily large field-of-view (FOV). Phased arrays are most often used as application-specific, receive-only detectors that operate in conjunction with a larger, uniform transmit coil such as the body coil of a clinical magnetic resonance imaging (MRI)/MRS system.[27–29] Each loop is connected to its own receiver channel often beginning with a low-noise gallium–arsenide field-effect transistor (GaAsFET) decoupling preamplifier embedded in the each loop (see ***Magnetic Resonance Spectroscopy Instrumentation***).

As in the case of the single coils, the multiple preamplifiers connected to each loop in the array must be protected at the loop element terminals from excessive induced voltages during transmission, and each loop must be actively detuned during the RF transmit pulse periods because the induced currents will otherwise oppose the transmit field. This is achieved by

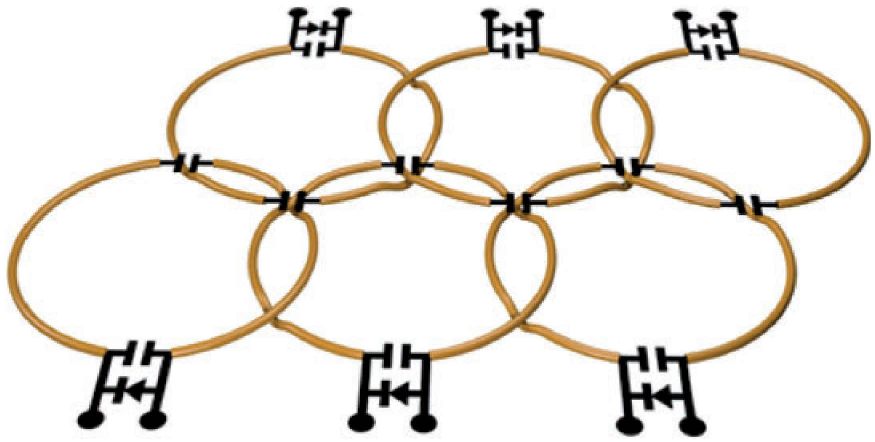


Figure 10. Six-channel phased array comprised overlapping split circular surface coils.

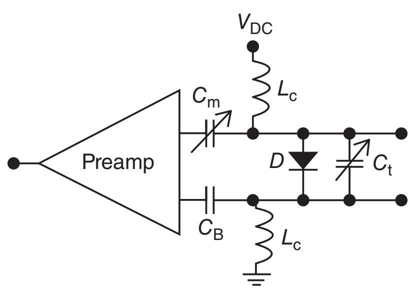


Figure 11. Parallel PIN diode (D) and preamplifier detuning circuit for one channel of a receiver phased-array. The diode is switched on by voltage V_{DC} from the scanner during excitation. The V_{DC} signal can be conveyed on the receiver output line, capacitively decoupled from the preamplifier. C_B, C_m, and C_t are tuning and matching capacitors.

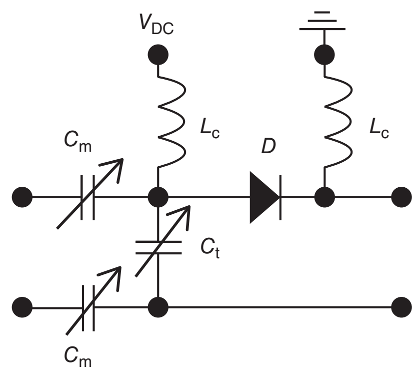


Figure 12. Series PIN diode detuning circuit for transmit phased array elements.

a scanner-activated high-voltage limit, low-noise, positive-intrinsic-negative semiconductor (PIN) diode circuit as shown in Figure 11. Basically, the scanner provides a voltage (V_{DC}, typically 5–15 V) during transmission that turns on the diode and shorts out the tuning capacitance, detuning the coil. Figure 10 depicts a single diode in each loop to represent this detuning circuit in a six-coil array with distributed tuning capacitance. Transceiver arrays that use the coil elements for both excite and receive will make use of a PIN diode T/R switch only, to switch the coil terminals sequentially between the power amplifier during transmission (to protect the receiver) and to the receiver during reception.

The loops in the array must also be isolated from their neighbors to decouple correlated sample noise and optimize parallel imaging performance by insuring that the signals from the multiple channels are as independent as possible. This decoupling is accomplished by cancelling the mutual inductance between neighboring loops by means of capacitive or inductive adjustments. This typically begins with careful adjustment of the overlap between adjacent loops, such as shown in Figure 10, to ultimately achieve an isolation of 12dB (network analyzer S12 measurement) or more for best coil operation. The larger transmit volume coil must also be detuned during signal reception. This detuning is also achieved by scanner-activated PIN diode circuits in the transmit coil that shunts signal to ground or produces an open circuit as in Figure 11 or 12, respectively.

Arrays bring multiple advantages to MRS. They can be tuned to more than one nucleus by double tuning each loop or by tuning adjacent loops to alternate frequencies. As receiver/detectors, arrays offer a means of covering larger ROIs more efficiently with electrically smaller, less radiating coil elements than large volume coils or loops. In this respect, phased-arrays may be considered volume coils in their own right, other variations of which are discussed later. Receiver arrays can greatly enhance SNR (7) as well as the speed of data acquisition through parallel imaging and multiband techniques.[27–29] As transmit elements, they can be used for optimizing B_1 homogeneity at higher B_0 field strengths, where wavelengths are short in tissue, for minimizing RF power deposition over an ROI by field profiling, B_1 shimming, and 'beam-steering' selective excitation techniques. For all of these reasons, the use of phased arrays is widespread, at least for ^{1}H MRI and MRS. Their availability for non-^{1}H nuclei, however, is more limited.

Volume Coils

Solenoids

Wire-wound solenoids are one of the earliest choices of volume coil design. Figure 13 shows a multiturn solenoid whose purpose is to extend and intensify the linear B_1 field of a single loop along the axis of the RF coil. Such a coil, with multiple turns around a volume, has significantly higher coil inductance than single-turn designs and thus requires a proportionately lower tuning capacitance. Owing to its high inductance, this design is most effectively used as a simple transceiver coil for small laboratory animals, excised tissue samples, or test tube or capillary samples bearing lower gyromagnetic ratio nuclei where a lower-$\boldsymbol{\nu}_0$ coil is required. In addition to being a good performer for small samples at lower frequencies, this is a simple and easy coil to build and use in the laboratory.

Crossed Pairs

Felix Bloch's crossed-coil design is shown in Figure 14.[3] This design is comprised of a pair of independent loops and can be used in two ways. Bloch connected one loop to his transmitter and the other loop to his receiver. When the loop planes are crossed or perpendicular as shown, the transmit field of one loop is geometrically isolated from the received field of the other loop. Another use for this loop pair is to generate a quadrature or circularly polarized transmit and/or receive field. In such a coil set, the transmit (receive) signal is divided into two and separately fed to (detected from) each coil with a 90° phase shift. Circularly polarized excitation provides a more uniform field over a larger FOV in the sample than using one or both coils without the quadrature excitation, and the SNR of the NMR signal detected in quadrature from the sample will be improved by as much as a $\sqrt{2}$-fold.

A third way to use this coil is as a double-tuned or multinuclear coil. In this case, one of the two loops is tuned to the $\boldsymbol{\nu}_0$ of a first nuclear species and the second loop is tuned to the $\boldsymbol{\nu}_0$ of a second nuclear species. Such a crossed-Bloch pair can be used to excite and/or detect two different MRS signals from two different species to perform heteronuclear decoupling, spectral editing, or nOe (as with the coil pair in Figure 9). These coils

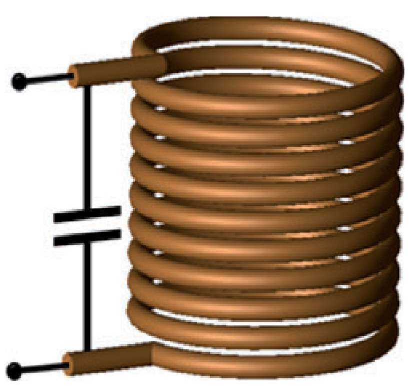


Figure 13. Multiloop wire solenoid with tuning capacitor connected across its terminals. The B_1 field is primarily in the vertical direction in the diagram.

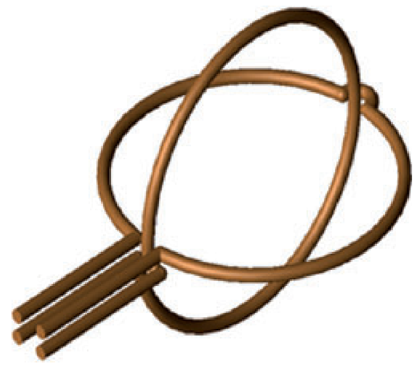


Figure 14. Crossed wire loops depicted as a two-coil orthogonal pair. The B_1 field is in the plane perpendicular to the four-feed terminals.

can be used as transmit-only coils, receive-only coils, or transmit and receive coils. Again however, the axes of this coil pair must be oriented substantially perpendicular to B_0, and despite having more open space than the solenoid that obstructs the magnet bore when used in vertical bore magnet configurations, access to its central sample space remains somewhat obstructed.

One potentially useful variation, which might be considered either as a three-channel phased array or as a form of crossed-coil design, is shown in Figure 15. Here, the crossed pair is opened and spread over a curved, anatomic surface, while still achieving a quadrature field over a portion of the sample space. This configuration can be useful as a quadrature transceiver for imaging. Alternatively, it can be combined with a second transceiver loop tuned to a low gyromagnetic ratio nucleus and nested at the intersection of the larger crossed pair to provide efficient, local ^{1}H MRI performance, and sensitive non-^{1}H MRS.[30,31]

Saddle Pairs

The series-wound saddle pair shown in Figure 16 can be viewed as a two-loop solenoid but wound to efficiently and conveniently accommodate a sample in a linear B_1 field perpendicular to the magnet's B_0 field, with the desired cylindrical access.[6,32–34] More desirable for some applications is the combination of two such saddle pairs with each pair rotated 90°, in the configuration pictured in Figure 17. This crossed saddle pair can be used in all of the ways possible with the crossed pair of Figure 14, but in a much more space-efficient design for performing NMR in cylindrical bore superconducting magnets. This coil can generate a fairly uniform circularly polarized B_1 over its entire volume perpendicular to the magnet bore and to the coil's cylindrical axis, by separately exciting each pair with two RF signals phase shifted by 90°, as discussed earlier. To minimize coupling and maximize isolation between the two saddle pairs for best SNR and quadrature performance and to cover the largest volume FOV, the overlap between the two pairs is adjusted to cancel or minimize the mutual inductance between the loops. The amount of overlap depends on the coil dimensions and the sample loading and is typically determined empirically by adjusting the size and orientations of the coils until a maximum isolation (e.g., an S12 > 12 dB measured on a network analyzer) is achieved. As with the crossed-Bloch

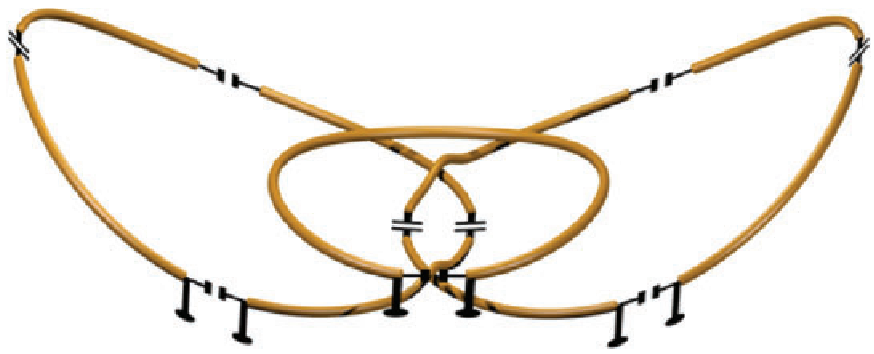


Figure 15. Quadrature surface coil formed from a crossed-coil pair with a third nested loop for use as a three-channel phased array or a multi nuclear applications with the inner coil used for MRS of a low-sensitivity nucleus.

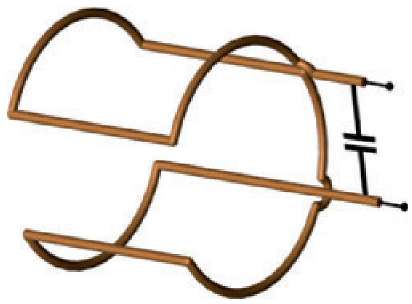


Figure 16. Series wound saddle pair (compare with Figure 8) with a tuning capacitor. The B_1 field is primarily in the vertical direction.

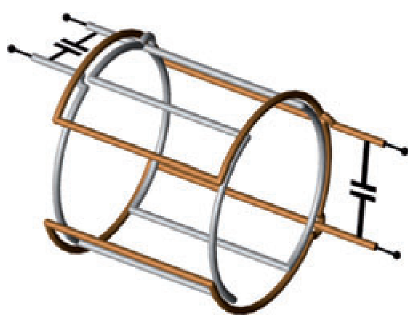


Figure 17. Two series-wound crossed wire saddle pairs rotated so that their B_1 axes are orthogonal to form a crossed saddle pair.

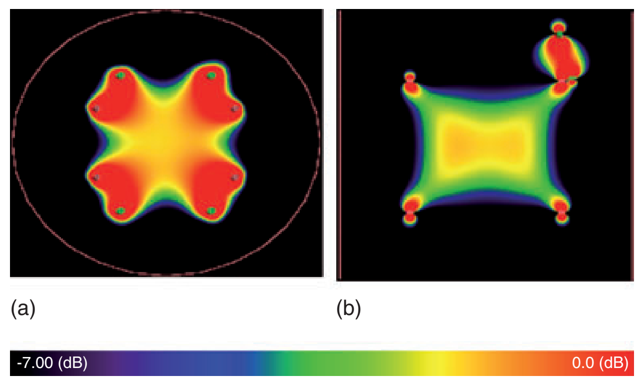


Figure 18. B_1 field in the (a) transaxial and (b) axial center planes for the circularly polarized crossed-wire saddle pairs of Figure 17. The log color scale for this figure applies to all following field contour figures.

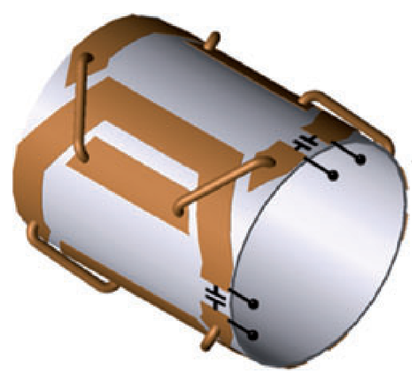
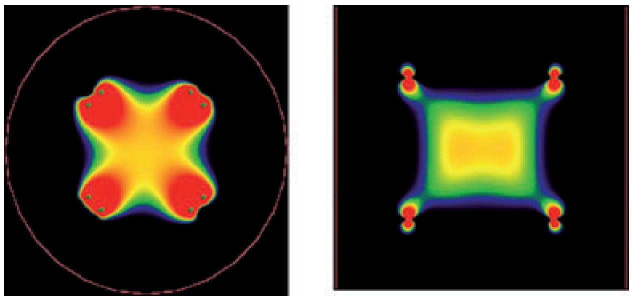


Figure 19. A crossed foil saddle pairs with wire jumpers spanning the crossover points.

pair, each saddle pair can also be used as separate transmit and receive coils, respectively or tuned to the NMR frequencies of two different nuclei for multinuclear MRS. Many modern MRS detector coils use variations of the crossed saddle pair design. The fields in the transaxial and axial central planes for this coil are shown in Figure 18.

Saddle pairs can be constructed from foil as well, as illustrated in Figure 19. Foil may be used for ease-of-fabrication reasons. For example, foil coil elements can be milled, etched, or glued to a supporting member with ease, accuracy, and precision from circuit board material or copper tape. The substrate material can add to the stability of the coil circuit as well. To avoid parasitic capacitance between overlapping loops separated by thin insulation, wire jumpers can be used (Figure 19). While the electrical quality factor Q ($=2\pi\nu_0$ L/R), a measure of the efficiency of an RF circuit, may be a bit lower with foil loops vs wire loops or copper tubing, the FOV coverage of a foil-etched coil is very similar to a wire wound coil of the same dimensions (compare Figures 18 and 20).

If the two opposing sides of each saddle coil are not connected in series but are instead isolated for connection to their own receiver preamplifier or RF power source, a four-channel cylindrical volume coil results, as shown in Figure 21. The



(a) (b)

Figure 20. B_1 field in the transaxial and axial center planes for the circularly polarized crossed-foil pairs in Figure 19.

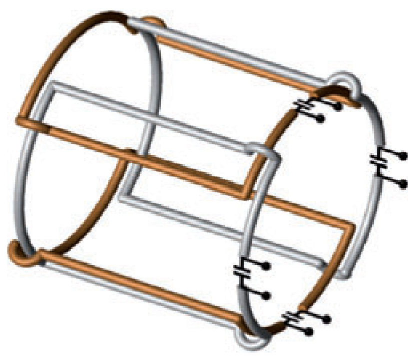


Figure 21. Four-loop wire volume coil formed from four saddle coil loops on a cylindrical former. The B_1 field is in the plane perpendicular to the cylindrical axis.

Figure 22. Four-loop volume coil fabricated from foil saddle coil elements with wire jumpers spanning the crossover points.

four channels can be used as a multichannel transmit and/or receive volume coil array, or for multinuclear MRS applications wherein one or more of the individual coil elements tuned to a different NMR frequency, to enable combined ^{1}H MRI, MRS, or decoupling experiments, as discussed earlier. Such designs mark the evolution toward multielement volume coils. Figure 22 shows a version fabricated with foil and wire jumpers.

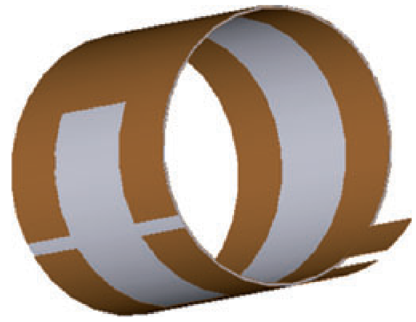


Figure 23. Alderman–Grant coil. The copper strut on top is mirrored on the bottom and the B_1 field is directed left-right through the window.

Slotted Tube Resonator

Another coil design fabricated from foil on a cylindrical former is the 'slotted tube resonator' of Alderman and Grant shown in Figure 23,[8] used for early 1.5 T MRI. It is composed of a pair of end-rings connected by two broad opposing conducting sections that tend to reduce inductance that facilitates tuning at higher $\boldsymbol{\nu}_0$, leaving windows that span about 100° on opposite sides of the tube (the 'slot'). Similar to in surface coils, the end conductors are split in four places to accommodate distributed tuning capacitors. Electrically, the structure is analogous to connecting two end-rings with distributed capacitance in parallel, via the broad axial sections to form a tuned LC circuit.

Birdcage Coils

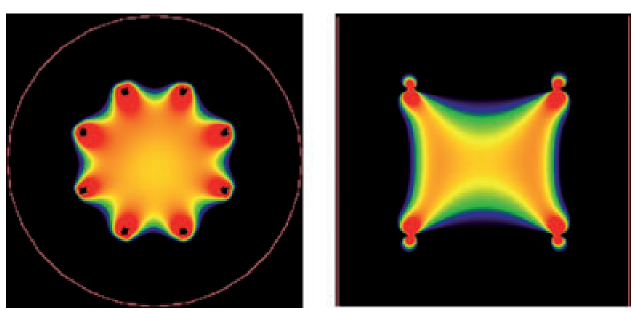
As noted in section titled 'Wire Coils', birdcage coils are wire or foil wound, lumped element coil circuits based on ladder rather than loop or solenoid circuits.[8,10–13,35–38] This coil circuit type achieves a uniform B_1 field within its typically cylindrical volume by a 360° azimuthal distribution of linear currents surrounding the volume. The return path for these linear current elements is on end rings, and the rungs provide current paths connecting them. In theory, an infinitely long birdcage fitted with an infinite number of rungs would achieve perfect field uniformity within the coil volume. However, a 'square' coil whose length equals its diameter or shorter is sufficient for most MRI and MRS, FOVs.

The electrical circuit of a birdcage coil is equivalent to a narrow-bandwidth filter circuit whose center frequency can be tuned to $\boldsymbol{\nu}_0$ with lumped tuning capacitances distributed either in the rungs (low-pass birdcage), with capacitance distributed in the rings ('high-pass birdcage'; Figure 24), or with lumped capacitance distributed in both the rungs and the rings (band-pass birdcage). Each design has performance advantages and disadvantages that depend on frequency and application.[10,11,38] The high-pass coil can achieve higher frequencies – an important feature – but can present sample loading problems. All the birdcage structures are symmetric and can be driven and/or the signal received as a circularly polarized coil for best SNR and FOV coverage. For quadrature operation, transmit and/or receive cables are attached 90° apart at the terminal positions shown, using the matching circuit of Figure 5. As in section titled '**Phased Arrays**', the RF power must be provided as two channels that are 90° phase shifted at the two coil ports.

Figure 24. High-pass bird cage coil with feed ports spaced at 90° for quadrature excitation or detection.

This can be accomplished using a single RF power amplifier and a 'quadrature hybrid' circuit. The same hybrid can be used to recombine the two, 90° phase-shifted receive channels representing the circularly polarized (B_1) NMR signal, to feed a single receiver channel in the most simple transmit and receive interface for these coils.

While ubiquitous transceiver head coils are still in use, the most common modern application for the birdcage is for uniform transmit coils used in conjunction with phased-array receiver coils. Because clinical transmit coils are very large, these big resonators require a significant number of capacitive breaks in both the rings and the rungs to render a band-pass coil. Such coils operate on the edge of self-resonance, are very radiative, and require large, 32-kW pulsed power amplifiers to drive them. The RF power deposition from these lossy radiators presents significant limitations to RF protocols. Accordingly, the utility of birdcage transmitters becomes limited for most larger preclinical and human applications at 3 T and above, the field at which many MRS applications begin. Birdcages almost always operate inside a Faraday shield (Figure 2). The center-plane birdcage B_1 fields calculated in Figure 25 show the characteristic FOV for this uniform transmitter. The end-rings tend to cancel the MRI/MRS signal, causing a concave B_1 field cut off at the ends of the coil. This can be advantageous, as it prevents signal aliasing from beyond the coil volume.



(a) (b)

Figure 25. B_1 field in the (a) transaxial and (b) axial central planes for a circularly polarized bird cage coil.

Birdcage coils can also be double tuned for multinuclear MRS. A common design approach for double tuning a birdcage is to add a second set of end rings, effectively creating two coils of different electrical lengths in a single circuit.[35]

Millipede Coils

The Varian Inc. millipede MRS coil was a favorite that is still produced privately by its inventor, Ernest Wong at Extend MR, LLC.[13,37] The millipede coil is analogous to a foil made low-pass birdcage design that maximizes the number of rungs. The tuning capacitance is displaced to each end of the coil rather than in the middle being distributed in the foil rung–ring overlap at the ends of each rung. The benefit of this design is its high uniformity over most of the volume of the coil. See the millipede coil and its field contours in Figures 26 and 27. It is operated mostly as a circularly polarized transceiver coil. As with a birdcage, its transmit and receive terminals are set 90° apart.

Litz Coils

Litz coils are a wire or foil wound, lumped element cage coil variant that are produced and sold to many users by Doty Scientific, Inc., Columbia SC.[12,36] 'Litz' meaning braided or woven refers to the woven appearance of these coils. This design uses capacitive overlaps by its foil rungs and rings to achieve a tuned cage-like mode of operation. While few MRS experimentalists

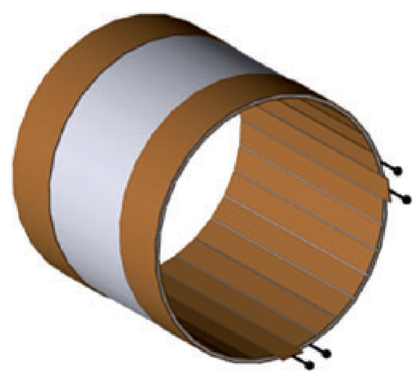
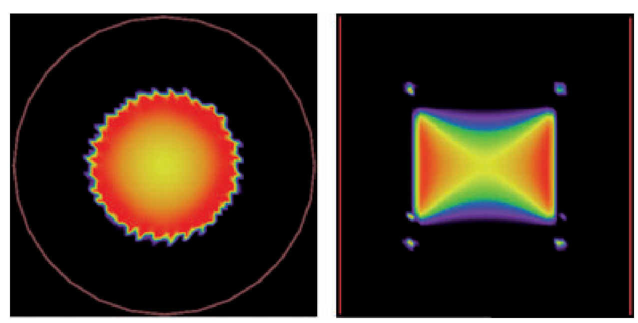


Figure 26. Millipede coil.



(a) (b)

Figure 27. B_1 field, (a) transaxial and (b) axial for circularly polarized millipede coil.

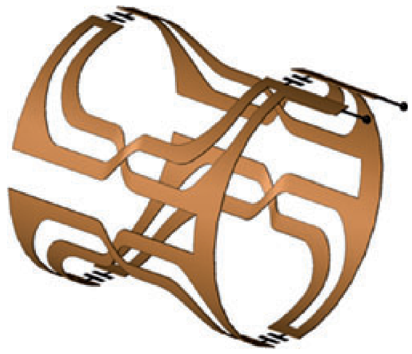


Figure 28. Litz coil configured for linear operation. The B_1 field is vertical through the windows, top and bottom.

build these intricate coils, they are well made by their manufacturer and are widely used in the mostly vertical bore and preclinical horizontal bore markets. For simplicity of viewing, a linear litz coil is shown in Figure 28 with the B_1 fields produced in Figure 29. Circularly polarized and double-tuned litz coils are more common and show good performance.[36] A linear cage-like field contour is apparent.

Transmission Line Coils

As noted earlier, the first MRS coil built by Robert Pound for Ed Purcell was a re-entrant klystron transmission line design based on microwave radar research.[4] Any circuit that propagates a TEM wave defines a transmission line, and the name 'TEM' is applied to this class of coil circuits. TEM coils include coaxial, stripline, and microstrip transmission lines, as well as resonant cavities and waveguides.[17–19,39–43] The primary objective for employing these designs is to achieve high efficiency and uniform current and field distributions at high B_0 and Larmor frequencies and large sample volumes including human systems. The lumped element circuits comprised discrete inductors and capacitors employed in the wire-wound coils described earlier are inherently low-frequency devices. Their design equations are approximated by Thevenin's and Kirchoff's equivalent circuit equations, and their fields are approximated by the Biot–Savart Law.

When circuit wavelengths exceed a tenth of a wavelength, they begin to radiate energy like an antenna. As the sample sizes remain the same and the wavelengths decrease with increasing B_0 and $\boldsymbol{\nu_0}$, lumped element designs must be abandoned in favor of transmission line designs and Maxwell's time-dependent EM field calculations. Similar to ultra-high-frequency (UHF) circuits in the telecommunications industry, UHF NMR coils benefit from transmission line (TEM) design. With such designs, there are few lumped elements other than trimmer capacitors and chokes (inductors). The bulk of the inductance, capacitance, conductance, and small resistances of the coil antenna are parameters in transmission line equations and are continuously distributed throughout the TEM circuit. The TEM coils have high Q values that provide for efficient operation at higher operating frequencies where lumped element devices become problematic or no longer work, but where MRS shows much promise.

An early example of a transmission line coil for MRS is the Schneider and Dullenkopf, slotted tube resonator pictured in Figure 30.[14,15] The transmission line nature of this coil is easy to see from the outer conductor serving as a shield, with a coaxial, albeit hollow, center conductor containing the sample space. The transmission line parameters are distributed over the length of this structure, between the shield and the center conductor plates. This design has evolved into a more useful and widely used 'TEM' design for ultra-high-field systems of $B_0 \geq 4$ T, as pictured in Figures 31 and 32.[17,18,39]

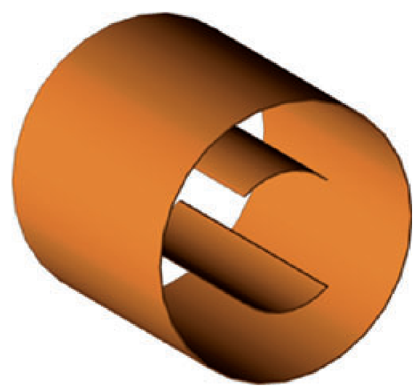


Figure 30. Slotted tube resonator.

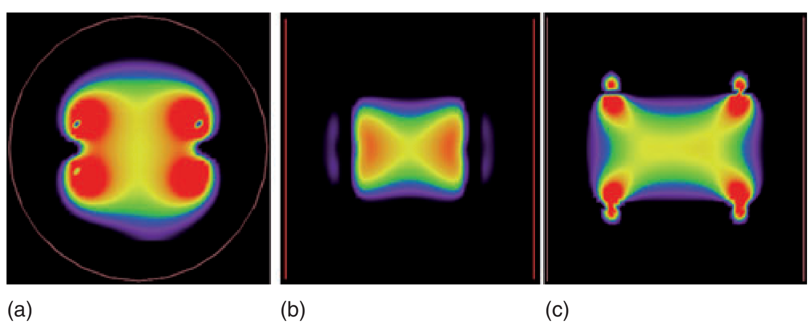


Figure 29. Transaxial B_1 in the (a), xz (b), and yz (c) axial central planes of the linearly polarized litz coil in Figure 28.

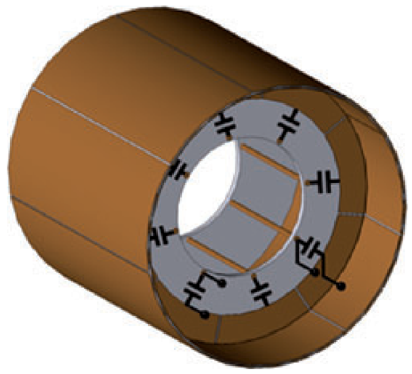


Figure 31. An inductively coupled coaxial line TEM coil.

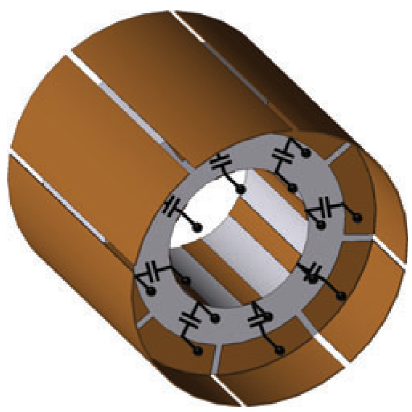


Figure 32. An actively coupled TEM coil formed from striplines.

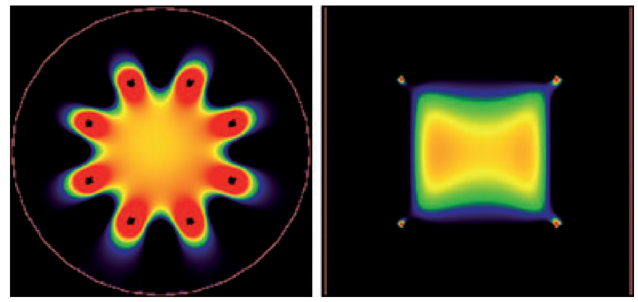


Figure 33. B_1 field in the (a) transaxial and (b) axial central planes of a circularly polarized TEM coil.

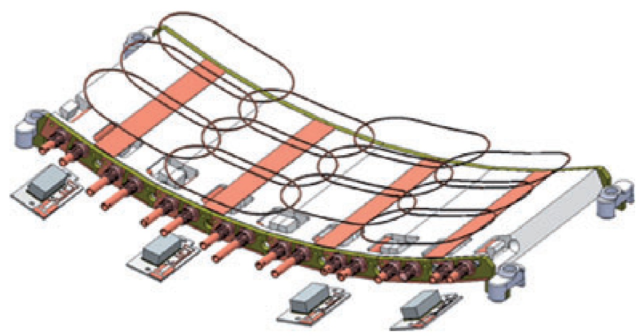


Figure 34. TEM surface array combined with a nested loop phased array. The loops are centered over the striplines.

TEM Coils: Coaxial Lines, Striplines, and Microstrips

Modern TEM coils are capacitively shortened and tuned, half-wave transmission line resonators that, while using similar principles to early tube resonator designs, possess practical features for modern utility, such that their final appearance bears little resemblance.[19,39–43] Among these features, the center conductor 'rungs' of these coils generate a highly uniform field like the rungs of a birdcage. Unlike the end-ring-connected birdcage coils, however, the rung conductors of the TEM coil are not connected at all. Their current return path is on the shield of the coil, as in a transmission line. This feature produces several benefits. First, each rung in the TEM coil is an electrically short and efficient, independent resonator: the TEM coil is effectively an array of these discrete resonant units. The useful frequency of this coil therefore depends on the inductance of a single rung only and not on the circuit size of the whole coil as in the case of the end-ring-coupled birdcage coil. TEM coils have been shown to produce efficient body coils to 8 T and above.[39,44]

Second, because each element operates independently of the next, the individual elements can be separately driven by a different transmit signal with 5 degrees-of-freedom, namely phase, magnitude, frequency, space, and time. This allows for B_1 shimming, beam-steering spatially selective excitation, parallel imaging, and multiband operation. In addition, because there are no end-ring field cancellations, the FOV for a TEM coil is longer and more uniform than a birdcage of equal length. Compare the TEM field contours of Figure 33 to the birdcage contours of Figure 25. As with the birdcage, these coils can be used as receivers, transmitters, or transceivers.

While cylindrical volume coils are shown in Figures 30–32 because TEM coils are essentially an array of independent transmission line elements, they can be conformed to many useful spatial forms. A common example is the TEM surface array (Figure 34) used for ultra-high-field body MRI and MRS.[39] The TEM coil can also be double tuned by tuning adjacent rungs to an alternate NMR frequency of a desired nucleus. As with most double-tuned coils, an ^{1}H frequency is typically chosen for scout imaging, landmarking, shimming, and sometimes decoupling, and a second, non-^{1}H nucleus is tuned for MRS. This coil works quite well because the coil couples to the same sample volume for both nuclei, achieving comparable sensitivity at two frequencies as its single-tuned counterpart owing to the independent operation of the individual resonators.

Figure 34 shows a commonly used TEM configuration of linear transmit elements paired with a loop array receiver. Linear transmit elements in a surface array such as this, or in a volume coil, are relied upon for generating a more uniform B_1 excite field than is achievable with small single elements, while a close-fitting phased array next to the sample acts as a sensitive signal detector. Note that when linear transmission line elements are used in this way, a geometric field decoupling is realized when the loop elements are centered over the line

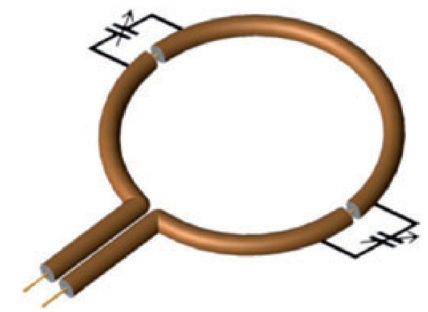


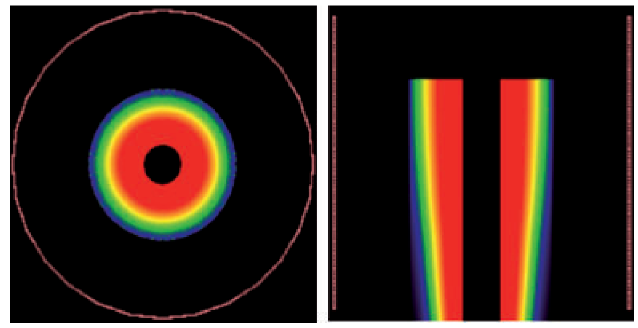
Figure 35. Transmission line, double-tuned surface coil.

elements. The assembly shown can be used in a number of ways. It can be used as a surface array for covering a larger ROI than achievable with single elements. When this is paired with another similar assembly on the opposite side of a body, it is very effective and efficient for body imaging, as compared to a whole body cylindrical coil of similar design.

Other Transmission Line Coils

TEM coils can take other forms as well. For example, the double-tuned surface coil pictured in Figure 35 is constructed from a section of semi-rigid coax cable, although the same coil could be executed with stripline or microstrip. In this coil, the continuous center conductor is tuned by circuit length and trimmer caps to a lower frequency, and the split shield is tuned with trimmer capacitors to a higher frequency. Most of the reactance of the coil is continuously distributed in transmission line manner to yield a balanced multinuclear surface coil.[1]

Another interesting transmission line design for high-resolution MRS is pictured in Figure 36.[19] As seen in Figures 36 and 37, a B_1 field of maximum intensity is generated at the tip of a resonant, halfwave coaxial line. Sample space is drilled into the dielectric between the outer and center conductor of the semirigid coaxial line section. For maximum efficiency, this dielectric can be selected to match the sample in order to minimize reflection and losses. This 'stubline' coil seems well suited for high-resolution, vertical bore magnets. The same approach could be extended to horizontal bore systems using dielectric



(a) (b)

Figure 37. B_1 field, both (a) transaxial and (b) axial, for stubline coils.

wave guides with access holes or apertures cut though the waveguide sidewalls at current nodes that couple to anatomy at those points with maximum efficiency. These structures can be double tuned by appropriate choices of dielectric inside and outside of the cavity.[19] The inside of the cavity 'sees' one wavelength and the outside of the cavity sees another, typically either air or tissue.[45]

Antennas

Finally, we conclude with a third coil-circuit class, the antenna.[20] While the wire and transmission line coils above are designed to store an RF magnetic field within or adjacent to their circuit, an antenna is designed to radiate its energy efficiently to the far field. Using an antenna may be appropriate when a sample ROI is in the far field ($>\lambda$). This condition occurs for MRI and MRS at the highest B_0 and largest physiological samples. This class of detectors has proved successful for reaching deep body ROIs at $B_0 = 7$ T (300 MHz) to 10.5 T (450 MHz), for example. Antennas comprised arrays of center-fed dipoles, as pictured in Figure 38, have been built and tested and found to perform favorably compared to TEM coil arrays under these highest field conditions for human MRI/MRS. Compare the field contours of the dipole in Figure 39 to the TEM contours in Figure 33. Because, like the TEM coils, these

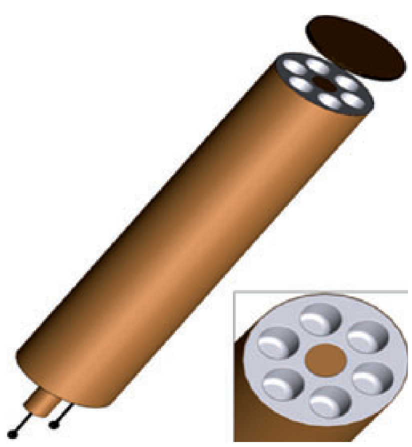


Figure 36. Open and short stubline coils.

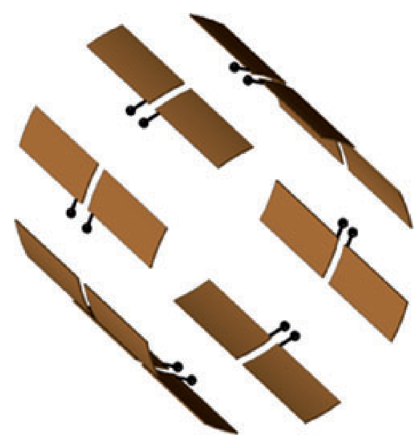
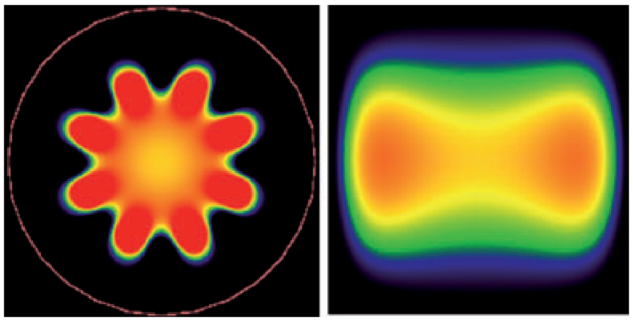


Figure 38. Dipole array arranged around a cylindrical sample chamber.



(a) (b)

Figure 39. B_1 field, (a) transaxial and (b) axial, in the central planes of the dipole array in Figure 38.

arrays have no end-rings, the axial field extends further, and more uniformly than a birdcage of equal physical length. However, the birdcage would not work at these highest frequencies and sample dimensions. Similar to TEM coils and other array structures, these dipole antennae can be double tuned by tuning alternating elements in the array to alternating frequencies.

Biographical Sketches

J. Thomas Vaughan Jr, B. 1957. BS Biology, 1982, BS Electrical Engineering, 1982, and PhD Biomedical Engineering, 1993. He has made a career of leading engineering efforts to advance high-field human MRI including 2 T at UT Southwestern beginning 1984, 4 T at UAB beginning 1989, 7 T at MGH/Harvard beginning 1995, and 7 T, 9.4 T, 10.5 T beginning 1999 in Minnesota. He has published approximately 125 papers and holds 50 patents. He has built and tested, in consultation with the original designer, all the coils discussed in this chapter. Currently, he is Director of MR Research at Columbia University.

Russell Lagore, B. 1986. He received his BS Electrical Engineering in 2010 and MS Biomedical Engineering in 2013 from the University of Alberta. At the University of Alberta, he studied magnetic field effects on preamplifier noise figure. He is currently an RF engineer and Research Fellow at the Center for Magnetic Resonance Research at the University of Minnesota.

Jinfeng Tian Received his PhD from University of Minnesota in Biomedical Engineering in 2007. After serving as an RF Engineer in MR Instruments Inc. for 2 years, he has been with the Center for Magnetic Resonance Research at University of Minnesota as a Research Associate. His research interests include MRI RF coil design and RF safety evaluation with numerical tools.

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