

Volume Coil Types and Design Principles  
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The development of new RF coil technology that can generate a radio frequency magnetic field in a large, body-sized volume has been a major constraint on the progress toward higher field magnetic resonance imaging. This coil technology evolved far from its original roots in NMR spectroscopy of test tube sized samples and took place in a historical context with assumptions and requirements that may no longer be appropriate to today's problems. In this presentation, I will give a somewhat personalized account of RF coil technology as I experienced it in my career. I will emphasize what I call the basis principles of good coil design, how they apply to birdcage and TEM resonators, and how one should approach the problems presented at current high fields (7T or so).

The term "RF coil" is a misnomer that carries over from earlier NMR spectroscopy days when one could make a probe by winding copper wire on a test tube and tuning it with a capacitor. The resonant circuit consisted of two physically separated, discrete components: the inductor that stored energy in its magnetic field and the capacitor that alternately stored energy in its electric field. To operate as an ideal lumped element circuit, these two components must be small compared to the wavelength of the operating frequency. Extrapolating from this model to body-sized coils, Hoult and Lauterbur [1] predicted that MR imaging would be limited to frequencies below about 15 or 20 Mhz. In particular, due to the finite speed of light, the current in the coil conductor would be properly phased along its length only if the total conductor length was restricted to about one eighth of a wavelength. Alternatively, the maximum operating frequency of a coil is limited by its internal stray capacitance.

Fortunately, these restrictions can be overcome at high frequencies by following a number of design principles. I state them categorically, followed by some motivation.

A. Minimize the total inductance.

1. Use a single series turn or multiple parallel turns.
2. Large conductors have less inductance than small conductors. Lowering the surface current density corresponds to lowering the tangential B-field strength, which lowers the energy density.
3. The purpose of the coil is to produce a RF magnet field in the volume of the sample. The coil's sensitivity is proportional to the B-field in the sample per unit current in the coil. Magnetic energy stored outside of the sample adds to the inductance without contributing to the signal. Therefore, make sure all the inductive elements contribute to the field in the sample volume.
4. High inductance corresponds to high impedances, which require high voltages per unit current.
5. Do not use capacitors that are inductive.

B. Increase the speed of light.

1. The wavelength for current traveling along a conductor is determined by the phase shift induced by the conductor's series inductance per unit length

interacting with its (stray) shunt capacitance per unit length. If a number of capacitors are inserted at equal intervals along a length of conductor and the value of each capacitor is chosen to resonate with the inductance of the intervening conductor segment, there will be virtually no phase shift induced along the conductor at that one frequency. The current has a constant amplitude along the conductor, i.e., its wavelength is arbitrarily long. The phase velocity of light can be made arbitrarily large at any given frequency, but the group velocity suffers accordingly.

2. The use of multiple series capacitors is the MOST IMPORTANT CONCEPT given in this lecture. With enough capacitors properly spaced, any sized structure can be made to resonate for any MRI obtainable field strength. Since we did not think that we could patent the multiple capacitor idea due to prior art, we did not publish our methods.
3. The use of multiple capacitors has some secondary advantages: the size of the net capacitance can control the size of currents in parallel paths; the location of the capacitors influences the orientation and strength of the electric fields that are induced in the sample; distribution the total voltage across multiple series capacitors eliminates voltage breakdown problems; by reducing the effect of stray capacitance, the coils are not detuned as much by the sample.

#### C. Maximize coil efficiency.

1. The SNR of a coil is inversely proportional to the square root of the power needed to produce a unit of RF magnetic field, which is, in turn, proportional to the equivalent series resistance of the loaded coil. The equivalent resistor is the sum of the inherent resistance of the coil conductor elements, the equivalent resistance due to magnetically induced eddy currents in the tissue, and the equivalent resistance due to dielectric losses.
2. Coil resistance should be minimized by using conductors with large surface area.
3. The tissue losses increase rapidly with the volume of tissue exposed to RF fields. Therefore, choose a coil geometry that excites only the region of interest.
4. Dielectric losses are minimized by the use multiple capacitors to minimize the electric field perpendicular to the sample surface.

#### The Birdcage Resonator

I conceived the birdcage resonator about one month after I joined GE Medical Systems in February of 1981 [2, 3, 4]. The working assumption was that we needed a highly homogeneous RF field for both the transmission and the reception. The fear was that if the coil introduced some subtle shading artifact, then the radiology might make a misdiagnosis. The later advent of receive-only surface coils with superior sensitivity and no uniformity indicates how far off base our assumption was.

The lowpass birdcage [3] is essentially a balanced lumped element delay line with  $N$  segments wrapped around a cylinder and joined end to end. The ending segments form the inductive elements and the straight rungs contain the capacitive elements. A

resonance occurs whenever the phase shifts add up to give a standing wave with an integer number of wavelengths around the cylinder. The highest frequency mode corresponds to the Nyquist condition with  $N/2$  wavelengths. The lowest frequency occurs for the single wavelength mode, wherein the currents in the straight rungs are a discrete representation of the ideal sinusoidal distribution of the surface current that would produce a uniform transverse RF magnetic field inside the cylinder.

The lowpass birdcage is one example of class of resonators capable of generating a uniform transverse magnetic field.  $N$  identical resonant elements spaced evenly around the cylinder such that they are coupled together will support the desired sinusoidal current distribution. The strength of the coupling determines how far apart the individual modes are split in frequency. In particular, the highpass version of the birdcage [4], which reverses the location of the capacitive and inductive elements, produces a uniform field at its highest frequency mode and the Nyquist mode at the lowest frequency. In both cases, the straight rung elements are coupled to one another by their mutual inductance and by the impedance of the ending element between each rung. In order to operate a body-sized birdcage inside a set of gradient coils, one must install an RF shield [5, 6] around the birdcage to prevent total detuning of the RF coil by the gradient coil. The shield is a necessary evil that greatly influences the operating frequency of the birdcage and degrades the coil sensitivity by producing a counter flowing current that diminishes the useful field in the sample volume and boosts the energy density of the field outside the coil.

Before I decided which version of the birdcage to use in the commercial Signa scanner, I investigated with Jim Bridges a third alternative. Starting with an empty shield, we installed sixteen elements shaped like towel racks spaced around the circumference. The bases of each towel rack were grounded to the shield and the straight segment was broken up by inserting several series capacitors. We called it a "coil in a can," but it later became more widely known as a TEM resonator [7]. We found that the mode spectrum was ordered in the same way as the lowpass birdcage but that the frequency separation between the uniform mode and the two-wavelength mode was only about half as large for the TEM as for the birdcage. This lower splitting occurs because the TEM resonator relies on only mutual inductive coupling compared to the additional coupling provided by the birdcage's ending impedance. I tried to increase the coupling by halving the space between the adjacent towel racks by building a 32-element version. This did not significantly increase the splitting. I was concerned that there was an increased risk of cross contamination of the homogeneous mode by the nearby second order mode when the coil  $Q$  was reduced by tissue loading. Hence, I abandoned the TEM resonator in favor of the lowpass birdcage. After Jim Bridges left GE and went to Advanced NMR, he overcame the small splitting issue by driving the "coil in a can" TEM resonator with a push-pull arrangement that does not couple to the second order mode [8]. The lowpass seemed to have a slight SNR advantage over the highpass birdcage when we were operating only in the horizontal linear mode. Unfortunately, the vertical mode of the lowpass behaved badly for large patients when we converted it to quadrature.

High Field Issues

The birdcage has served the MRI community well for nearly twenty-five years, but its utility is limited at fields higher than about 3T. As the progress in phased arrays and SENSE type coils demonstrates, a uniform reception sensitivity is definitely not needed. As a receive coil, a birdcage can be thought of as a multiple element dephased array. A voxel located at the center of a birdcage resonator emits its signal equally to all of the rungs and the birdcage responds by combining all these signals with equal weighting and properly phased to give the same SNR that a similar sized phased array would obtain. In contrast, for a voxel located far from the center, the signal is very large in the proximal rungs and relatively weak for the distal rungs. The birdcage responds to this mix of signals by weighting them all the same but dephasing them in such a way that there destructive interference among the strong signals until the net signal has the same SNR as the center voxel. Essentially because the birdcage lacks the additional detection hardware and combinational software, it dumbs down the high SNR possible in the near field of the phased array and delivers a middling SNR.

The birdcage cannot satisfy the requirements of a transmit coil at high frequency such as needed at 7T. It is hardwired to provide a RF field that is approximately uniform in the absence of a patient. When this field is applied to a patient, the dielectric and conductive properties of the tissue so greatly distort the field that its initial uniformity is irrelevant. In fact, one might hope for a coil configuration that purposely generates a very non-uniform field whose variations are chosen to counteract in part the distortions induced by the tissue properties. To accomplish this end, the transmit coil of the future would need to be made up of a large number of independent coil elements, each driven by its own small power amplifier, each controlled by programmable waveform generator. To determine how such an array would be programs, one could perform a coil calibration imaging sequence during prescan that characterized the interaction between each coil element and the specific patient.

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