



Magnetic Resonance Imaging RF Coils Systems Testing and Measurement

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Abstract

The radio-frequency (RF) detection coil is a transducer device that excites the nuclear magnetic resonance (NMR) signal and then detects it in the magnetic resonance spectroscopy (MRS) measurement. The signal-to-noise (SNR), spatial and temporal resolution, spatial uniformity, and the efficiency with which RF excitation power is used and lost to the measurement sample are all dependent on the RF coil's performance. The success of the MRS measurement is largely determined by the proper choice and application of the RF detection coil. There are many coil design choices available and we need to fit the best options for the specific application. Among coil design choices there are transmission line coils, wire coils (surface coils, volume coils), TEM coils. The types of

surface coils are single loop solenoids, multiple-loop arrays, and the types of volume coils solenoids, crossed pairs, saddle pairs, phased arrays, cage coils, Litz coils. TEM coils are coaxial lines, strip lines, and microstrips. TEM coils can be capacitively shortened and tuned; half-wave transmission line resonators are similar to the slotted tube resonator but with a number of practical features that lend themselves to modern utility. Typical features are where the center conductor 'rungs' of these coils generate a highly uniform field like rungs of a birdcage. The TEM coil can be double-tuned by tuning alternating rungs to alternating Larmor frequencies of the desired nuclei. The Larmor or precessional frequency in MRI refers to the rate of precession of the magnetic moment of the proton around the external magnetic field.

1. Magneto Resonance Gradient Loop Saddle Coil structure and functionality

The radio-frequency (RF) detection coil is a transducer device that excites the nuclear magnetic resonance (NMR) signal and then detects it in the magnetic resonance spectroscopy (MRS) measurement. The signal-to-noise (SNR), spatial and temporal resolution, spatial uniformity, and the efficiency with which RF excitation power is used and lost to the measurement sample are all dependent on the RF coil's performance. The success of the MRS measurement is largely determined by the proper choice and application of the RF detection coil. There are many coil design choices available and we need to fit the best options for the specific applications. Among coil design choices there are transmission line coils, wire coils (surface coils, volume coils), TEM coils. The types

of surface coils are single loop solenoids, multiple-loop arrays, and the types of volume coils solenoids, crossed pairs, saddle pairs, phased arrays, cage coils, Litz coils. A saddle coil system can be constructed by crushing two equal rectangular coils on a cylindrical shell. The same current is carried by each coil, and the direction of the flowing current may be set in Helmholtz or anti-Helmholtz configuration. The magnetic flux field ("magnetic field") is generated by a saddle coil system and is inspected in the whole space in two different spatial planes. The uniformity of the magnetic field is analysed by using Taylor series and the optimum geometry can be obtained from second-order Taylor expansions. The saddle coil geometric parameters are optimised for best performance. The deviations of the magnetic field which is produced by an optimum-compensating saddle coil system are constructed with typical NMR field-cycling magnetic field in homogeneities. The basic components of any MRI system: The main magnet produces the B_0 field, necessary to align the spins and achieve equilibrium. Gradient coils enable image encoding in the x, y, and z direction (frequency, phase, and slice-encoding directions, etc.). The RF coil is the part of the MRI system that excites the aligned spins and receives an RF signal back from the sample. All the components are controlled and interfaced with the user via a console (Fig. 1).

In the case that RF Rx coil is saddle coil the signal acquired using Rx coil is MR (Magneto resonance) signal that carries information about the imaging object in the magnetization form. We analyse dual frequency actively decoupled saddle coil. The actively decoupled saddle coil is made from copper wire (volume coil). There are four fixed value capacitors ($C_{A1}, C_{A2}, C_{B1}, C_{B2}$) which are distributed along the coil to maintain the constant current distribution within the coil. The end of the saddle coil is connected to variable capacitor (C_1)

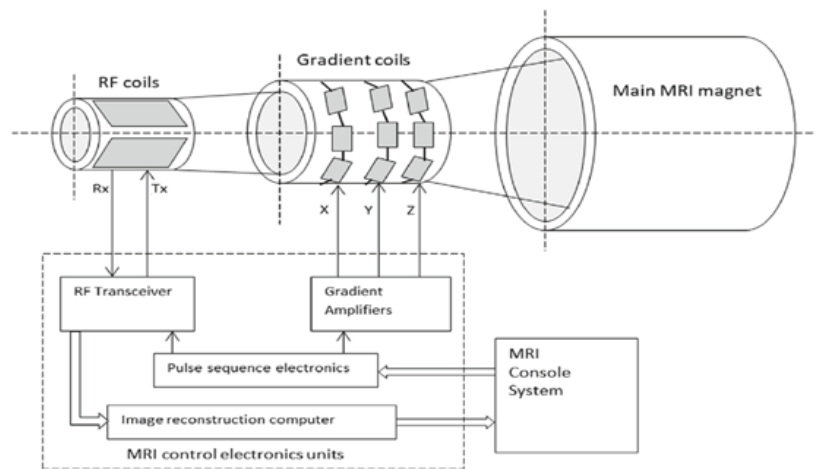


Figure 1: MRI scanner system (main MRI magnet, gradient coils, RF coils, control units and console system)

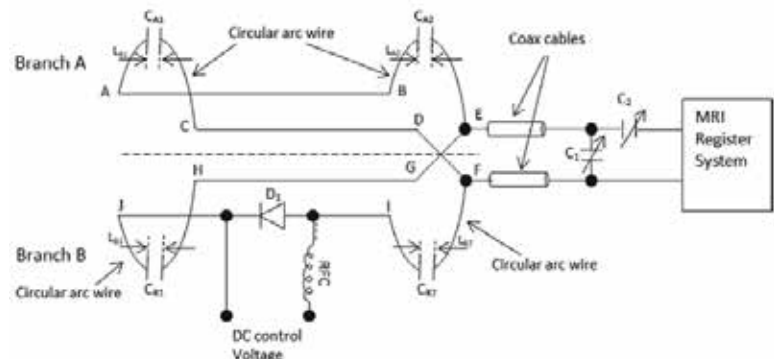


Figure 2: Dual frequency actively decoupled saddle coil schematics

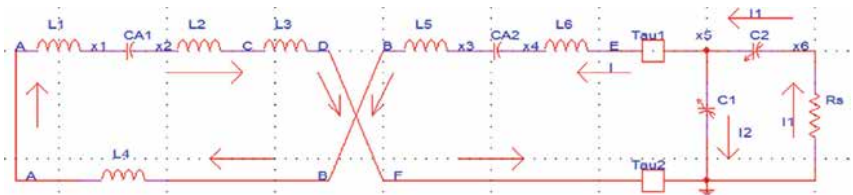


Figure 3: Half saddle coil (branch A), coax cables, and tuning networks and register system equivalent circuit

and another variable capacitor (C_2). The two capacitors implement type of resonator. The decoupling circuit is implemented by RF choke (RFC) C_{choke} and PIN diode (D_1) and it compatible with the dual frequency circuit. The RF choke (RFC) C_{choke} and PIN diode (D_1) are inserted in the leg of the sample coil and controlled by a separate coaxial cable. We can divide our volume saddle coil to two sub sections A and B and by supplying DC control voltage to RF choke (RFC)

C_{choke} and PIN diode (D_1) connects or disconnect the saddle coil section B (Fig. 2).

The two coax cables which connect the tuning capacitors (C_1, C_2) to our half saddle coil (branch A) are represented by two delay lines τ_1, τ_2 , respectively. We consider two coax cables represented delay lines are delayed in currents and not voltage. The Half saddle coil (branch A), coax cables, and tuning networks elements (C_1, C_2) and register system equivalent circuit are presented (Fig.3).

Assumption: The derivatives of I_1 and I_2 do not affected from the delay in time_{T_1}, T_2 .

We get a system set of four differential equations (neglecting delay in time_{T_1}, T_2):

$$\frac{dl}{dt} = I_x; \quad \frac{dl_1}{dt} = -\frac{1}{C_2 \cdot R_s} \cdot I_1 - \frac{1}{C_1 \cdot R_s} \cdot I_2; \quad \frac{dl_x}{dt} = \frac{1}{C_1 \cdot (\sum_{k=1}^6 L_k)} \cdot I_2 - \left(\frac{\sum_{k=1}^2 \frac{1}{C_{Ak}}}{(\sum_{k=1}^6 L_k)} \right) \cdot I$$

$$\frac{dl_2}{dt} = -\frac{1}{C_2 \cdot R_s} \cdot I_1 - \frac{1}{C_1 \cdot R_s} \cdot I_2 - I_x$$

2. PIN Diode Controlled Dual-Tuned MRI Coil Systems

Magneto Resonance (MR) of nuclei other than hydrogen is investigated. MRI uses the movement of protons within a magnetic field to generate an image. Within the constant magnetic field of an MRI scanner, tissues that contain free hydrogen nuclei (protons) generate varying signals when pulses of radiofrequency (RF) energy are applied to them. The common scheme of using trap circuits for dual-tuned operation results in increased coil losses as well as problems in obtaining optimal tuning and matching at both frequencies. By using PIN diodes, we switch the coil between two resonance frequencies. The dual-tuned coil design utilized PIN diodes to switch between different capacitance values for setting two difference resonance frequencies (Fig. 4).

The switching between lower resonance frequency and higher resonance frequency of the dual-tuned coil is done by switching ON and OFF PIN diodes D_1 . We consider that the PIN diodes (D_1) are ideal; when diodes D_1 are in OFF state then have very high resistance $R_{D1-OFF} \rightarrow \infty$, and ON state with very low resistance $R_{D1-ON} \rightarrow \epsilon$ (Fig. 5a/b).

The equivalent circuit for the lower resonance frequency (5a) and the equivalent circuit for the higher resonance frequency (5b) is presented.

The switching of D_1 diodes is done by bias voltages V_{Bias-1} and V_{Bias-2} . D_1 diodes get the same voltages V_{Bias-1} and V_{Bias-2} at the same time. D_1 diodes are ON state for $V_{Bias-1} > V_{Bias-2}$ ($V_{Bias-1} = +12v$, $V_{Bias-2} = v0$ (GND)) or $D1$ diodes are OFF

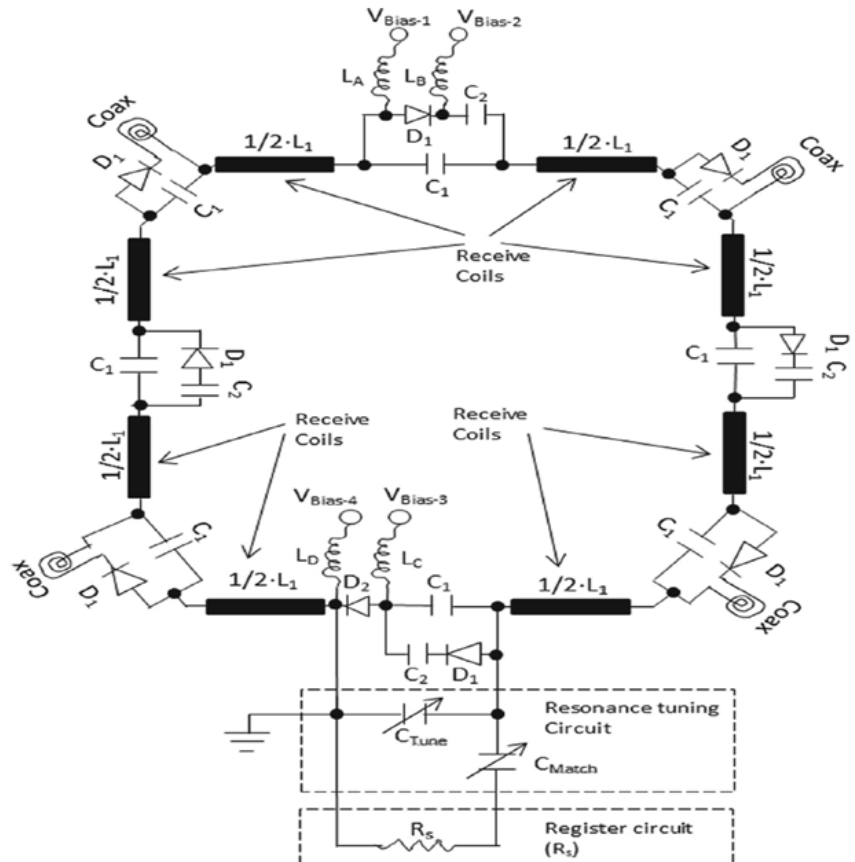


Figure 4: Dual-tuned coil with PIN diode switching

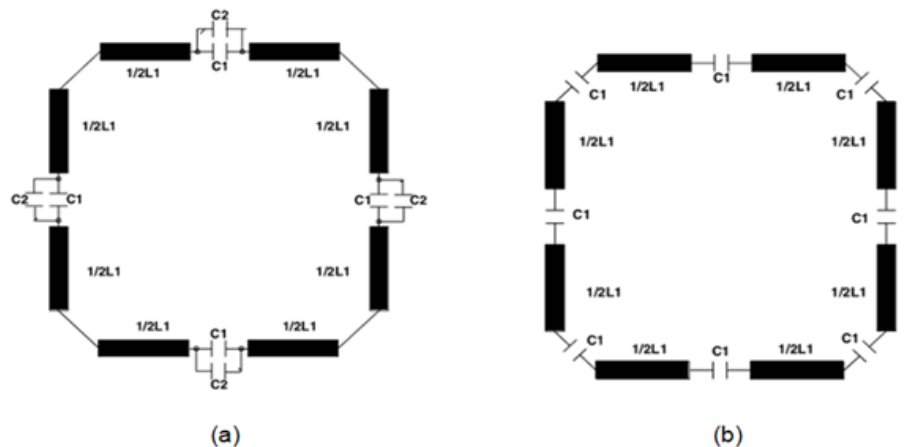


Figure 5a/b: Dual-tuned coil with PIN diode switching's equivalent circuit (a) lower resonance frequency, (b) higher resonance frequency.

state for $V_{\text{Bias-1}} < V_{\text{Bias-2}}$ ($V_{\text{Bias-1}} = -12\text{v}$ and $V_{\text{Bias-2}} = 0\text{v}$ (GND)). For low resonance frequency of the coil, D_1 diodes are at ON state. For higher resonance frequency, D_1 diodes are at OFF state. The receive coils in our dual tuned coil structure are rectangular cross section segments or thin tape segments (8 segments). For full functionality of dual tuned coil with PIN diodes switching system, we need to connect the coil structure to resonance tuning circuit and a register system which is represented by 50-ohm resistor (R_s). Diode D_2 gives us the ability to get full system work, D_2 need to be in OFF state (other option is ON state). D_2 is ON state for $V_{\text{Bias-3}} > V_{\text{Bias-4}}$ ($V_{\text{Bias-3}} = +12\text{v}$, $V_{\text{Bias-4}} = 0\text{v}$ (GND)) or OFF state for $V_{\text{Bias-3}} < V_{\text{Bias-4}}$ ($V_{\text{Bias-3}} = -12\text{v}$, $V_{\text{Bias-4}} = 0\text{v}$ (GND)). We consider D_1 diodes at the corners of dual - tuned coil structure is connected to circular coax transmission lines in serial.

The circular coax transmission lines function as delay lines for the case D_1 diodes at ON state (low resonance frequency), $V_{\text{Coax}} \rightarrow 0$; $I_{\text{Coax}}(t) \rightarrow I_{\text{Coax}}(t-\tau)$ which is delay (τ) in current flows through the coax where the voltage on the coax cables is very small. The circuit for Dual-tuned coil circuit for lower resonance frequency circuit (D_1 diodes are ON state (short) and D_2 is at OFF state (disconnected)) is presented (Fig.6).

We get the set of DDEs (Delay differential equations) for dual tuned coil system where D_2 is at OFF state and D_1 diodes are ON state (lower resonance frequency case).

We consider the coax transmission lines are functioned as short conductor then dual resonance frequency coil's corner capacitors are shorted

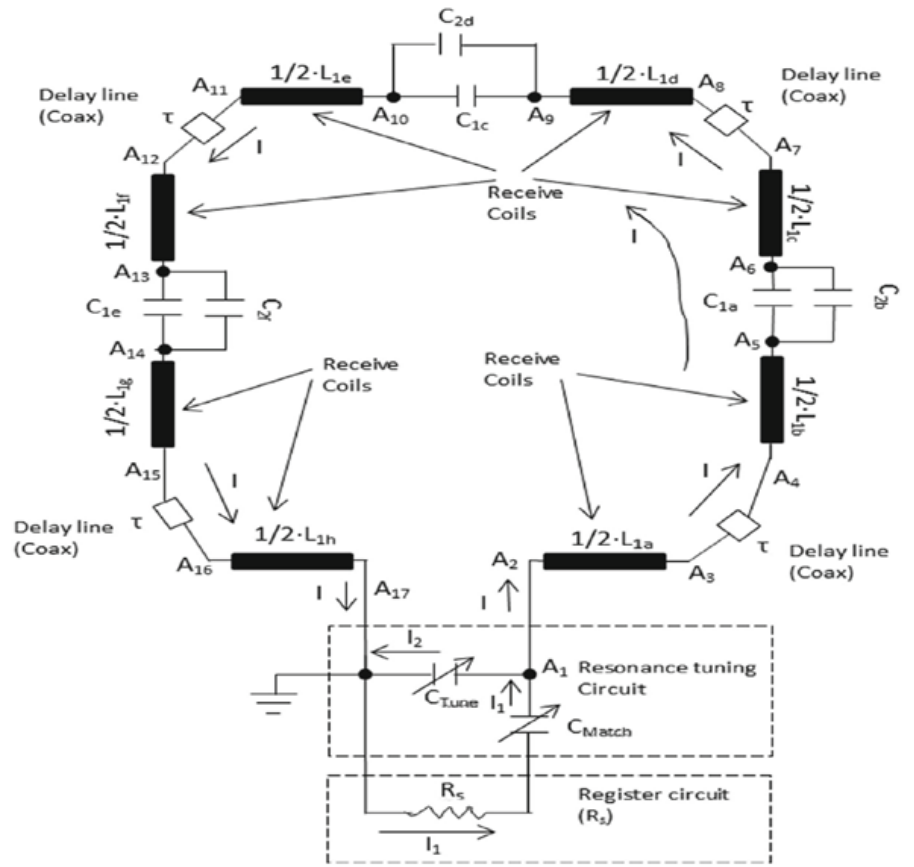


Figure 6: Dual-tuned coil circuit for lower resonance frequency circuit (D_1 diodes are ON state (short) and D_2 is at OFF state (disconnected)) is presented

We consider the derivations of

$$\frac{d^n}{dt^n} I_{\frac{1}{2}L_{1n}}(t - k \cdot \tau) \quad \forall n = 1, 2, 3; n \in \mathbb{N}$$

do not affect from the time delay

$$k \cdot \tau \quad \forall k = 1, 2, 3, \dots; k \in \mathbb{N}$$

$$\text{then } I_{\frac{1}{2}L_{1a}} = I'$$

$$\frac{dI'}{dt} = \frac{1}{4 \cdot L_1 \cdot C_{\text{tune}}} \cdot I_1 - \frac{1}{4 \cdot L_1 \cdot C_{\text{tune}}} \cdot I - \frac{1}{4 \cdot L_1 \cdot C} \cdot \sum_{k=1}^3 I(t - k \cdot \tau)$$

$$\frac{dI_1}{dt} = -\frac{1}{R_s} \cdot \left(\frac{1}{C_{\text{match}}} + \frac{1}{C_{\text{tune}}} \right) \cdot I_1 + \frac{1}{R_s \cdot C_{\text{tune}}} \cdot I; \quad \frac{dI}{dt} = I_1$$

Reference

O. Aluf, Advanced Microwave RF Antennas and Circuits, Nonlinearity Applications in Engineering, Springer, Edition 2, 2025, In two Volumes, ISBN: 978-3-031-58699-6, EAN: 9783031586996.

<https://link.springer.com/book/9783031586996>

$$L_{1a} = L_{1b} = L_{1c} = L_{1d} = L_{1e} = L_{1f} = L_{1g} = L_{1h} = L_1$$

$$C_{1a} = C_{2b} = C_{1c} = C_{2d} = C_{1e} = C_{2f} = \frac{1}{2} \cdot C; C = C_{1a} + C_{2b}; C = C_{1c} + C_{2d}; C = C_{1e} + C_{2f}$$