The shielding of RF MRI coils using double-sided EMI shield

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Abstract

We present, here, a device for shielding the RF MRI coils to eliminate their undesired radiation, and to restore the natural resonant frequency f_0 in the presence of a biological sample. It is presented for a planar surface RF MRI coil. Nonetheless, this approach is also valid for conformal coils as well.

Introduction

The influence of any biological sample on an RF MRI coil results in a shift of the resonant frequency f_0, which makes the receiver coil less sensitive at f_0. Different samples cause different frequency shifting. It occurs due to electric conductivity σ of the sample, which in presence of a time-varying field experiences **eddy currents**, which in turn induce fields that cancel the original magnetic field. Besides, a time-varying magnetic field also induces an **electric field**, which sets up a capacitive coupling between the sample and the RF coils. This introduces parasitic capacitances in the circuit, and thus alters the resonant frequency. The approach, described here, shields the effects of a variety of samples; restores the natural resonant frequency f_0 , and eliminates the undesired capacitive coupling among the phased array coils and other electronic parts.

The approach uses the **CTLM** (**C**oaxial **T**ransmission **L**ine **M**odeling) technique of shielding; a double-sided **EMI** (**E**lecro**M**agnetic **I**nterference) shield is constructed and filled with a ROHACELL substrate, with a very good **Loss Tangent** $\tan\delta$ and $\varepsilon_r=1$.

Firstly, one constructs a coaxial model whose inner conductor dimensions match those of the RF Coil. The shield (outer conductor) of the coaxial, which is usually of the order of a few skin depths, e.g. ≥ 10 , is then split longitudinally into two identical halves; after flattening, one serves as the upper shielding and the other as the lower shielding.

The upper shielding is to prevent radiation and mutual capacitive coupling, while the lower is to eliminate the negative effects of the sample on the RF coil. Both flattened halves are then soldered at the middle via a small bridging segment whose length equals the diameter of the coax. On the opposite of that connection a tiny gap of ≤ 1 mm is made. The shield sandwiches the RF coil and the substrate. There are two different ways of connecting the shield to the coil, which are:

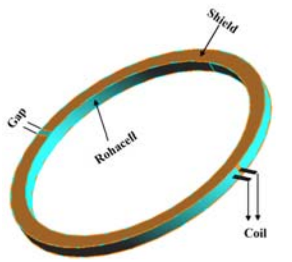

Fig.1 An unbalanced connection with a ROHACELL filling

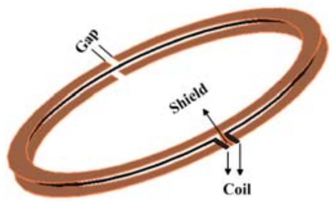

Fig.2 A balanced connection, the filling has been removed

- An unbalanced connection: one terminal of the RF coil is soldered to the bridging segment of the shield, and grounded, Fig.1.
- A balanced connection: only the bridging segment, which lies between the coil terminals, is grounded Fig.2.

In MRI, unbalanced connections are more frequently used; however, a balanced connection has many applications as well.

The most important thing is to have the shield gap opposite to that of the RF coil, to prevent the **common mode**, otherwise it affects the upper and lower shielding, and cancels out the induced signal in the RF coil.

When shielding is used, any induced eddy currents will flow on the shield's outer surface instead of the coil, and are eventually earthed. Therefore, the quality factor $Q=P_{stored}/P_{dissipated}$ is barely affected [1]. Any coil can be modeled as an RLC tuning circuit [1, 2]. The resonance effect occurs when inductive and capacitive reactances are equal, and the voltages (imaginary) developed across them cancel each other out. A peak of real voltage occurs at resonance.

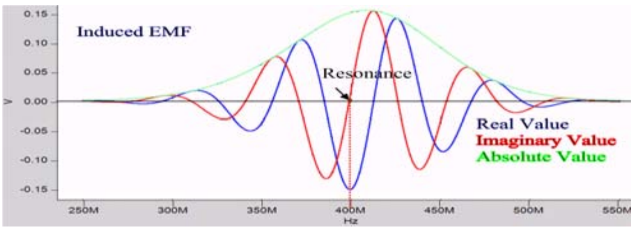

Fig.3 An unshielded coil, no sample introduced resonance at 400 MHZ

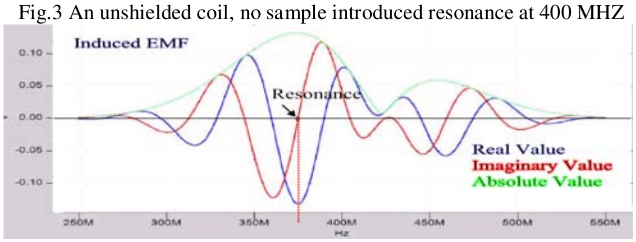

Fig.4 An unshielded coil, with a sample introduced resonance at 375 MHZ

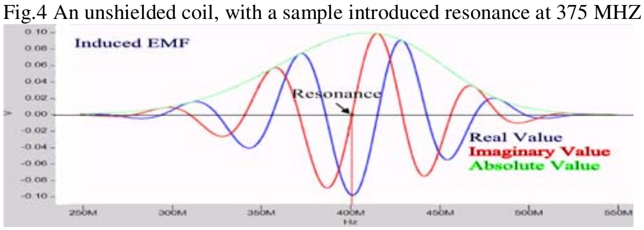

Fig.5 A shielded coil, with a sample introduced resonance at $\cong$ 400 MHZ

How does this approach work?

Poynting's theorem [2] is applied to a volume, v bounded by a closed surface, s. The complex power, P_s delivered by the source in v equals the sum of power, P_f flowing out of s (radiated), the time-average power, $P_{d(av)}$ dissipated in v, plus the time-average stored in v.

$$P_s = P_f + P_{d(av)} + j2(W_{m(av)} + W_{e(av)}) \quad (1)$$

The radiated power is

$$P_f = \frac{1}{2}\oiint_s \vec{E}\times\vec{H}^* \cdot d\vec{s} \quad (2)$$

The average dissipated power in v is

$$P_{d(av)} = \frac{1}{2}\iiint_v \sigma|E|^2 dv \quad (3)$$

The time-average stored magnetic power in v is

$$W_{m(av)} = \frac{1}{2}\iiint_v \mu|H|^2 dv \quad (4)$$

The time-average stored electric power in v is

$$W_{e(av)} = \frac{1}{2}\iiint_v \varepsilon|E|^2 dv \quad (5)$$

Here σ, ε, and μ are respectively the conductivity, permittivity, permeability of the sample. E, and H stand for electric and magnetic field intensities respectively. From Eq. (1) to Eq. (5), one notices that eliminating the electric field eliminates the radiation power and thus the capacitive coupling, which under an optimal circumstance implies that $P_s \approx W_{m(av)}$.

Numerical results

A generic, planar RF coil that has an f_0 of 400 MHz is constructed with a doubled-sided shield. A four layer cylinder models human tissues, skin, bones, and blood. The simulation was carried out using an FDTD numerical solver, SEMCAD X.

Fig. 3 depicts the situation when no sample is loaded, an unshielded coil is used, it resonates at 400 MHz, where a 150 mV real voltage (blue) is registered, and zero reactive voltage (red) is detected at f_0=400 MHz. Fig. 4 shows the influence of a sample on an unshielded RF coil, which resonates at 375 MHz, where a 130 mV signal is registered. Fig. 5 depicts the approach of restoring the original frequency in the presence of the sample, using the above described "double-sided EMI shield". It is seen to resonate at nearly 400 MHz., with a signal-peak of 112 mV. The shield attenuates the received signal a bit, yet it is an effective tool against frequency shifting.

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References: [1] The ARRL Handbook for Radio Communications 2008; [2] Stutzman et al, Antenna Theory & Design, JW (1981).