Resonance Shift Decoupling: A Potential Alternative to Low Input Impedance Preamplifiers

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Introduction

The use of phased-array coils has become desirable for efficient magnetic resonance imaging in both clinical and preclinical biological investigations. Multi-channel receivers are now commonplace, allowing the use of phased-array coils with many elements. Coupling becomes increasingly problematic with more elements, degrading the independence of coil sensitivities required for parallel imaging [1]. This ultimately results in reduced image SNR. Geometric overlap-decoupling can be used for adjacent elements; however non-adjacent elements require loop current minimization. Low input impedance (Z_{in}) preamplifiers have traditionally been used for this purpose [2]. In this work, we tested an alternate method involving a transistor amplifier that is built into the coil to minimize loop current and maximize gain, while simultaneously achieving low-noise performance.

Theory

The noise performance of a receive chain is dominated by the noise figure of the first stage (F_1) if a large first-stage gain (G_1) is achieved. This is evident from the following cascade equation describing the overall system noise figure F:

$$F = F_1 + \frac{F_2 - 1}{G_1} + \frac{F_3 - 1}{G_1 G_2} + \frac{F_4 - 1}{G_1 G_2 G_3} \cdots$$

where F_i represents the noise figure and G_i represents the gain of the i^{th} stage. Although the first stage of the cascade is typically considered the preamplifier, it is actually the coil itself: the noise is associated with sample and coil loss and the gain is achieved by the resonance phenomenon.

Loop current minimization techniques require a large coil impedance to minimize induced magnetic flux that results in coupling. Thus an ideal array element would have infinite loop impedance, infinite gain, and minimal noise figure. An ideal amplifier has these attributes, so connecting an untuned loop to it would produce the desired result (Fig. 1a). Unfortunately, such an amplifier does not exist and thus other methods must be investigated. A transistor is a reasonable approximation to an ideal amplifier (Fig. 1b) since it has a large transconductance (G) and a low input capacitance, C_{gs} (i.e. large Z_{in}). However, there exist specific impedance values as seen from the gate-source transistor terminals that can optimize the design for either high-gain (Z_s) or low-noise (Z_{opt}) operation (Fig 1c).

Methods & Materials

An initial dual-turn surface coil that consisted of the inductive loop and a DC biased transistor (FHC40LG) (as in Fig. 1b) was manufactured using a LPKF ProtoMat C100/HF for use on a 7.0 T Bruker Biospec MRI system. Because the transistor match was not optimized, noise performance suffered. We therefore extrapolated Z_s and Z_{opt} parameter values from the data sheet for 300 MHz operation. Constant noise and gain circles were plotted with the Smith-Chart software (Fig. 2). Breaking capacitors were chosen to achieve a compromise between large gain and low-noise operation, which is shown as point 1 in Fig. 2. When the loop was combined with the transistor, however, the loop and the coil structure formed a resonance frequency at nearly 300 MHz (point 2 in Fig. 2). This design results in a good gain and noise figure, however the resonance near the Larmor frequency has the undesirable effect of a large loop current. Preliminary phantom images were acquired with improved SNR. All of the low-noise transistors that were investigated had similar requirements for Z_s and Z_{opt} for the frequencies associated with current MRI field strengths.

The circuit design was modified to achieve reduced loop current. Another compromise of Z_s and Z_{opt} was set to point 2 in Fig. 2, which corresponds to the self-resonance frequency of the dual-turn coil. When the transistor was added this time, C_{gs} added with the equivalent coil capacitance (C_{tune}) to shift the resonance to a lower frequency and thus reduced the loop current, resulting in a resonance shift decoupled (RSD) coil element (Fig. 3). Loop currents of the RSD coil and a coil terminated with a commercially available low-Z_{in} preamplifier were compared via S_{21} bench measurements. Phantom images were acquired to compare the noise performance.

Results & Discussion

The low-Z_{in} preamplifier method produced 16 dB of decoupling while the RSD coil that was tuned to 284 MHz achieved 13 dB of decoupling. This indicates that the described design has the potential to provide effective element decoupling for arrays with a large number of elements. This would allow the cost and complexity of phased-array design to be substantially reduced. However, the aggressive shift in the resonance frequency beyond that due to C_{gs} alone compromised noise performance. The dominating factors in the RSD coil performance however are ultimately the transistor properties such as minimum noise figure, C_{gs}, and the location of Z_{opt} and Z_s for the Larmor frequency of interest. Further design and component optimization may provide both effective decoupling and outstanding noise performance, greatly simplifying and reducing the cost of MRI phased-array design.

References

[1] Pruessmann KP, et al. *Magn Reson Med* 42: 952-962, 1999. [2] Roemer PB, et al. *Magn Reson Med* 16: 192-225, 1990.

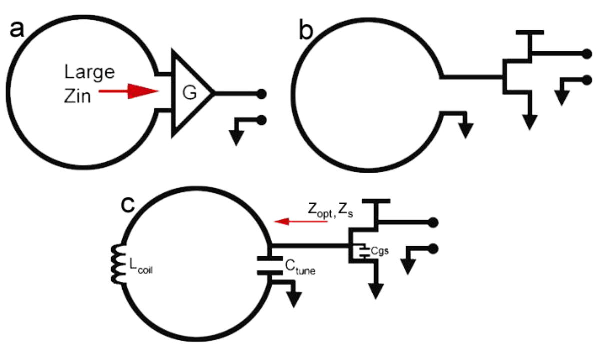

Fig. 1. Various steps in the theoretical design process.

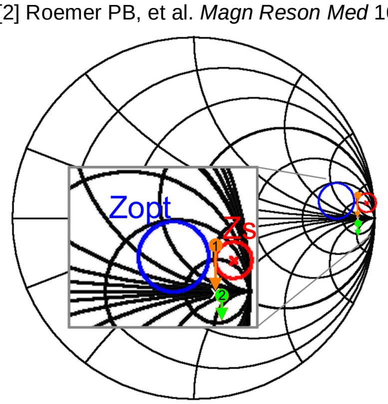

Fig. 2. Smith chart plots of noise and gain impedance match values.

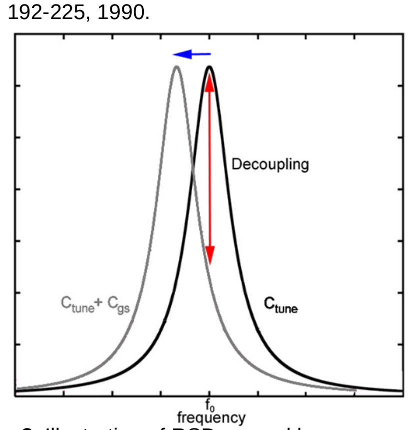

Fig. 3. Illustration of RSD caused by connecting a self-resonant loop to a transistor.