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Noise properties of a NMR transceiver coil array

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Abstract

The use of multiple radiofrequency (RF) surface coil elements has applications in both fast parallel imaging and conventional imaging techniques. Through implementation of a simple magnetic decoupling network, 50 Ω matching can be achieved in both the transmitter and receiver chains, enabling the use of conventional RF power amplifiers and preamplifiers for transceive applications. Unlike phased array coil arrangements using low impedance preamplifiers for decoupling, the noise correlation between 50 Ω coils decoupled with discrete components has not been characterized. We have measured the dependence of coil quality factor (Q-factor) and noise correlation on coil separation and shown these quantities to be consistent with theoretical arguments, at least at 4 T (170 MHz). Our results suggest that a coil system for transmission and reception of NMR signals with 50 Ω coils can be built to take advantage of all the benefits of conventional array coils and with the added advantages of using conventional amplifiers.

Keywords: Transceive; Surface coils; Phased array; Decoupling; Noise correlation

1. Introduction

Surface coils have long been a popular choice of MRI receiver coil due to their ability to trade off field-of-view (FOV) for increased signal-to-noise ratio (SNR) in the received signal. This increased SNR is a direct result of the increased sensitivity (defined as the ratio of the magnetic field to the current which generates it) in close proximity to the surface coil. This sensitivity decreases with distance from the coil, following a power-law dependence. This spatially varying sensitivity of a surface coil used as a receiver is commonly referred to as its sensitivity profile. By the principle of reciprocity, such a surface coil used for transmission would have the same sensitivity profile for excitation, and if used in transmit/receive mode, the total sensitivity is multiplicative. The introduction of multiple receiver coils [1] (commonly referred to as a phased array) allowed larger volumes to be covered with increased image SNR through the judicious recombination of simultaneously acquired signals from independent receivers interfaced to these coils. Through the use of multiple surface coil sensitivity profiles, fast parallel imaging techniques (such as SMASH [2] or SENSE [3]) may also be implemented. The placement of surface coil elements within the array will influence the effectiveness of the parallel imaging approaches. It has been shown that minimizing the FOV overlap between the individual surface coil elements results in an improvement in parallel imaging performance, as this gives more orthogonal sensitivity profiles [4]. The separation between surface coil elements that optimizes parallel imaging, will in general, have no overlap (to achieve more orthogonal sensitivity profiles). As a result, surface coil arrays that are SENSE optimized will rarely have an overlap of 10% to provide natural decoupling between nearest neighbour coils [1] (see Section 2.1). Therefore, the predominant electrical engineering problem in array design for parallel imaging lies in eliminating this coupling between surface coil elements.

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Parallel imaging can be achieved in transmit [5], receive [2], or potentially in transceive mode. Parallel imaging with transcieve coils has the potential to improve speed-up factors by virtue of a higher order power-law dependence of the magnetization profile as mentioned earlier. However, the coupling between multiple surface coils is a critical issue in both transmitting and receiving the NMR signal. The prevalent method of decoupling coils utilizes low input impedance preamplifiers to effectively open each surface coil element to minimize the current and interference between all coils within the surface coil array [1,4]. Noise correlation between such coils has been well characterized but these parallel imaging arrays are restricted to receive-only mode. It has also been shown [6] that decoupling networks can be used to eliminate coupling between a surface coil pair. Applying this technique to parallel imaging arrays provides a method to implement transceive applications without coil placement restrictions and maintains the use of conventional 50 Ω RF power amplifiers and preamplifiers. However, the noise correlation properties of coils decoupled in this manner have not been explored. The goal of this study is to characterize the effectiveness of a decoupled coil pair as a potential building block for parallel and conventional imaging arrays. The Q-factor of each coil, as well as noise correlation between them is measured to characterize the transceive properties of the coil pair and to obtain the noise correlation matrix between two elements decoupled in such a way.

2. Theory

A brief description of the various mechanisms of surface coil coupling and the electrical engineering methods used to minimize their effects are required. Both individual surface coils [7,8] and surface coil arrays [1,9] have had extensive theoretical analysis in the past. To begin, we consider a simple square surface coil pair loaded with a conductive spherical phantom. An alternating current in one coil will generate a back electromotive force (*emf*) that will induce a counter-alternating current in the other coil if there is mutual inductance between the coil pair. The positive feedback nature of this coil–coil interaction (characterized by the magnetic coupling coefficient k_m) splits each coil's single resonance peak into two about the original resonance frequency as shown in Fig. 1 [2]. This coil–coil interaction couples the signal and the noise between each coil and is commonly referred to as *extrinsic* coupling. A second coupling mechanism between the coils arises through random ionic motion within our conductive sample, which simultaneously induces an *emf* in each coil. This *emf* will vary in phase and magnitude between the two coils depending on the distance from each coil to the ionic current element (see below). This coil–sample coupling is commonly referred to as *intrinsic* coupling and produces correlated noise in each coil. Coil–sample coupling also increases the total dissipated power of the coil pair, thereby decreasing the SNR.

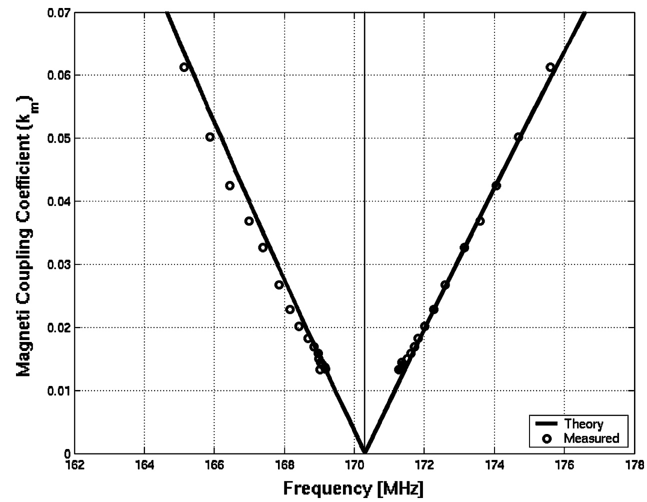

Fig. 1. The resonance peak splitting between a surface coil pair as a function of their normalized separation (*d*/*L*) compared to results from theoretical simulations (magnetic coupling coefficient estimated by surface integral of vector potential [1], and resonance splitting through ABCD network model [9]).

2.1. Decoupling methods

There are three common methods for minimizing *extrinsic* coupling in practice. First, natural decoupling can eliminate the mutual inductance by overlapping the coil pair by approximately 10% ([1, p. 208], "Fig. 9 shows the coil overlap where the mutual inductance goes to zero (coil separation about 0.9 d), but the correlated noise (ke) does not"). Although this design provides transceive capability, it is restricted to a unique separation and cannot be optimized for fast parallel imaging techniques. Second (as previously mentioned), low input impedance preamplifiers may be used to reduce the current in each coil by effectively opening each coil at its feed [1,4]. This method reduces the effects (specifically cross-talk) of *extrinsic* coupling to negligible amounts (see p. 198, Eq. (4) in [1]), yet it does not eliminate the mutual inductance between the coils. This technology is also limited to receive-only mode since current is required to generate a magnetic field within the patient. Finally, decoupling networks can use conventional low-loss reactive circuit components to counter the electrical effects of mutual inductance. It was shown by Wang [6] that a capacitor network with equal and opposite reactance to the mutual inductance can be used (as shown in Fig. 2) to eliminate the effects of this magnetic coupling. This method provides transceive capability for arbitrary element separation.

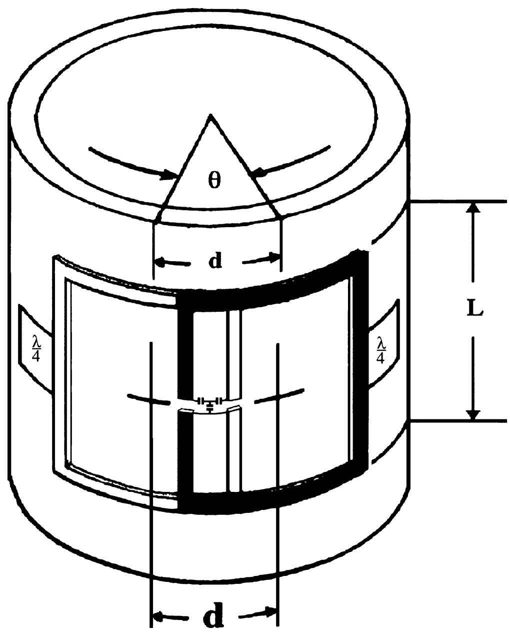

Fig. 2. The apparatus consists of a square surface coil pair conformed to subtend 60° about an acrylic cylindrical shell (od 17.78 cm).

Furthermore, this design maintains the use of commercially available 50 Ω preamplifiers and transmit RF power amplifiers.

2.2. Noise correlation

When two signals (v_1 and v_2) with stochastic noise (denoted by ˆ) are added, the total noise power (σ_f) will be greater if there is noise correlation (σ_{12}, also commonly called covariance):

$$\hat{f} = \hat{v}_1 + \hat{v}_2, \tag{1}$$

$$\sigma_f^2 = \sigma_1^2 + \sigma_2^2 + 2\sigma_{12}. \tag{2}$$

The optimal combination of signals from multiple receiver coils to attain maximum SNR has been thoroughly examined. Duensing and colleagues determined a method of recombining coil signals such that the *extrinsic* coupling was not detrimental to SNR. However, there was no method they could determine that eliminated the detrimental effects of *intrinsic* coupling [10]. As mentioned, the source of this noise correlation is due to common current paths between the coil pair within the patient and can be characterized by an increase of patient loading (transfer impedance [7]). The increase in dissipated loss that is attributed to increasing noise correlation should correspond to a decrease in each coil's Q-factor.

The correlation coefficient (r) is ideal for quantifying the noise correlation between two coils, and can be expressed as the following (using Bra-Ket notation to denote the inner product):

$$r = \frac{\sigma_{12}}{\sqrt{\sigma_1^2\sigma_2^2}} = \frac{\langle \hat{v}_1 - \bar{\hat{v}}_1 | \hat{v}_2 - \bar{\hat{v}}_2 \rangle}{\sqrt{\langle \hat{v}_1 - \bar{\hat{v}}_1 | \hat{v}_1 - \bar{\hat{v}}_1 \rangle \langle \hat{v}_2 - \bar{\hat{v}}_2 | \hat{v}_2 - \bar{\hat{v}}_2 \rangle}}. \tag{3}$$

It is precisely this statistic that is used in constructing the noise correlation matrix (Ψ_{ij}) [3] to account for coupling between receivers when reconstructing sensitivity encoded images. By employing phase sensitive quadrature demodulation, we are able to obtain real (Re) and imaginary (Im) components of our signal. There exists a relative phase between each channel that will maximize the noise correlation between similar channels ($r(\mathrm{Re}_1, \mathrm{Re}_2)$ and $r(\mathrm{Im}_1, \mathrm{Im}_2)$) for common-mode signals. It should be noted that the remaining *intrinsic* correlated noise power within real and imaginary channels of the system is phase shift invariant (contrary to [7]).

3. Materials and methods

3.1. Coil construction

Each square surface coil is constructed of 50 μm thick copper tape with a width (w) of 1.27 cm. The dimension of each coil side (L) is 10.64 cm, which is constant through out this study. Coil dimensions were chosen so as to subtend 60° on the outside surface of a 17.78 cm id, 20.32 cm od acrylic cylinder as shown in Fig. 2. The distance between the center of each coil (d) is used as a measure of separation, similar to the approach used in [1]. The inter-element spacing between the coil pair is zero when $d = L + 2w$ ($d/L = 1.24$). Quarter wave baluns [11] are used to isolate coils from shield ground. Without proper balancing the transmission line shield voltage potential would rise above ground potential due to direct conductive coupling, and the shield would act as an antenna and increase the amount of dielectric loss [12] to the sample as well as the noise correlation detected in the system. The 50 Ω quarter wavelength baluns are matched to each coil on opposing sides as shown in Fig. 2. To match each coil to the 50 Ω balun, a parallel to series capacitance ratio method is used [13] for each coil separation. The required matching and balancing network reduces the loaded Q-factor of each coil from 19.6 to 8.9. These measurements are made using a HP4396A network/spectrum/impedance analyzer and shielded RF field probes (in the case of the matched and balanced configuration, a single RF probe is used to receive and the coil itself is used to transmit).

Each surface coil is tuned, matched, and balanced in isolation prior to decoupling. The decoupling network is placed at the virtual ground between the coil pair as shown in Fig. 2. The electrical components of this decoupling network are shown in Fig. 3A. As the inter-element spacing increases, 20 AWG copper wire is used to connect the decoupling capacitance (C_m). Additional tuning capacitors are used to counteract the

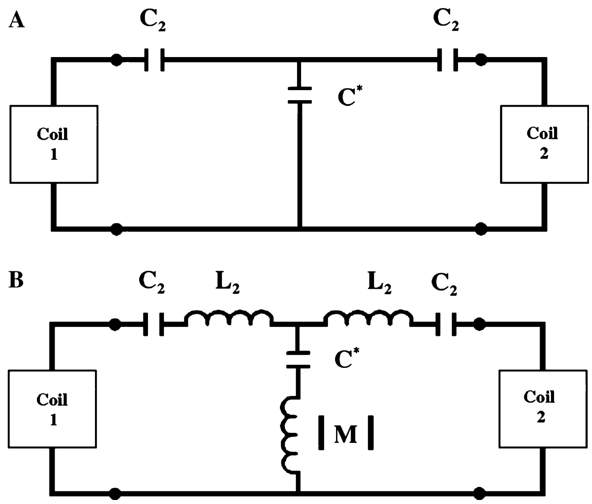

Fig. 3. Schematic of magnetic decoupling network employed between the surface coil pair, as shown in Fig. 2. (A) Physical schematic of components used. (B) The equivalent circuit accounting for inductive coupling (M), and increase in inductance of the coil (L_2) due to the 20 AWG connective wire.

additional reactance (L_2) caused by these connecting wire segments. The decoupling capacitor is always centered between the coils, which made the tuning capacitance (C_2) symmetric as shown in Fig. 3. Since the Q-factor decreases upon loading (but resonant frequency does not shift), the decoupling network is attached prior to loading to ensure greater accuracy in tuning C_m. Distributed low-loss ceramic capacitors are used on each coil to reduce dielectric loss such that there is no resonance frequency shift upon loading with the spherical phantom. This is similar to many multi-purpose surface coil designs where retuning to compensate for loading is undesirable or impractical. Furthermore, the decoupling capacitance is inversely proportional to the square of the operation frequency. If detuning were allowed to occur, the decoupling network would not eliminate all magnetic couplings. This design also provides more accurate estimation of the decoupling capacitance (C_m) required through tuning in the unloaded case (providing a greater Q-factor). A 4% saline 17.78 cm od, 17.14 cm id phantom is placed in the center of the cylinder for mimicking head loading and is aligned with the coil pair for symmetry upon varying coil separation. If the phantom was offset from center the loading of the coil would vary as a function of coil separation further complicating the matching of the coil pair at every separation.

3.2. Measurements

A 4395A HP network/spectrum/impedance analyzer was used to obtain Q-factor measurements from the bench-top. The Q-factor is measured as the ratio of the resonance frequency to the full width at half maximum. Due to the large decrease in Q upon matching and balancing, the Q-factor measurements are made without baluns attached to provide a more accurate Q-factor measurement relative to the error.

Noise correlation between the two coils was measured as a correlation coefficient from time series data. The noise correlation data were acquired for 1 s and repeated five times for each separation. All noise correlation data are obtained within the Faraday cage shield of the MR scanner room. However, the apparatus was placed outside of all imaging gradient fields to eliminate any system-fluctuation induced noise correlation. Each coil was attached to a Varian broadband preamplifier (15–400 MHz) before being sampled with a 250 kHz bandwidth centered at 170.3 MHz using a 4 T Varian Unity INOVA console.

4. Results and conclusions

The peak splitting of resonance modes of the coil pair follows the predicted response from simulations as shown in Fig. 1. Resonance splitting was measured every 1.27 cm about the cylinder, until an angle of π was reached between the coil pair. The concordance with simulations confirms that the estimated magnetic coupling used to confirm the relationship of decoupling capacitance C_m is accurate. The decoupling network proposed by Wang [6] was effective in decoupling the surface coil pair. It was shown by Roemer et al. [1] that the magnetic coupling can be both positive and negative depending on whether or not the coil overlap is less or greater than the natural decoupling overlap, respectively. The decoupling capacitive reactance (X_{C_m}) required to eliminate resonance peak splitting at each separation was indeed found to be equal and opposite to the reactance of the estimated absolute mutual inductance (X_{M}) such that the introduction of inductive decoupling was not required. The cross-correlation between X_{C_m} and X_{M} was −96.5%, ($p = 1.6\mathrm{E}{-6}$, confidence interval $\alpha = 0.95$, $N = 11$).

A decoupling network was tuned for each separation whereupon Q-factor measurements were made. The Q-factor of a single loaded coil in isolation was measured to be 19.6, whereas, the Q-factor of the coils when they were naturally decoupled (~10% overlap) was 18.6. The decrease in Q-factor of the *extrinsically* decoupled coil pair can be attributed to the increased mutual impedance the two coils share due to *intrinsic* coupling. The Q-factor as a function of coil separation is shown in Fig. 4.

The correlation coefficient was calculated with both phase and temporal delays. Although there was a relative phase difference between each coil that maximized the correlation coefficient (the recorded result), there was no noise correlation that was greater or more significant (p value) when temporal delays are introduced.

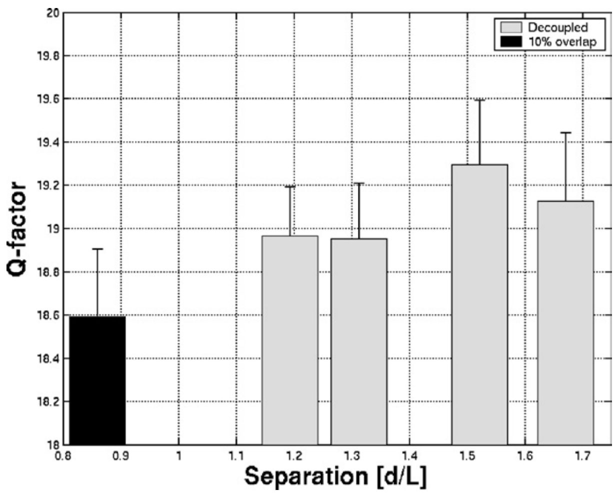

Fig. 4. Quality factor vs. normalized separation (d/L) between the loaded surface coil pair.

The measured noise correlation as a function of normalized coil separation is shown in Fig. 5. For natural magnetic decoupling (~10% overlap) the noise correlation was finite (8%) and due entirely to *intrinsic* coupling. As the two coils depart from the position that gives rise to zero mutual inductance, the *extrinsic* coupling between them drastically increases the noise correlation between the (now) coupled coils. In contrast, the noise correlation between the decoupled coils monotonically decreases. The magnetically decoupled r curve has identical characteristics to k_e curves of similar coils [1,14], and shares no characteristics with its k_m curve, suggesting that it represents the noise correlation generated solely from *intrinsic* coupling. Furthermore, both the real and imaginary channels behave exactly the same way as expected.

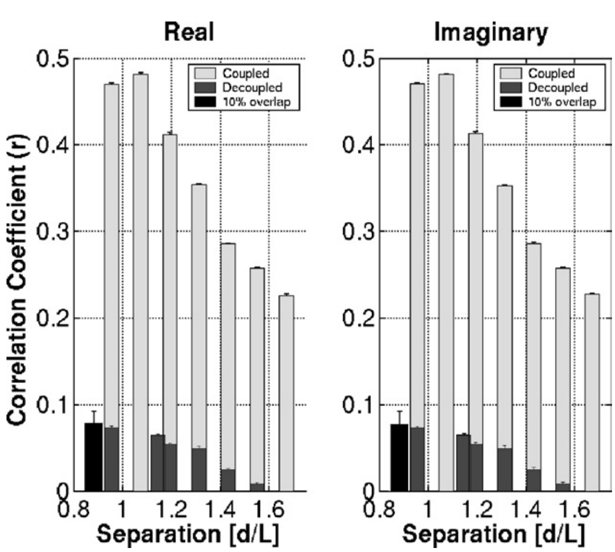

Fig. 5. Noise correlation vs. normalized separation (d/L) between the loaded surface coil pair.

5. Discussion

Extensive prior work has been done in theoretically describing noise correlation between a magnetically decoupled surface coil pair [7,14–17]. We have shown that both *extrinsic* and *intrinsic* coupling influences noise correlation between a surface coil pair in the presence of a lossy high dielectric constant spherical sample. The existence of noise correlation due entirely to *intrinsic* coupling between a surface coil pair has been experimentally confirmed by employing a purely capacitive decoupling network. The spatial dependence of this noise correlation has been measured and shown to share similar characteristics with the k_e curve as presented by Tropp [14].

All measurements characterizing the transceive capability of this coil pair reflect similar trends. The Q-factor increases as noise correlation decreases with increased coil separation. This is an expected result since the source of noise correlation between coils arises not only due to coil–coil *extrinsic* magnetic coupling (which has been eliminated through implementing a decoupling network), but also from *intrinsic* coupling arising due to an increase in sample loading. For conventional sum-of-squares reconstructions, the noise correlation variation with separation also presents a method for optimizing phased arrays, since *intrinsic* noise correlation's effect on SNR reduction cannot be eliminated [10].

The ability to magnetically decouple a surface coil pair through use of decoupling networks provides the ability to transmit and receive independently without the use of low input impedance preamplifiers while allowing arbitrary coil separation and not introducing additional noise correlation. These three characteristics of such a decoupling strategy make this surface coil pair a useful building block for a transceive array provided the next nearest neighbour *extrinsic* coupling is negligible. For long linear arrays, next nearest neighbour coupling will likely be low for most human sized coil designs. For 2-D arrays, decoupling networks joining all nearest neighbours in pair-wise fashion can be adjusted in an iterative manner (or deterministically from a knowledge of the k_m matrix).

The most important point of this experiment is that the noise correlation properties of coils decoupled with this capacitive scheme are similar to those using low impedance preamplifiers, a fact not previously explored. This allows for the use of 50 Ω transmitter amplifiers and 50 Ω receiver preamplifiers to build parallel imaging optimized arrays. At magnetic fields of 4 T and higher in humans, it is useful to have the ability to tailor the RF amplitude and phase in a controlled manner, and transceive arrays allow exactly that. We are in the process of characterizing such an 8 element head coil, where each element is driven with phase increments of a $2\pi/8$ rad, as well as with adjustable amplitudes. This degree of tailored excitation is not possible with conventional birdcages. Furthermore, the ability to transmit and receive using the coil elements imposes a stronger axial dependence on each coil's sensitivity than in a typical receive-only phased array. This stronger spatial dependence has appealing characteristics for the implementation of parallel imaging.

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