New matching networks for coil and preamplifier

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Introduction: MRI signal is usually detected with a 50 Ohm-matched coil (conductor loop together with a matching network (=CMN)) and amplified by a preamplifier for further processing. The preamplifier has another matching network (=PMN) in order to transform the 50 Ohm input impedance to the optimal source impedance Z_{opt} of the transistor [1]. This is based on the assumption that the investigated object and preamplifier are the dominant noise sources in this case [2]. However, with receiving coils in a coil array becoming increasingly smaller, the noise contributed from their matching networks cannot be neglected [3]. Here we present two methods to simplify CMN and PMN, thereby the noise contributed from the matching networks is reduced. This approach is highly relevant to micro imaging and coil arrays with a large amount of small coil elements.

Method: The *normally used matching method (NM method)*, which is common for coil and preamplifier, is shown in Fig 1 (top). The CMN is between interface A1 and B1 and the PMN is between interface C1 and D1. The conductor loop is on the left side of interface A1 and the preamplifier is located between interface C1 and E1. Between the coil and preamplifier, a phase shifting network transforms the preamplifier's low input impedance $Z_{in} = (R_{in} + j * X_{in})$ into a low resistance $R_S(\approx R_{in})$, while the 50 Ohm of the coil is kept. This is shown on the interfaces B1 and C1 in Fig 1. The resonance of L_{11} and C_{12} at the working frequency supplies a large impedance Z_L in the coil loop needed for array decoupling. Note that it is not necessary to delete the reactance of Z_{in}. Considering a preamplifier whose first stage is made by a FET, X_{in} is normally inductive and R_{in} is a very small resistance. Being equivalent to an inductor (inductance=X_{in}) and a very small resistor (resistance=R_{in}), Z_{in} can resonate with CMN directly, creating large impedance Z_L and omitting the phase shifting network. This is achieved by replacing the matching inductor L_{11} in NM method with a capacitor, while the PMN is not changed and the new CMN still matches the coil loop to 50 Ohm. This method is named *capacitor matching method (CM method)* and is shown in Fig 1 (bottom left). This CM method can be further simplified by integrating C_{23} and C_{24} into one capacitor C_{33}, with $1/C_{33} = 1/C_{23} + 1/C_{24}$, while the other variable capacitors are changed to fixed capacitors too. This is not a mere simple integration but means that CMN and PMN are combined into an integrated matching network (IMN) and the 50 Ohm interface is omitted. This *integrated capacitor matching method (ICM-method)* is shown in Fig 1 (bottom right).

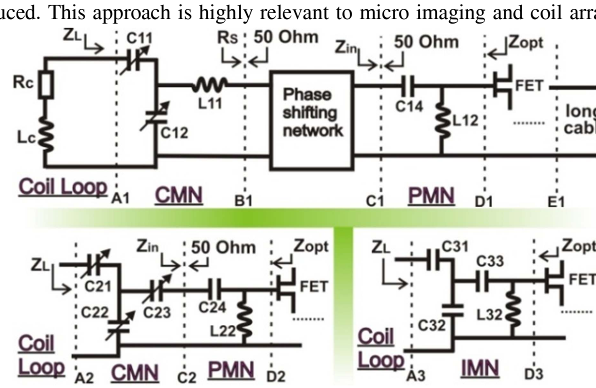

Fig 1. The comparison between the NM method (top), CM method (bottom left) and ICM method (bottom right).

Experiments: The NM, CM and ICM methods were compared through self-made coil loops (side length=2.3cm) and self-designed preamplifiers (GaAs FET, double stages, $gain = 28dB, N_F = 1dB, Z_{in} = 1.2 + 14.3j$), which allows the changes in its PMN. A low-pass π matching network was chosen for phase shifting in NM method. During the integration from CM to ICM method, the trimmer values were obtained with the help of a calibrated conductor loop and two overlap-decoupled pickup coils. These two pickup coils were also used for S21 measurement for preamplifier decoupling evaluation. MRI experiments were done in a 3T Siemens scanner (MAGNETOM Trio, Siemens Healthcare, Germany) on a cubic phantom (distilled water +1 .955g/L $CuSO_4$ + 3.6g/L NaCl). Images were obtained from single channel to compare the SNR difference of different matching methods with FLASH sequence (TR/TE=600/10ms, flip angle 25°, FOV=150X150 mm, 3mm X 35 slices). Two ICM matched coils at a distance of 1.3 cm were used to show the preamplifier decoupling performance with SE sequence (TR/TE=500/10ms, flip angle 90°, FOV=135X135mm with 3mm slice), configured to save uncombined images from different channels. Image SNR was calculated by MATLAB (Mathworks, Natick, Massachusetts, USA) with mean signal from 2000 points divided by standard deviation of noise measured outside of the phantom. To compare the SNR improvements of CM and ICM methods at different coil sizes, square coils with side length of 0.5cm, 1cm, 1.5cm, 2cm and 2.5cm were simulated in ADS (Advanced Design System 2011, Agilent Technologies), with the assumed trimmer quality factor of 200 and fixed capacitor quality factor of 400.

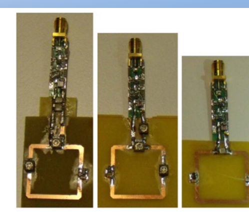

Fig 2. Photos of three matching methods with related loops and preamplifiers. From left to right: NM, CM and ICM

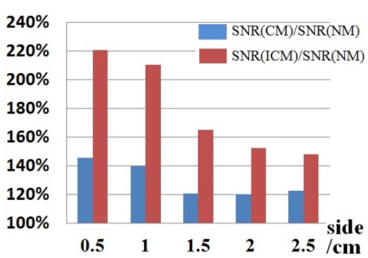

Fig 3. SNR improvements of two new matching methods on square coils with different side lengths.

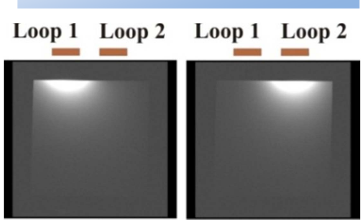

Fig 4. Uncombined images from a two coil array matched by ICM.

Results: The coil loops matched in NM, CM and ICM methods with preamplifiers were fabricated and shown in Fig 2. The decoupling coefficients measured with two pick up coils were 25dB, 26 dB and 32dB for NM, CM and ICM methods, respectively. The SNR comparison of the images obtained from the three methods are: SNR(CM)/ SNR(NM)= 138% and SNR(ICM)/ SNR(NM)= 167%, on average at different coil-phantom positions. The simulated SNR comparison between two new methods and NM method are shown in Fig 3. The uncombined images from a two coil array are show in Fig 4.

Discussion and Conclusion: Compared to the NM method, CM and ICM methods lead to higher image SNR. As every passive component contributes noise, smaller number of components in CM and ICM means less noise contribution. Besides this reason, the SNR increase from CM to ICM method originates from the fixed capacitors, which always have higher quality factors than the trimmers. The 6dB improvement of the decoupling also resulted from the higher quality factor. There is almost no coupling between two loops, as can be seen from uncombined phantom images recorded with a two-coil array (see Fig. 4). As seen from Fig 3, the SNR improvement of the two new methods increases as coil size decreases. Unfortunately, the CM and ICM methods cannot be used for coil loops whose body noise resistance is larger than $X_{in}^2/50$ because the reactance of the capacitors would become positive. However, this limitation can be ignored since the CM and ICM method's impact on image SNR decreases for large coils, where NM method should be preferred.

In conclusion, the CM and ICM matching methods are helpful for building small receive coils. In addition, the IMN method is a first step towards the design of the coil and the preamplifier as a combined system for better image performance and miniaturization of the MR receiver.

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Reference: [1] Roemer, P. B., W. A. Edelstein, C. E. Hayes, S. P. Souza, and O. M. Mueller. "The NMR phased array." Magnetic resonance in medicine 16, no. 2 (1990): 192-225. [2] Hoult, David I., and R. E. Richards. "The signal-to-noise ratio of the nuclear magnetic resonance experiment." Journal of Magnetic Resonance (1969) 24.1 (1976): 71-85. [3] Kumar et al., "Noise figure limits for circular loop MR coils." Magnetic Resonance in Medicine 61, no. 5 (2009): 1201-1209.