A method to calculate the noise factor of the receive coil matching network

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Introduction: In ordinary MRI experiments using a local receive coil, major noise sources in signal reception are the sample and the preamplifier [1]. An additional noise contribution from the coil matching network (=CMN) is typically small compared to these sources and is usually ignored. Receive-array coils allow for high local SNR and improved imaging speed. However, with larger numbers of receive channels, and consequently smaller sizes of individual coil elements, noise contributions from the CMN also need to be taken into account [2] [3]. Up to now, a quantitative description of a CMN noise contribution for practical consideration does not exist. Here, we present a method to calculate the noise factor of CMN. Approximations are also developed to allow the estimation of a CMN's noise factor on the RF bench. Based on the calculations, a method to reduce the CMN noise factor is presented.

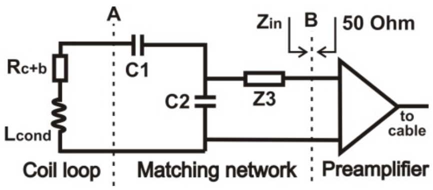

Fig1. The coil model with its matching network consisting of C1, C2 and Z3.

Method: A representative circuit diagram (coil model), which consists of the coil conductor loop, CMN and preamplifier, provides the basis for our calculations and approximations and is displayed in Fig 1. The conductor loop receiving MR signal, is represented by inductance L_{cond} and resistance R_{c+b}. R_{c+b} is the sum of conductor loop resistance R_c and body noise resistance R_b. The equivalent model of the conductor loop is assumed to have two power sources. One source is for original signal induced from the changing magnetization in the sample, while the other only creates original noise from thermal noise of R_{c+b}. Although the original signal and noise degrades in CMN, their ratio remains preserved. So the original signal degradation can be expressed by the calculation of the original noise degradation. The original noise degradation is obtained with preamplifier input impedance $Z_{in} = 50\ Ohm$, which is correct for any Z_{in} since the CMN noise factor is not with respect to Z_{in}. Based on the above considerations it is possible to obtain the signal and noise power at interface B and then acquire an analytic expression for CMN noise factor.

Numerical calculations for verity the analytic results were performed using MATLAB (Mathworks, Natick, Massachusetts, USA). MRI experiments were performed in a 3T Siemens scanner (MAGNETOM Trio, Siemens Healthcare, Germany) with gradient echo (GRE) sequences (TR/TE=570/10 ms, pixel bandwidth: 200 Hz, flip angle: 15°, FOV=20x20 cm). Working with fixed phantom conductivity, the different SNR of the images obtained with the same conductor loop ($\emptyset = 2.5$cm) and same preamplifier (Siemens Healthcare, Erlangen, Germany) reflects different CMN noise factor. The conductivities of the phantom ($CuSO_4$+NaCI+distilled water in a cubic container) were varied to represent different body noise resistance values.

Results: As the full analytic solution is difficult to be displayed here, following simplifications can be made. In the CMN shown in Fig 1, Q_{C1}, $Q_{C2}$and Q_{Z3} are the quality factors of components C_1 , C_2 and Z_3 (can be capacitor or inductor), respectively. The sum of the reactance of C_1 and C_2 is written as X_{C12}. The reactance of the conductor loop at working frequency ω_0 is written as $X_{cond}(= \omega_0 L_{cond})$. Assuming high quality factors $Q_{C1} = Q_{C2} = Q_C \gg 1$, and similar Q_{Z3} and Q_{C2}, the N_F of the CMN can be calculated analytically. Neglecting small quantities of second and higher order, the noise factor can be approximated by

$N_F(Z_3 \text{is indcutor}) \approx [-2L_{cond}(2/Q_C + 1/Q_{Z3})\Delta\omega + (R_{c+b} + Q_C X_{C12})]/R_{c+b}$ and

$N_F(Z_3 \text{ is capacitor}) \approx [-2L_{cond}(2/Q_C - 1/Q_{Z3})\Delta\omega + (R_{c+b} + Q_C X_{C12})]/R_{c+b}$.

Here $\Delta\omega = \omega_r - \omega_0$ and ω_r is the coil loop resonance frequency when no preamplifier is connected. ω_r can be measured by two pick-up coils decoupled by overlap.

Consider a coil loop matched to 50 Ohm at 123 MHz, with $X_{cond} = 50$ nH. Calculated analytical results of the noise factor N_F with respect to $\Delta\omega/\omega_0$ are shown in Fig 2. The main plot shows N_F with respect to different R_{c+b} with $Q_C = 100$ and $Q_{Z3} = 67$, while the insert shows N_F with respect to different Q_{Z3} with $R_{c+b} = 0.1$ Ohm and $Q_C = 100$. The right part of the triangle in each curve is Z_3 being an inductor and the left part is Z_3 being a capacitor. Points having $X_{c1} < 0$ were discarded, therefore different curves cover different ranges of $\Delta\omega/\omega_0$. As seen in the main plot, our approximations (black dotted lines) represent exact solutions (red solid line) very well. Also, MR images were recorded from phantoms with different conductivity, which means different body noise resistance. Using a least squares fit, slopes of the experimental SNR change with $\Delta\omega$ for different conductivities and chosen Z3 were estimated. It was found that the experimental results matched well with the calculated theoretical results shown in Fig 2.

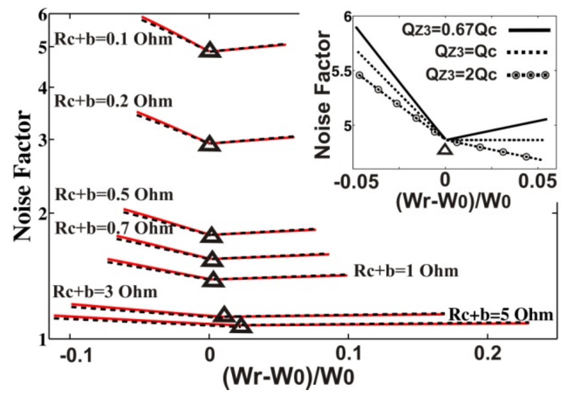

Fig 2. The noise factor of CMN with respect to $\Delta\omega/\omega_0$. Approximations: Black dotted line, Exact analytical calculations: Red solid line.

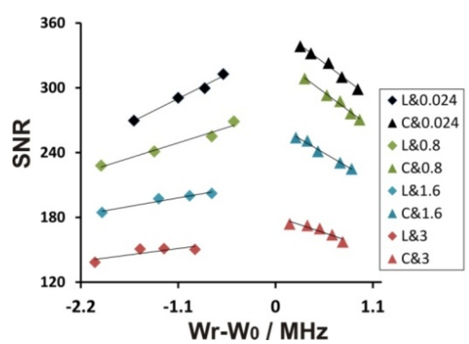

Fig 3. Image SNR with respect to $\Delta\omega$. The label, e.g. 'L&0.024' means that the image is obtained from the coil whose Z_3 is inductor and the phantom conductivity is 0.024 S/m.

Discussion and Conclusion: Obtained formulas are useful for coil construction by allowing the estimation of the CMN noise factor N_F on RF bench examinations. As shown in Fig 2, CMN N_F is far larger than the preamplifier $N_F(\approx 1.1$ for 0.5dB) in the case of very small R_{c+b} and should not be ignored anymore. It was found that the CMN N_F changes with $\Delta\omega$, therefore it is possible to reduce the CMN noise factor by **A**djusting $\Delta\omega$, the **D**ifference between coil loop resonant **F**requency and coil working **F**requency, to the optimal $\Delta\omega_{opt}$ (abbreviated as **ADFF**) at which the CMN will contribute the least possible noise. Relations of the CMN noise factor and different component quality factors (see the insert of Fig 2) also help to find $\Delta\omega_{opt}$. It is possible to prove that the commonly used decoupling method with low input-impedance preamplifier has $\Delta\omega$ very close to $\Delta\omega_{opt}$. As seen from Fig 2 and Fig 3, the smaller R_{c+b} is, the larger SNR improvement can be obtained through the ADFF method. Therefore, the method is expected to be very helpful in optimizing small receive coils, as needed in arrays with a high number of channels, or for micro imaging application.

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Reference: [1] Keil, Boris, et al. "A 64-channel 3T array coil for accelerated brain MRI." Magnetic resonance in Medicine (2012). [2] M.Pavan, et al., " Noise contributions in receive array coils." Proc. ISMRM 2012, p433. [3] Kumar et al., "Noise figure limits for circular loop MR coils." Magnetic Resonance in Medicine 61, no. 5 (2009): 1201-1209.