Capacitive approach to restore decoupling between channels for four-element MR coil array

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Multi-channel coil arrays are increasingly being used to improve Signal-to-Noise Ratio (SNR) in Magnetic Resonance Imaging (MRI). The decoupling between coils is an important parameter in array design. Indeed coupling between elements affects resonance frequency of each single coil and decreases it sensitivity. Many solutions were developed to achieve decoupling between elements of multi-channel coil array. In this paper, we present a capacitive solution to restore channel decoupling of a specific four-channel receiver coil array using common conductors. The principle of an effective decoupling was first demonstrated by circuit simulations of |S|-parameters. A receive-only four-channel coil array was design for rat head MRI. Experimental |S|-parameter measurements validated the proposed capacitive approach by restoring decoupling between elements and particularly between external loops.

Introduction: The Nuclear Magnetic Resonance (NMR) multi-channel coil arrays introduced in 1990 [1] are used to increase sensitivity of MR coils while preserving a large volume of exploration. The basic idea is to use the juxtaposition or overlapping of several single loops. These loops are usually connected to low input impedance and low noise preamplifiers to reduce the coupling between the different elements of the array and to preserve SNR respectively. Recent works described other decoupling techniques such as strip transmission line array [2], overlap geometries [3] and capacitive decoupling networks [3, 4]. A different solution for two-channel coil arrays was also described. In this setup, single elements of the array are not juxtaposed or overlapped but elements are electrically jointed with a common conductor [5, 6, 7]. Based on this specific approach, a newly designed four-channel coil array was proposed [8].

Principle: The equivalent electrical circuit of a four-channel coil array based on common conductor architecture is presented in Fig.1 where inductors are representing conductor sections. At the operating frequency, C_d and C_c capacitance values are chosen to compensate imaginary impedance parts of the L_d and L_c inductance values, respectively. At the operating frequency, without L_{gnd} parasitic inductances, these perfect compensations of impedance allow zeros of transmission between channels by deporting the ground plane to the black points labelled 3. This virtual ground plane between loops ensures the decoupling between channels [7, 8]. The three parasitic inductors L_{gnd} coming from the experimental design of the coil array deteriorate zeros of transmission and lead to coupling between elements. In this paper, we propose a solution to restore the decoupling between elements by adding a C_{gnd} capacitor in series with the central L_{gnd} inductance. This solution is illustrated in "*Simulated results*" section with circuit simulations of Fig. 1 topology and in "*Experimental results*" section with measurement of a four-channel coil array described in "*Design*" section.

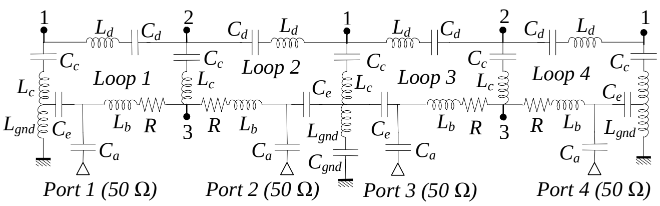

Fig. 1 *Equivalent electrical circuit of a four channel coil array with common conductors.*

Design: A specific four-channel receiver coil array was built on a flexible substrate and glued on a cylinder to fit the animal morphology. Loop inductors were realized with 35 µm thick and 4 mm width section of copper tape. Each element of the coil consists in a rectangular loop with 20x24 mm^2 internal and 28x32 mm^2 external dimensions. To decouple the receiver array from the transmitter coil, each loop integrates an active decoupling circuit made with two DH80055 PIN diodes. Four 50 Ω BNC cables were soldered at the four loops inputs to connect the coils with the acquisition data cabinet integrating transmit/receive switches and preamplifiers of MR system. The C_a and C_e were chosen to be adjustable elements for the four-channel coil array. Matching of each loop to 50 Ω at f_0=300 MHz (proton Larmor frequency at 7 T static magnetic field) was realized using varicap diodes BB149. The capacitance values of these diodes ranged from 2 pF to 22 pF for a bias voltages between -30 V and 0 V. Fixed capacitors are non magnetic case A series 100 ATC capacitors. All fixe capacitors are experimentally adjusted to minimize all $|S_{ij}|$ dB parameters. A photograph of the coil array is shown in Fig. 2.

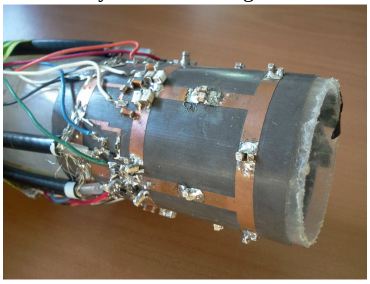

Fig. 2 *Photograph of the four channel coil array built for rat head.*

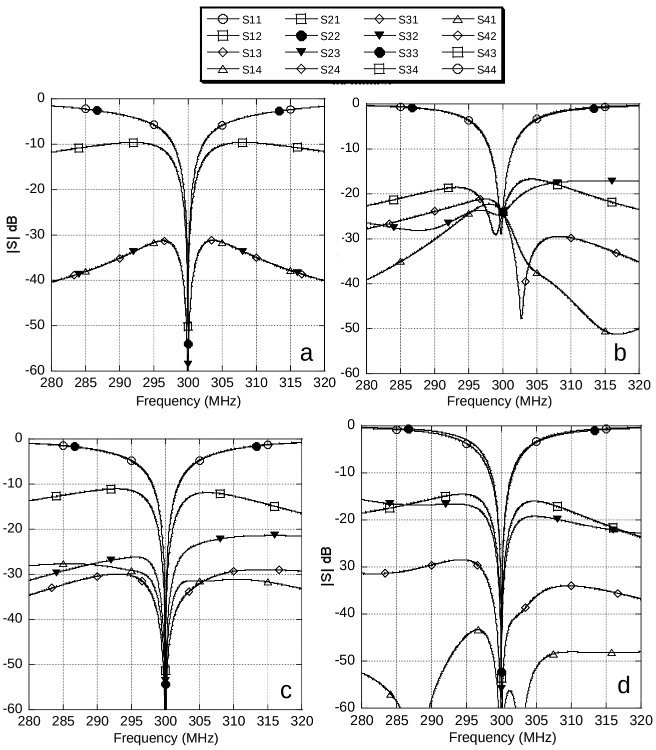

Fig. 3 S*imulations of the four-channel coil array.*
a Without parasitic inductance
b With parasitic inductances L_{gnd} fixed to 10 nH
c With capacitors C_{gnd} in series with L_{gnd} fixed to 28.1 pF
d With one central capacitor C_{gnd} in series with the central L_{gnd} and fixed to 28.1 pF

Simulated results: Circuit simulations realised with Designer software. The influence of L_{gnd} and C_{gnd} component values on the $|S|$ parameters is shown in Fig. 3. Simulations were achieved for an arbitrarily chosen value of coil inductors and for a 300 MHz operating frequency. The equivalent electrical circuit of Fig. 1 was simulated with all L_b, L_c, L_d and L_{gnd} values equal to 10 nH; R was fixed to 1.1 Ω. To compensate imaginary part of inductors at the operating frequency, capacitors C_c and C_d were fixed to 28.1 pF. C_a and C_e capacitors were fixed to 4.5 pF and 24.2 pF respectively. Fig. 3*a* shows the ideal coil with perfect decoupling between channels at the operating frequency without L_{gnd}-C_{gnd} components: all $|S_{ij}|$ parameters are lower than -30 dB at the operating frequency. Fig. 3*b* presents the decoupling deterioration due to the three 10 nH parasitic inductances L_{gnd}: $|S_{ij}|_{dB}$ parameters are increased at the operating frequency and curve minima are shifted. Fig. 3*c* and Fig. 3*d* present the restoration of decoupling between channels by additional C_{gnd} capacitors equal to 28.1 pF. It can be noticed on simulated results that a single C_{gnd} capacitor on central section (Fig. 3*c*) is sufficiency to restore an acceptable decoupling. This solution was chosen for practical realisation of the array.

Experimental results: |*S*|-parameter measurements of the four-channel coil array were carried out with an Agilent E5071C four-port VNA (Agilent Technologies Inc., Santa Clara, CA, USA). Results of coil array described in "*Design*" section are presented in Fig. 4*a*. $|S_{ii}|$-parameters perform better than -20 dB and characterizing each coil matching at the resonant frequency. $|S_{14}|$ parameter value is about -7 dB which is not sufficient to decouple external loops. This $|S_{14}|$-value is accentuated by magnetic coupling compare to circuit simulation (Fig 3.b). All other coupling parameters are inferior to -16 dB at the centre frequency. In Fig. 4*b* a 24.8 pF capacitor is added in series with the central L_{gnd} of the coil array. At the operating frequency, the $|S_{ij}|$-parameters are below -20 dB. These values allow the use of the coil array for MR imaging without coupling between channels.

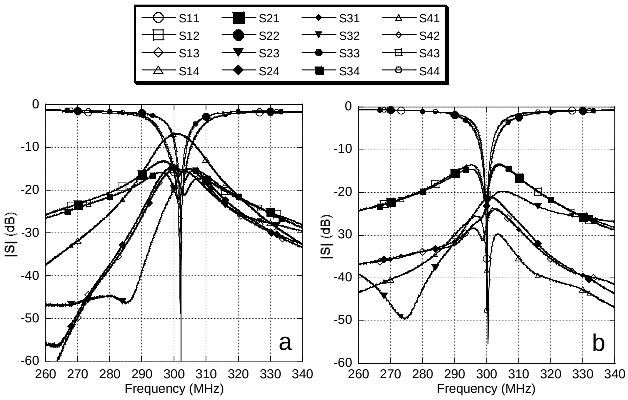

Fig. 4 |*S*| *parameters measurements of the four-channel coil array.*
a Without C_{gnd}
b With one central capacitor C_{gnd} = 24.8 pF

Conclusion: A capacitive solution to achieve decoupling between elements of a particular four-channel coil array based on common conductors was first demonstrated by circuit simulations of an equivalent electrical circuit. The solution was validated by the experimental |*S*|-parameters measured on four-channel coil array dedicated to rat head: a coupling lower than -20 dB was obtained between elements at the operating frequency.

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