Volume coil based on hybridized resonators for magnetic resonance imaging
C. Jouvaud, Redha Abdeddaim, B. Larrat, Julien De Rosny

To cite this version:
C. Jouvaud, Redha Abdeddaim, B. Larrat, Julien De Rosny. Volume coil based on hybridized resonators for magnetic resonance imaging. Applied Physics Letters, American Institute of Physics, 2016, 108 (3), pp.023503. <10.1063/1.4939784>. <hal-01284853>
Volume coil based on hybridized resonators for magnetic resonance imaging
C. Jouvaud, R. Abdeddaim, B. Larrat, and J. de Rosny

Citation: Appl. Phys. Lett. 108, 023503 (2016);
View online: https://doi.org/10.1063/1.4939784
View Table of Contents: http://aip.scitation.org/toc/apl/108/2
Published by the American Institute of Physics

Articles you may be interested in
Magnetic resonance imaging using linear magneto-inductive waveguides
Journal of Applied Physics 112, 114911 (2012); 10.1063/1.4768281

Analysis of the resolution of split-ring metamaterial lenses with application in parallel magnetic resonance imaging
Applied Physics Letters 98, 014105 (2011); 10.1063/1.3533394

Nonlinear split-ring metamaterial slabs for magnetic resonance imaging
Applied Physics Letters 98, 133508 (2011); 10.1063/1.3574916

Experimental demonstration of a $\mu = -1$ metamaterial lens for magnetic resonance imaging
Applied Physics Letters 93, 231108 (2008); 10.1063/1.3043725

Wireless power transfer inspired by the modern trends in electromagnetics
Applied Physics Reviews 4, 021102 (2017); 10.1063/1.4981396

Wireless power transfer based on dielectric resonators with colossal permittivity
Applied Physics Letters 109, 223902 (2016); 10.1063/1.4971185
Volume coil based on hybridized resonators for magnetic resonance imaging

C. Jouvaud,1,2 R. Abdeddaim,1,3 B. Larrat,1 and J. de Rosny1
1ESPCI ParisTech, CNRS, PSL Research University, Institut Langevin, 1 rue Jussieu, F-75005 Paris, France
2CEA, DAM, GRAMAT, F-46500 Gramat, France
3Aix-Marseille Université, CNRS, Centrale Marseille, Institut Fresnel, UMR 7249, Avenue Escadrille Normandie-Niemen, 13397 Marseille Cedex, France

(Received 9 November 2015; accepted 30 December 2015; published online 12 January 2016)

We present an electromagnetic device based on hybridization of four half-wavelength dipoles which increases the uniformity and the strength of the radio-frequency (RF) field of a Magnetic Resonant Imaging (MRI) apparatus. Numerical results show that this Hybridized Coil (HC) excited with a classical loop coil takes advantage of the magnetic hybrid modes. The distribution of the RF magnetic field is experimentally confirmed on a 7-T MRI with a gelatin phantom. Finally, the HC is validated in vivo by imaging the head of an anesthetized rat. We measure an overall increase of the signal to noise ratio with up to 2.4 fold increase in regions of interest far from the active loop coil. © 2016 AIP Publishing LLC. [http://dx.doi.org/10.1063/1.4939784]

Imaging objects and bodies with non-invasive techniques is a challenging field in physics. In this area, Magnetic Resonance Imaging (MRI) is one of the major diagnostic modalities based on the quantum spin of nuclei.1 Under a static magnetic field \( B_0 \), spins tend to align either positively or negatively, giving rise to a positive net magnetization along \( B_0 \). At equilibrium, this magnetization is constant and aligned with \( B_0 \). When spins are excited by a transient transverse radio-frequency (RF) magnetic field \( B_1 \), generated by a coil at the Larmor frequency, they start to precess around \( B_0 \) at this resonance frequency. Precession occurs at the Larmor frequency inherent to each atom’s nucleus. Transitions occur between the up and down spin states and spins start to precess, all in phase, giving rise to a net magnetization in the plane transverse to \( B_0 \). At the end of the emission, the nuclei spins relax and emit a decaying RF magnetic field at the Larmor frequency. The image is constructed following spatial encoding of the radiofrequency signals by the transient application of magnetic field gradients. The quality of an image defined by many interdependent variables (pixel size, contrast, signal to noise ratio (SNR), and homogeneity) is driven by all IRM parameters: sequence, gradients, and geometry. The transmit and receive RF apparatus impact the signal to noise ratio and the homogeneity of the image, because they mainly govern the B1 field strength and distribution. Two main types of coils have been proposed to emit and receive the B1 field: the surface coil showing strong SNR but bad homogeneity (Helmholtz coil) and volume coils which mainly interact with the B1 field, generated by a single loop coil. We propose a simple electrodynamic model to describe the hybridization between 4 resonators yields two modes that mainly interact with the B1 field, generated by a single loop coil. We propose a simple electrodynamic model to describe the hybridization. Numerical simulations are performed using CST Microwaves Studio to survey the effect of coupling between the excitation coil and the hybridized coil (HC). This HC is validated experimentally and numerically on a phantom made of gelatin. Finally, we show how HC enhances the image quality of a rat head compared to a loop coil.

A structure composed of several resonators presents different eigenmodes. This is due to the strong near field coupling between the resonators. Fundamental modes of the individual resonators are then hybridized. To build our magnetic coil, we apply this effect to electric dipole resonators. First, let us consider a resonator which mainly interacts with the electric field and shows a fundamental mode at the angular frequency \( \omega_0 \) (see Fig. 1). When two such resonators are close enough, the fundamental mode of the two resonators is split into two modes that resonate at two different frequencies \( \omega^+ \) and \( \omega^- \). Due to mirror symmetry, one of the eigenmodes is symmetric, while the other one is antisymmetric. Contrary to the symmetric mode, the currents in the resonators of the antisymmetric mode are opposite in direction. Consequently, this mode can be regarded as magnetic. In our case, this antisymmetric mode is found at a lower frequency...
The magnetic field of the system is tuned to 300 MHz and matched to 50 \( \Omega \) using a matching network circuit. The 3-cm diameter loop coil generates a magnetic field that is mainly parallel to the (Oy) axis. The magnetic field generated by the loop coil alone is shown in Fig. 2(a). The magnetic field drops off quickly. Under a quasi-static assumption, the magnetic field decreases as \( 1/\sqrt{R^2 + d^2} \), where \( R \) is the radius of the loop, and \( d \) is the distance from the loop axis. When the loop is in the near field of the four resonators, we clearly observe two strong resonances on the reflection parameter. Indeed, due to the antisymmetry plane of the system with respect to the (Oyz), only the eigen-modes \( (\pm\pm\pm) \) and \( (\pm\pm\mp) \) can be excited. The magnetic field pattern of the first resonance in the plane (Oxy) and (Oyz) is shown in Figs. 2(c) and 2(d). We observe, as expected, that the first resonance corresponds to the magnetic mode \( (\pm\mp\mp) \). The field in the region inside the 4 resonators is much more homogeneous than the one obtained with the coil loop alone in the transverse section (Oxy). Also, the field is now extended over all the resonator length in the longitudinal sections (Oyz) and (Oxz).
Experimental validations of HC are performed in a 7 T small animal MRI scanner (Pharmascan, Bruker Biospin, Germany) equipped with 300 mT/m gradient coils. The HC is used in transmit-receive mode. The hybridized coil is made out of thin metallic rods and tubes, composed of non-ferrous materials: pure copper. Four resonators are attached inside a PMMA cylinder to form a right parallelepiped (see Fig. 3(a)). The length of the rods is roughly 375 mm.

Experimental validations of HC are performed in a 7 T small animal MRI scanner (Pharmascan, Bruker Biospin, Germany) equipped with 300 mT/m gradient coils. The HC is used in transmit-receive mode. The hybridized coil is made out of thin metallic rods and tubes, composed of non-ferrous materials: pure copper. Four resonators are attached inside a PMMA cylinder to form a right parallelepiped (see Fig. 3(a)). The length of the rods is roughly 375 mm.

FIG. 3. (a) Experimental MRI test bed with the gelatin phantom. (b) The loop coil. (c) Sections of a cylindrical gel with (first row) and without (second row) the HC. From left to right, the transverse, sagittal planes. (d) Four SNR profiles with (dashed line) and without (continuous line) HMC. The four axes are shown in Fig. 3(c).

FIG. 4. (a) Schematic view of the rat inside the MRI bed, the resonators, and the loop coil. (b) In vivo demonstration of the sensitivity and homogeneity improvement of a single loop coil. Axial (a) and (b), coronal (c) and (d), and sagittal (e) and (f) views of a rat head from the same spin-echo acquisition at 7 T without (a), (c), and (e) and with (b), (d), and (f) hybridized rods positioned around the rat. The following settings were used for this turboRARE sequence: 128 $\times$ 128 matrix size, 30 slices with voxel size of 275 $\mu$m $\times$ 235 $\mu$m $\times$ 500 $\mu$m, RARE factor = 8, TE/TR = 24 ms/3000 ms, NA = 8. The position of the various 3D regions of interest used to calculate mean signals is drawn as white dashed rectangles. The region of interest used to calculate the standard deviation of the noise is the red dashed square.

| ROI#  | In depth | Lateral | Longit. |
|-------|----------|---------|---------|
|       | 1        | 2       | 3       | 4       | 5      | 6       |
| SNR w/o HC | 33.2     | 4.2     | 30.8    | 14.4    | 31.6   | 27.7    |
| SNR w/ HC  | 31.7     | 10.2    | 30.2    | 18.7    | 31.0   | 27.1    |

TABLE I. Comparison of SNR with and without HC obtained on 6 ROI.
because the Larmor frequency of proton at 7 T equals 300 MHz. When the dielectric material under study is included between the four resonators, the resonance frequencies of the hybridized modes are slightly modified. To mitigate this effect, the length of the resonators can be adjusted. To that end, each resonator is made of a copper rod which can slip into a copper tube. This allows a fine-tuning of their resonance frequency. The side length of the square cross section is 35 mm. The MRI acquisitions were done using a 30 mm-diameter copper surface coil etched on a 1 mm-thick dielectric substrate. Even if the loop matching is much better when coupled to the 4 resonators (see Fig. 2), it is not yet sufficient for a MRI application. To reach efficiency close to 100%, we use a classical matching network made of tunable amagnetic capacitors to match perfectly for every acquisition. First, HC concept is validated with a homogeneous phantom. The lastest is a cylinder made of gelatin whose diameter and length are equal to 50 mm and 110 mm, respectively. The phantom is placed between the four half-wavelength resonators.

Fig. 3(b) presents MRI images of axial, sagittal, and coronal planes of the gel obtained with and without the hybridized resonators. The MRI sequence used to obtain these image is a fast T1-weighted 2D gradient echo sequence (isotropic resolution 0.5 mm, matrix size 100 × 100 × 150, TE = 3.8 ms, TR = 1321 ms, and flip angle 50°). The air bubbles inside the gel generate black dots in the image. We observe that the hybridized mode provides images that are much more homogeneous and extended than the one obtained with the loop alone. However, we observe that coupling with one resonator at the bottom is not as good as predicted by theory. This is because the gel partially shields the electrical field which is the main origin of coupling. Moreover, due to the complex environment of an MRI cavity, the coupling of the 4 resonators is harder to obtain. Note that we did not observe any susceptibility artifacts. They only occur in the very close vicinity of the 4 resonators. Fig. 3(d) shows SNR along several axes through the gel. The in vitro SNR profiles were calculated as follows: First, the signal was averaged over a 6 × 6 voxels at each position. Second, the signal to noise ratios were obtained by dividing this signal by the standard deviation of the signal measured in a 15 × 10 × 150 voxel box located in a region where no signal is expected (outside the gel). We observe a significant enhancement of the SNR in all the volume of HC.

Finally, the HC has been validated in vivo by imaging the head of a rat. The operation is performed under general anesthesia. The rat is placed on a bed support inside the PPMA cylinder. The head is located in the center of the HC (see Fig. 4(a)). Fig. 4 shows the 3 projections of a 3D T2 spin-echo acquisition. During the same experiment, this is compared with the metallic rods removed from the PPMA cylinder and the loop coil retuned and rematched (Fig. 4(b)).

We gratefully acknowledge the financial support of the French Government-funded technological research organization CEA/DAM. This work was supported by the French National Agency (ANR) with the Grant OPTTRANS (No. 2010 BLAN 0124 04), by the LABEX WIFI (Laboratory of Excellence ANR-10-LABX-24) within the French Program Investments for the Future under reference ANR-10-IDEX-0001-02 PSL* and by the Institut Carnot STAR funding through the project CMRI.
1Z.-P. Liang and P. C. Lauterbur, *Principles of Magnetic Resonance Imaging* (SPIE Optical Engineering Press, 2000).
2S. E. Solis, R. Wang, D. Tomasi, and A. O. Rodriguez, *Phys. Med. Biol.* 56, 3551 (2011).
3C. E. Hayes, W. A. Edelstein, J. F. Schenck, O. M. Mueller, and M. Eash, *J. Magn. Res.* 63, 622 (1985).
4P. B. Roemer, W. A. Edelstein, C. E. Hayes, S. P. Souza, and O. Mueller, *Magn. Res. Med.* 16, 192 (1990).
5U. Katscher and P. Bornert, *NMR Biomed.* 19, 393 (2006).
6A. Hoyos-Idrobo, P. Weiss, A. Massire, A. Amadon, and N. Boulant, *IEEE Trans. Med. Imaging* 33, 739 (2014).
7J. B. Pendry, D. Schurig, and D. R. Smith, *Science* 312, 1780 (2006).
8J. M. Algarin, M. J. Freire, M. A. Lopez, M. Lapine, P. M. Jakob, V. C. Behr, and R. Marques, *Appl. Phys. Lett.* 98, 014105 (2011).
9M. J. Freire, L. Jelinek, R. Marques, and M. Lapine, *J. Magn. Res.* 203, 81 (2010).
10M. S. Khennouche, F. Gadot, B. Belier, and A. de Lustrac, *Appl. Phys. A* 109, 1059 (2012).
11X. Radu, D. Garray, and C. Craeye, *Metamaterials* 3, 90 (2009).
12A. Christ, O. J. F. Martin, Y. Ekinci, N. A. Gippius, and S. G. Tikhodeev, *Nano Lett.* 8, 2171 (2008).
13S. J. Orfanidis, *Electromagnetic Waves and Antennas* (Rutgers University, 2014).
14R. Abdellah, N. Ourir, and J. de Rosny, *Phys. Rev. B* 83, 033101 (2011).
15J. T. Bushberg and J. M. Boone, *The Essential Physics of Medical Imaging* (Lippincott Williams & Wilkins, 2011).