Analysis of a Flexible Dual-Channel Octagonal CoIl System for UHF MRI

NICOLA CURRELI1, MATTEO BRUNO LODI2, (Member, IEEE), ANDREA MELIS2, CLAUDIO PUDDU3,4, SERGIO CASU2, ALESSANDRO FANTI2,5, (Member, IEEE), NIKOLA DJURIC6, (Member, IEEE), ALESSANDRA RETICO4, AND GIUSEPPE MAZZARELLA2,5, (Senior Member, IEEE)

1Functional Nanosystems, Istituto Italiano di Tecnologia, 16163 Genova, Italy
2Department of Electrical and Electronic Engineering, University of Cagliari, 09123 Cagliari, Italy
3POLARIS, Department of Infection, Immunity and Cardiovascular Disease, University of Sheffield, S10 2RX Sheffield, U.K.
4INFN-CA, Complesso Universitario di Monserrato, 09042 Cagliari, Italy
5Faculty of Technical Sciences, University of Novi Sad, 21000 Novi Sad, Serbia
6Department of Power, Electronic and Telecommunication Engineering, Faculty of Technical Sciences, University of Novi Sad, 21000 Novi Sad, Serbia

Corresponding author: Alessandro Fanti (alessandro.fanti@unicax.it)

This work was supported in part by the Advanced Techniques and Data Mining in Magnetic Resonance Imaging (MRI) [Advancing Magnetic Resonance Imaging and Data Analysis (2015–2017–nextMR)] through INFN (CSN5).

ABSTRACT Nowadays, MRI is focused on using ultra-high static magnetic fields (>7 T) to increase the signal-to-noise ratio. The use of high fields, on the other hand, requires novel technical solutions as well as more stringent design criteria for specific absorption rate levels, reducing radiative effect and coil resistance. In this paper, two flexible RF coils for 7 T human magnetic resonance, and 298 MHz ultra-high frequency operations were analyzed and characterized. Imaging of lower human limbs is regarded as a case study. The lumped element theory and subsequent numerical simulations were used to fine-tune the single-coil element and the dual-coil array design, respectively. Here, we demonstrate how the shape, size, configuration, and presence of the sample influence the coil performance. The penetration depth of the \( B_1 \)-field and the specific absorption rate values have been determined numerically using two numerical surface phantoms: saline and a multilayer human tissue. A preliminary study in the presence of a saline solution phantom has been carried out to develop and validate the dual-coil system. The frequency response of the dual-coil array was measured to assess its robustness when coupled to twelve human volunteers. We found that our design is robust to variations in the anatomical properties of the human thighs, and hence to coil bending. The presented approach can be useful for the implementation of flexible devices with high sensitivity levels and low specific absorption rate.

INDEX TERMS Coils, optimization, flexible printed circuits, magnetic resonance imaging, phantoms.

I. INTRODUCTION MAGNETIC resonance imaging (MRI) is a gold-standard technique for the assessment of several human diseases, and it is capable to offer a wide range of anatomical and functional information [1], [2]. MRI aims to provide clinically relevant images with high-contrast using the quantum principle of nuclear magnetic resonance [3]. Over the past 20 years, medical research has focused on the development of ultra-high-field (UHF – 7 T or higher) MRIs, thanks to their signal-to-noise ratio (SNR), which is almost double than that of a 4 T MRI [4]. They also allow acquiring high-resolution images in a short time [5]. One of the fundamental components of MRIs are the radiofrequency (RF) coils, which are responsible for transmitting and receiving signals from different areas of the human body and play a major role in determining the images’ SNR [2]. RF coils can be mainly categorized into volume and surface coils [6]. The volume coils surround the tissue under exam (TUE) and have good RF homogeneity, which extends over a large area [6]. On the other hand, the surface coils are specifically designed for localized body regions and are placed in proximity of the region of interest (RoI). For this reason, the geometry of a surface coil is a very important feature in medical applications.
Different circular, square, and decagonal surface coil configurations have been investigated, showing uniform field at low frequencies (<127.74 MHz) [7]. Hexagonal coils have been proposed [8], [9], [10]. However, these shapes exhibited lower performance with stronger applied static magnetic fields, as the coil current path is electrically large, due to the limited wavelength (~1 m @ 300 MHz, in air), increasing the coil radiation losses [2], [11]. Therefore, the octagonal configuration has been proposed [12], allowing the current to flow more easily in UHF, avoiding high current densities at the corners of the coil, [12], [13], minimizing the lateral skin effect [14], achieving larger magnetic sensitivity [15], and providing improved SNR [13]. Moreover, the octagonal geometry benefits from a low manufacturing cost due to the ease of implementation, e.g., through the screen-printing of the elements [16]. As a consequence, octagonal surface coils are receiving more and more attention, becoming the preferred choice over volume coils, despite the reduction of the field of view (FoV) [6]. Hence, increasing the FoV is one of the main goals to make the surface coils suitable for use in MRI: array coils allow covering a larger RoI than that obtained from a single surface coil, combining the local sensitivity of every single element [17], [18]. A variety of array coil designs have been proposed for UHF MRI, including microstrip elements [19], loop coils [20], dipole antennas [21], and dielectric resonant antennas [22]. Microstrip and multichannel transmit/receive (TX/RX) loop coils with independent phase and amplitude control, show a SNR on the RoI suitable for achieving specific absorption rate (SAR) levels within the IEC limits [23]. Similarly, dipole antenna arrays allow SAR levels of up to ~60% compared to other techniques [24]. On the other hand, dielectric resonant antenna arrays, which use both resonant and non-resonant high permittivity structures, exhibit a sensitivity ~25% higher than a loop gap resonator [25]. In the aforementioned designs, the elements of the array are decoupled using active strategies, which calls for metasurfaces design [20], [26] or electronics such as preamplifiers or decoupling circuits, i.e., complicated hardware solutions which implies a larger probe size, thus occupying a larger area [27], [28]. In addition, due to the materials and manufacturing techniques used, the probe costs inevitably increase [29]. For these reasons, using simple multiple loop coils as array elements is the most commonly used method for designing array coils [6]. Indeed, based on the investigated RoI, this design enables ease of implementation and allows choosing the optimal dimension and number of loops for the array. In this way, the array can be placed in close proximity to the RoI, allowing a higher SNR to be achieved even at larger depths (up to 60 mm), corresponding to better image quality [29], [30].

In this context, the realization of flexible coils is pivotal in view of the implementation of small size wearable RF arrays, offering several benefits [16], [31]. For example, a flexible receiver RF array can be tuned according to the subject (e.g., considering intra-subject anthropometric variability), providing optimal conformal fitting by increasing the filling factor [32], [33], as well as improved comfort for the patient during the MRI exams [32]. At the same time, flexible coils result in a high FoV and a significant increase in SNR levels since the design is less sensitive to both dielectric and conductive properties of the subject [34]. However, in UHF, the anatomical features of the subjects could lead to alterations in the electromagnetic response of the flexible probes. In fact, the generated $B_1$-field and the coil impedance become simultaneously a complex function of the size, shape, and material of the coil, as well as the interaction of the coil with the patient, affecting its robustness and its frequency response [35], [36]. Furthermore, at 7 T, the size of the biological TUE becomes comparable, or even greater, to the working wavelength, so the interaction between biological tissues and the electromagnetic field becomes much stronger [37]. Therefore, it is of great importance to avoid inhomogeneity in the spatial distribution of the transmitted RF fields which could lead to high induced currents and produce large phase-shifted contributions to the $B_1$-field, with consequent critical problems in patient safety [2], [38], [39], [40].

In this work, the in-depth analysis of a two-channel octagonal geometry flexible matrix coil for 7 T MRI is presented. In order to perform the analysis, we have implemented and characterized a proof-of-concept array configuration. Our analysis shows that this design allows for robust and effective performance on a wide range of subjects, ensuring a complete and passive geometric decoupling without the use of preamplifiers, resulting in a smaller probe size, and therefore, a closer fit to the sample during the exam.

II. COIL AND ARRAY DESIGN

A simple but effective design methodology is presented for the case study of a two-channel array for imaging human thighs and femoral regions. The early-stage analysis and design of the UHF single RF coil, for the 7 T MRI dual-coil
array, has been performed by using the LC circuit theory, treating the coils as a combination of inductive ($L_{coil}$) and capacitive ($C_{coil}$) lumped elements [41]. The lumped element circuit model is shown in Fig. 1a. As shown in Fig. 1b and Fig. 1c, two octagonal coils [24] with a thickness $a_{coil} = 0.6$ cm, side $L = 3.44$ cm and a radius $r_{coil} = 4.5$ cm have been considered. In order to match the coil to the 50 $\Omega$ impedance of the RF source (and of the input stage of the receiver), at the resonant frequency of $f_r = 298$ MHz, a tuning capacitance and a matching section are required [41]. $C_{coil}$ can be estimated from the resonance condition of the equivalent LC circuit [42]:

$$C_{coil} = \frac{1}{\omega_0^2 L_{coil}}$$ (1)

where $\omega_0 = 2\pi f_r$ is the resonant angular frequency, while $L_{coil}$ can be calculated according to the octagonal loop formula (Supplementary material) [12]. However, at UHF the coil electrical length is not small enough for the lumped-constant description to be accurate [2], hence Eq. 1 loses its accuracy, and also the current distribution could not be constant [2]. In addition, the search in the solution space is complicated by the possible influence of the bending of the flexible coil on the frequency response of the loaded array [2]. In order to get a constant current distribution along with the coil, we propose to divide the coil into $N = 4$ equal sections, each one shorter than $\lambda/10$ at $\omega_0$ [43], and a connecting tuning capacitor ($C_t$) between each consecutive pair of sections given by:

$$C_t = N \cdot C_{coil}$$ (2)

In this way, the current in each section is constant and, because of the capacitors, is the same in each section. The value of $C_t$ given by Eq. 2 can be assumed as a starting point for a pre-tuning procedure. In order to match the coil, a matching capacitance ($C_m$) is connected in parallel to the coil circuit as shown in Fig. 1. Unlike $C_t$, the value of matching capacitors is an unknown function of the coil input impedance, i.e., the load properties, which depends on the electromagnetic characteristic (namely, permittivity and conductivity) and the size and shape of the TUE [26]. Therefore, we performed tuning and matching optimization analysis using full-wave simulations via CST software to find suitable working conditions for MRI coil and array [44]. The values of $C_t$ and $C_m$ have been optimized by imposing $f_r$ and input matching conditions ($S_{11} < -15$ dB at $f_r$) by performing extensive numerical experiments given a set of coil radii and arms lengths [38]. Thus, it has been verified the strong mutual coupling between the coils and the TUE, carrying out the simulation in presence of the sample taken into account, i.e., the saline solution phantom and the human thigh as shown in Fig. 1d.

However, the proposed design approach could be adapted for other body areas. The two-channel array was used to image a larger RoI, with respect to a single loop coil. The array consists of two identical octagonal coils with a space between the two elements of $W = 0.75 \cdot r_{coil}$, which agrees with the theory of Roemer et al. [45], and to similar octagonal arrays operating at lower frequencies [46]. In this way, the strong influence on the input impedance is reduced [47], thus facilitating the implementation process and the robust performance against several types and shapes of loads. Indeed, by stabilizing the resistance, it is possible to retain almost constant quality factor ($Q_{tun}$) and sensitivity values. The curvature radius of the coil, defined in the geometrical reference frame shown in Fig. 1d, was not a free parameter in the CST (Microwave Studio, Simulia, 3DS, GE) simulations. In fact, for every given curvature, the values of $C_t$ and $C_m$ must be re-optimized, i.e., the described design procedure should be carried out for different body regions. However, given the pre-design with the lumped theory and the fast numerical tuning and matching, the proposed methodology could be easily scaled and re-adapted for different cases, thus being of interest to other researchers in the field of UHF MRI [2], [26].

**III. ARRAY CHARACTERIZATION AND ANALYSIS**

Since the resonant condition and the input match depend on the given sample materials, as well as on the curvature of the sample, the proposed dual-coil array was studied considering two different surface phantoms, represented in Fig. 2, as usually done in the literature. In the first model (Fig. 2a), a homogeneous cylinder (sample S1), with $r_{sample} = 6$ cm and $l_{sample} = 16.8$ cm, filled with a 0.1 M saline solution ($\varepsilon_r = 79 – j41.62$). This case study was considered the benchmark for the experimental validations of the realized dual-coil array for 7 T MRI. The second model (Fig. 2b) consists of a surface layered phantom that mimics the essential geometric and electromagnetic features of a human thigh, built as a multilayer cylinder (sample S2). A cylinder having the same dimensions as sample S1, with four layers (i.e., bone, muscle, fat, skin), models the human thigh (see Table 1) [48]. The performance of the designed dual-coil array for 7 T MRI together with the models was numerically analyzed with the purpose of testing the quality of field homogeneity and evaluating the SAR exposure of the proposed dual-coil array.

**A. IN SILICO TEST WITH BIOLOGICAL SURFACE PHANTOM**

From the periphery to the center, the phantom is composed of skin, fat, muscle, and bone [48], [49], [50], as shown in Fig. 2b. The physical parameters of the four human tissues used in the S2 samples are reported in Table 1 for the sake
TABLE 1. Physical properties of tissues at 298 MHz [48].

| Material       | Thickness (cm) | Conductivity (S m⁻¹) | Permittivity εr | Density (kg m⁻³) |
|----------------|---------------|----------------------|-----------------|------------------|
| Fat            | 2.2           | 0.0764               | 11.75           | 911              |
| Muscle         | 3             | 0.7700               | 58.23           | 1090             |
| Skin           | 0.1           | 0.6404               | 49.92           | 1190             |
| Cortical bone  | 0.7           | 0.0824               | 13.45           | 1908             |

TABLE 2. Anthropometric characteristics of the volunteers.

| Subj. ID | Diameter (cm) | Length (cm) | Weight (kg) | Height (m) |
|----------|---------------|-------------|-------------|------------|
| SO1      | 17.2          | 46          | 70          | 1.76       |
| SO2      | 16.8          | 48          | 80          | 1.75       |
| SO3      | 19.4          | 45          | 72          | 1.74       |
| SO4      | 19.0          | 47          | 72          | 1.67       |
| SO5      | 16.8          | 43          | 76          | 1.73       |
| SO6      | 16.2          | 45          | 70          | 1.69       |
| SO7      | 16.2          | 41          | 77          | 1.76       |
| SO8      | 19.7          | 42          | 85          | 1.80       |
| SO9      | 18.4          | 42          | 72          | 1.74       |
| SO10     | 14.9          | 42          | 74          | 1.70       |
| SO11     | 14.3          | 48          | 71          | 1.66       |
| SO12     | 16.4          | 41          | 70          | 1.78       |

Avg. ± std. 17.3 ± 1.8 44.1 ± 2.4 71.8 ± 5.4 1.73 ± 0.6

of completeness [48]. The designed dual-coil array for UHF MRI at 7 T was numerically tested with respect to the different geometry of the TUE and considering the influence of the curvature radius. The S2 sample was used to assess the coil sensitivity as a function of the phantom depth, according to [51], [52]. Furthermore, the SAR distribution and levels in the phantom were analyzed as follows [38]:

\[
\text{SAR} = \int_V \frac{\sigma |E|^2}{\rho} \, dV \tag{3}
\]

where \(\sigma\) is the tissue conductivity, in S m⁻¹, \(\rho\) is the tissue density in, kg m⁻³, and \(E\) is the induced electric field vector, in V m⁻¹, in the volume \(V\). The SAR is volume-averaged according to IEEE/IEC 62704-1 regulation [53].

B. REALIZATION AND EXPERIMENTAL ASSESSMENT

In order to confirm the results obtained through the simulations, a plastic phantom (sample S3) having the same dimensions of the sample S1 containing the saline solution was used. The designed dual-coil array for UHF MRI has been realized using conductive copper tape (metal thickness \(\varepsilon_t = 3.4\), \(\mu_t = 1\)), and each double coil is shown in Fig. 1c.

The experimental measurements of the performances of the realized dual-coil array have been performed using a Hewlett Packard, 8720C Vector Network Analyzer [54]. The instrument was calibrated using the Open-Short-Load (OSL) method. Furthermore, experimental measurements were carried out on 12 human volunteers. The complex-valued scattering parameters (S11) of the coil array was investigated on human thighs in order to demonstrate the robustness of the proposed design. The anthropometric characteristics of the subjects are reported in Table 2. This study was performed following the principles outlined in the Helsinki Declaration of 1975, as revised in 2000. Subjects were informed about the aim of the study, and they provided their informed consent.

IV. RESULTS

A. TEST ON SALINE SOLUTION

The dual-coil array for 7 T UHF MRI was firstly designed to resonate at 298 MHz and to be matched in presence of a saline solution phantom, having a radius of 6 cm (sample S1 in Fig. 2a). The guessing values for \(C_t\) are first derived from Eq. 2. Subsequently, the optimized \(C_t\) and \(C_m\) have been obtained through numerical simulations, as explained in Sect. II.B, and were found to be \(C_t = 5.4\) pF and \(C_m = 1.5\) pF, for each coil. These values are in accordance with devices having similar coil dimensions, and working frequency [20], [31], [55]. We found that the optimal spacing between the coils, required to passively decouple the array element, was found to be \(w = 6.75\) cm, i.e. \(0.75 \cdot r_{coil}\), for the geometry reported in Fig. 1b, confirming the results obtained by numerical simulations (Sect. II). The experimental measurements were then performed with the realized dual coil array on the sample S3, and the results were compared to the simulations. The simulated and experimental frequency responses of the \(|S_{11}|\) have been found in good agreement with simulation results (i.e., a maximum of 5 MHz shift and \(\sim 3\) dB difference), as shown in Fig. 3a. In order to understand and investigate how the TUE geometry affected the design and performance of the UHF array, the two-channel array was re-designed for three different values of curvature radii namely \(|BC| = 6, 8, 10\) cm (at a distance \(r_{sample} = |AD| = 6\) cm), considering a 0.1 M saline solution sample S1, as shown in Fig. 1c. The results found from the numerical design procedure are reported in Table 3. The corresponding \(S_{11}\) vs. frequency curves are shown in Fig. 3b. From the simulation results reported in Table 3 and in Fig. 3c it can be observed that as the radius of curvature increases (i.e., as the sensor becomes flatter), the effect on the frequency shift is marginal, suggesting a negligible effect of the capacitive coupling with the load. Indeed, the value of \(C_t\) is almost constant, with a maximum variation of about 3%. This result suggests that a correct selection of the coil capacitors ensures a robust design for different loads. The input resistance of the loop requires a slight variation of \(C_m\) by increasing the radius of curvature, given an almost constant real part of the input impedance (less than 1%), and a negligible change in the complex impedance (\(>0.50\) \(\Omega\) on average). The simulations about the field penetration in the saline phantom have been reported in the Supplementary materials. The provided numerical calculations proved the robust behavior of the dual coil array. For this reason, we selected a set of capacitance values capable of effectively match and tune the coil for any lBC. This solution lowers the complexity of implementation, thus not requiring adaptive tuning strategies. Having demonstrated that the validity of the tuning and matching procedure, the numerical investigation focused on demonstrating that the conformal shape of the flexible array could improve the RF
FIGURE 3. (a) Comparison between simulated and measured magnitude of $S_{11}$ of the dual-coil array in presence of the saline phantom S3 ($|BC| = 6$). (b) Simulated reflection coefficient at the input of both coils of the dual coil sensor for the different curvature radii and capacitance values, which are reported in Table 3.

TABLE 3. Simulations results.

| Material | $|BC| = 6$ cm | $|BC| = 8$ cm | $|BC| = 10$ cm |
|----------|---------------|---------------|---------------|
| $C_t$ (pF) | 5.4           | 5.6           | 5.6           |
| $C_m$ (pF) | 1.5           | 3.25          | 4.4           |
| $f_r$ (MHz) | 297.95        | 297.95        | 298.1         |
| $|S_{11}|$ (dB) | -31.21        | -41.466       | -36.44        |
| $Z_{11}$ ($\Omega$) | 50.43 - j0.59 | 51.28 - j0.74 | 50.23 - j0.22 |

B$_1$-field sensitivity in the more complex and realistic case of the human-like phantom (sample S2) shown in Fig. 2b.

B. ANALYSIS AND PRELIMINARY CHARACTERIZATION FOR HUMAN MRI

The behavior and performances of the dual-coil array for UHF MRI at 7 T were investigated numerically by simulating the sensor loaded by the S2 sample. The array response was tested in silico by varying the thickness of fat and muscle layers for different curvature radii, considering the bone and skin layers size to be constant. The thickness variation of fat and muscle is known to strongly affect the coupling, field homogeneity, and SAR patterns [56], [57]. In order to investigate how the subject variability affects the UHF MRI exam, the maximum B$_1^+$ efficiency and the SAR inside the S2 sample were considered and reported in Fig. 4.

Fig. 4a–4c show a considerable asymmetry of the B$_1^+$ efficiency since at 7 T the wavelength is in the same range as the sample size. In fact, shorter wavelengths (and higher conductive currents) lead to greater phase changes [55]. Moreover, from Fig. S2, it is also possible to notice a discontinuity in the B$_1^+$-field, near the surface between muscle and fat. Indeed, a larger field intensity is observable, due to the higher conductivity of the muscle. Deeper, in the muscle the field efficiency is reduced due to the higher conductivity [58]. By observing the relationship B$_1$/E (Fig. S8 in Supplementary Materials), and on the basis of the SAR models of Fig. 4d–4f, it can be noticed that the coil efficiency decreases in the skin-muscle layer since the E-field is higher in this position.

On the other hand, the coil’s optimal matching is demonstrated on the phantom depth ($\sim 35$ dB) and decreases as the bending increases. The volume-averaged SAR levels are shown in Fig. 4d–4f and the distribution is more inhomogeneous than the deposited power observed in the saline phantom case (Fig. S1, Supplementary material). For the human thigh phantom, given a curvature radius of $|BC| = 6$ cm, the computed maximum local SAR is lower than 4 W kg$^{-1}$, within the SAR limit of 10 W kg$^{-1}$ [53]. Given the high sensitivity and the safe SAR levels, the robustness of the proposed UHF array was evaluated. The array was tuned and matched as previously described. The B$_1^+$-fields profiles in Fig. 5 and Fig. S4 (Supplementary materials) are in agreement with ref. [6], and at $\sim 3$ cm and $\sim 4$ cm of depth in the tissue phantom S2, independently by the radius of curvature. The simulation and experimental results for the S-parameters are shown in Fig. 6. From the simulations results, for a fixed radius of curvature ($|BC| = 7$ cm, $C_t = 4.7$ pF, $C_m = 15$ pF), the variation of muscle and fat tissues thickness for 2 cm and 3 cm has a negligible effect on the resonance frequency (a maximum difference of 5 MHz and a minimum of $\sim 2.5$ MHz is found, as shown in Fig. S3 in Supplementary material), but affects the matching, as shown in Fig. 6. However, by comparing the in silico results with the experimental array $|S_{11}|$ in presence of the subjects, we found a variance of 2.5 MHz for $f_r$, at $\sim 15$ dB while $>20$ MHz from simulations. Therefore, both the array matching and tuning are not significantly affected by the anthropomorphic variability. The Q$_{fact}$ of the coil was quantified for the loaded and unloaded coil from the $S_{21}$ parameter vs. frequency. The relation between the loaded Q$_{fact}$ and the unloaded Q$_{fact}$ is an indicator of the coil sensitivity, and is reported as a measure to compare coil loops in their maximum efficiency to detect the MR signal [59]. For the unloaded coil a Q$_{fact}$ of 96 is obtained (Fig. S5, Supplementary material). Thus, even the loaded quality factor ranges from 20 to a maximum of $\sim 35$, for an average value of 24.577 (see Fig. S6, in Supplementary material). By comparing our findings to...
previous literature results (Supplementary material sec. 6), in [26] a four octagonal elements array, working at 298 MHz, showed a ratio of unloaded to loaded $Q_{\text{fact}}$ varying from 3.2590 (34/11) to 1.204 (65/54), whilst, a similar array working at 64 and 127 MHz, presented values in the range 6.7–11.4 when loaded (see Table S1). On the other hand, by considering the $B_1^+$-field efficiency, our flexible dual-coil system achieve a maximum value of $\sim 1 \mu T/\sqrt{W}$, which is generally higher than the efficiencies (0.15-0.4 $\mu T/\sqrt{W}$) for other UHF MRI systems (supplementary material, Table S2) [60], [61], [62], [63]. These results in the observed range of variation do not hamper the proposed design of the dual-coil array for UHF MRI at 7 T, in the case of human thighs confirming the robustness of the array.

V. CONCLUSION
In this work, we implement and fully characterize a flexible array of two octagonal coils for the imaging of the human thigh as a case study. The design methodology is based on passive decoupling of the elements and a numerical framework to achieve robust performances. Here, the proposed dual-coil UHF array for MRI was analyzed and validated both in silico and by using a saline solution. We carried out a numerical study on a multilayered human thigh phantom to evaluate the $B_1$-field distribution, homogeneity and SAR patterns. The influence of fat and muscle composition on the sensor response was also considered. The numerical simulations were validated by experimental measurements on
human subjects. We found that the designed system has a constant $B_1$-field sensitivity at $3 \text{ – } 4$ cm of depth as IBC increases. The sensitivity of our array is higher (>0.5) deeper in the TUE, i.e., 5 mm vs. 3 cm [16]. The SAR levels lower as the array flattens, are always under the threshold values [53]. Moreover, we found better results than the traditional loop array, made of circular surface coil working at 300 MHz, which demonstrated an asymmetric SAR distribution and a high peak value of 53.8 W kg$^{-1}$ [64]. These promising results together with an unloaded to loaded $Q$ which demonstrated an asymmetric SAR distribution and an array, made of circular surface coil working at 300 MHz, Moreover, we found better results than the traditional loop array, made of circular surface coil working at 300 MHz, which demonstrated an asymmetric SAR distribution and a high peak value of 53.8 W kg$^{-1}$ [64].

These promising results together with an unloaded to loaded $Q$ which demonstrated an asymmetric SAR distribution and an array, made of circular surface coil working at 300 MHz, Moreover, we found better results than the traditional loop array, made of circular surface coil working at 300 MHz, which demonstrated an asymmetric SAR distribution and a high peak value of 53.8 W kg$^{-1}$ [64].

ACKNOWLEDGMENT

(Nicola Curreli, Matteo Bruno Lodi, and Andrea Melis contributed equally to this work.)

REFERENCES

[1] B. Behzadnezhad, J. Andreae, S. A. Hurley, C. Filla, E. Mueller, B. D. Collic, N. Behdad, L. Populin, and A. B. Mcmillan, “Subject-specific, non-invasive helmet-restraint RF coil for awake, non-human primate MRI imaging,” IEEE J. Electromagn., RF Microw. Med. Biol., vol. 3, no. 3, pp. 177–183, Sep. 2019.

[2] R. H. Caverly, “RF aspects of high-field magnetic resonance imaging (HF-MRI): Recent advances,” IEEE J. Electromagn., RF Microw. Med. Biol., vol. 5, no. 2, pp. 111–119, Jun. 2019.

[3] G. Liney, MRI in Clinical Practice. London, U.K.: Springer, 2007.

[4] J. T. Vaughan, M. Garwood, C. M. Collins, W. Liu, L. DelBarre, G. Adrian, P. Andersen, H. Merkle, R. Goebel, M. B. Smith, and K. Ugurbil, “7T vs. 4T: RF power, homogeneity, and signal-to-noise comparison in head images,” Magn. Reson. Med., vol. 46, no. 1, pp. 24–30, Jul. 2001.

[5] C. Niu, Q. Wang, Y. Hu, Y. Wang, F. Tang, F. Liu, and S. Crozier, “Numerical design of high-efficiency whole-body gradient coils with a hybrid cylindrical-planar structure,” IEEE Trans. Biomed. Eng., vol. 66, no. 6, pp. 1628–1636, Jun. 2019.

[6] J. T. Vaughan and J. R. Griffiths, RF Coils for MRI. Hoboken, NJ: USA: Wiley, 2012.

[7] R. Rojas and A. O. Rodriguez, “Numerical study of the optimal geometry of MRI surface coils,” in Proc. 29th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc., Aug. 2007, pp. 3890–3893.

[8] J. Lazovic, D. S. Stojkovic, C. M. Collins, Q. X. Yang, J. T. Vaughan, and M. B. Smith, “Hexagonal zero mode TEM coil: A single-channel coil design for imaging multiple small animals,” Magn. Reson. Med., vol. 53, no. 5, pp. 1150–1157, 2005.

[9] M. K. Chauhby, M. Gupta, R. Harsh, and T. Bhuiya, “Multi-channel hexagonal surface coils for 1.5 T MRI scanner,” in Proc. Int. Conf. Commun. Syst. Netw. (ComNet), Jul. 2016, pp. 236–240.

[10] N. Ugle, T. K. Bhuiya, R. Dapkar, D. M. Shinde, and S. Bagchi, “Design and comparison of rectangular, square and hexagonal RF coils for 1.5 T MRI system,” in Proc. 9th Int. Conf. Comput., Commun. Netw. Technol. (ICCCNT), Jul. 2018, pp. 1–4.

[11] G. Barisano, F. Sephehrband, S. Ma, K. Jann, R. Cabeen, D. J. Wang, A. W. Toga, and M. Law, “Clinical 7 T MRI: Are we there yet? A review about magnetic resonance imaging at ultra-high field,” Brit. J. Radiol., vol. 92, no. 1094, Feb. 2019, Art. no. 20180492.
...
ANDREA MELIS received the bachelor’s degree in biomedical engineering from the University of Cagliari, Italy, in 2017. He worked as an Assistant Researcher with the University of Cagliari. His research interests include EM modeling and developing of RF coils at low and high frequencies, and design and realization of WSN systems for the monitoring of industrial processes, such as bread manufacturing.

CLAUDIO PUDDU received the bachelor’s degree in biomedical engineering from the University of Cagliari, Italy, in 2014, and the master’s degree from the University of Genova, Italy, in 2017. He is currently pursuing the Ph.D. degree with the University of Sheffield, U.K. He worked as an Assistant Researcher with the University of Cagliari, from 2013 to 2014. In 2017, he was a Visiting Student at the National Institute for Nuclear Physics (INFN), Pisa, Italy. His research interest includes EM modeling and developing of RF coils at low and high frequencies.

SERGIO CASU graduated in electronic engineering. He received the M.Sc. degree in telecommunication engineering from the University of Cagliari, in 2014, and the Ph.D. degree in electronic engineering and computer science, in 2018. His research interests include electromagnetism involving the design and characterization of resonant cavities as biological reactor, design of RF coil for ultra-high field MRI, design of antennas for UHF RFID, design of antennas for biomedical applications, and use of microwaves for farmland disinfection.

ALESSANDRO FANTI (Member, IEEE) received the Laurea degree in electronic engineering and the Ph.D. degree in electronic engineering and computer science from the University of Cagliari, Cagliari, Italy, in 2006 and 2012, respectively. He worked as a Postdoctoral Fellow with the Electromagnetic Group, University of Cagliari, from 2013 to 2016, where he is currently an Assistant Professor. Since 2016, he has been a Scientific Associate with the National Institute of Nuclear Physics (INFN) Cagliari Division. He has coauthored more than 120 scientific contributions published in international journals, conference proceedings, and book chapters. Since 2020, he has been acting as a Principal Investigator of the IAPC Project, funded with € 5 million by the Italian Ministry of Economic Development (MISE), within the AGRIFOOD PON I&C (2014–2020). His research interests include the use of numerical techniques for modes computation of guiding structures, optimization techniques, analysis and design of waveguide slot arrays, analysis and design of patch antennas, radio propagation in urban environment, modeling of bio-electromagnetic phenomena, and microwave exposure systems for biotechnology and bio- agriculture. He is a member of the IEEE Antennas and Propagation Society, the Italian Society of Electromagnetism, and the Interuniversity Center for the Interaction between Electromagnetic Fields and Biosystems. He is also an Associate Editor of the IEEE JOURNAL OF ELECTROMAGNETICS, RF AND MICROWAVES IN MEDICINE AND BIOLOGY.

NIKOLA DJURIC (Member, IEEE) was born in Novi Sad, Serbia, in 1973. He received the M.Sc. and Ph.D. degrees in telecommunication and signal processing from the Faculty of Technical Sciences (FTN-UNS), University of Novi Sad, in 2003 and 2009, respectively. He was a Teaching Assistant and an Assistant Professor, from 1997 to 2010 and from 2010 to 2015, respectively. Since 2015, he has been an Associate Professor at FTN-UNS, where he has also been a Full Professor, since 2000. He is the author of more than 70 journal and conference papers. His research interests include computational and applied electromagnetics, with experience in environmental electromagnetic field (EMF) measurement and monitoring for EMF compliance testing, estimation, and simulation of EMF whole-body exposure.

ALESSANDRA RETICO received the Ph.D. degree. She has been a Researcher at INFN, since 2005. She developed and validated on large samples automated systems for lung nodule detection on screening CT. She was the Coordinator of the Artificial Intelligence in Medicine Research Project, from 2019 to 2021. She was also the coordinator of several INFN-funded research projects in this field (SEVEN, 2011–2012, and TESLA, 2013–2014), focused in technological development for ultra-high-field MRI and the Pisa Unit Coordinator of the Advanced Techniques and Data Mining in MRI (nextMR) Project, from 2015 to 2017. She was a Research Unit Coordinator of the Project “Supporting an early autism spectrum disorder diagnosis through the support vector machine approach” funded by the Italian Ministry of Health and the ARIANNA Project “Interdisciplinary research platform for neuroimaging analysis in Autism Spectrum Disorders.” 2016–2018, funded by PAR-FAS 2007–2013, where an informatics platform has been developed to facilitate MRI data collection and thus neuroimaging-based research on Autism Spectrum Disorders (ASD). Her main research interests include medical imaging (RX, CT, MRI, and PET-MR), and image processing and analysis with standard and innovative techniques, including machine-learning and deep-learning based methods. She is a member of the “Train the Brain” consortium and of the Enhancing Neuro Imaging Genetics through Meta-Analysis (ENIGMA)–ASD working group.

GIUSEPPE MAZZARELLA (Senior Member, IEEE) received the degree (summa cum laude) in electronic engineering from the Università Federico II of Naples, in 1984, and the Ph.D. degree in electronic engineering and computer science, in 1989. In 1990, he became an Assistant Professor at the Dipartimento di Ingegneria Elettronica, Università Federico II of Naples. Since 1992, he has been with the Dipartimento di Ingegneria Elettrica ed Elettronica, Università di Cagliari, first as an Associate Professor and then, since 2000, as a Full Professor, teaching courses in electromagnetics, microwave antennas, and remote sensing. He is the author (or coauthor) of over 100 papers in international journals and a reviewer for many EM journals. His research interests include efficient design of large arrays of slots, power synthesis of array factor, with emphasis on inclusion of constraints, microwave holography techniques for the diagnosis of large reflector antennas, use of evolutionary programming for the solution of inverse problems, in particular problems of synthesis of antennas and periodic structures.

Open Access funding provided by 'Università degli Studi di Cagliari' within the CRUI CARE Agreement