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Stepped Impedance Resonators for High Field Magnetic Resonance Imaging

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Abstract

Multi-element volume radio-frequency (RF) coils are an integral aspect of the growing field of high field magnetic resonance imaging (MRI). In these systems, a popular volume coil of choice has become the transverse electromagnetic (TEM) multi-element transceiver coil consisting of microstrip resonators. In this paper, to further advance this design approach, a new microstrip resonator strategy in which the transmission line is segmented into alternating impedance sections referred to as stepped impedance resonators (SIRs) is investigated. Single element simulation results in free space and in a phantom at 7 tesla (298 MHz) demonstrate the rationale and feasibility of the SIR design strategy. Simulation and image results at 7 tesla in a phantom and human head illustrate the improvements in transmit magnetic field, as well as, RF efficiency (transmit magnetic field versus SAR) when two different SIR designs are incorporated in 8-

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element volume coil configurations and compared to a volume coil consisting of microstrip elements.

Keywords

MRI; RF coils; TEM; microstrip; stepped impedance resonators (SIR)

I. INTRODUCTION

High-field magnetic resonance imaging (MRI) has become a popular modality in research centers due to the benefits of potentially higher signal-to-noise ratio, contrast-to-noise ratio and spectral resolution for varying applications [1-4]. In these systems, RF coils are the magnetic field producing elements used to inductively excite and receive the nuclear magnetic resonance (NMR) signal in the subject.

This nuclear magnetic induction is most efficient at the field strength dependent Larmor frequency; coils must resonate at Larmor frequencies of ^1H protons (42.58 MHz/Tesla) to take advantage of the signal-to-noise benefits of 7T (298 MHz) or higher field strength MRI systems.

However, to achieve the aforementioned benefits of high field MRI, various RF challenges need to be overcome. As operational frequencies increase with an increase in field strength, the wavelengths decrease and become shorter than the dimensions of the object being imaged leading to constructive and destructive interferences of the fields in the sample [5, 6]. Increase in frequency also leads to an increase in radiative losses to the coil's surrounding field leading to RF inefficiencies [7, 8].

RF losses in the tissue loading are another area of concern at higher frequencies. Losses due to the magnetic field induced eddy current density and electrical field displacement current density both increase with increasing field strength [9-12].

To overcome some of these issues, techniques such as RF shimming [13-15] and Transmit SENSE [16] can be used to manipulate the RF waves in the sample while RF signal losses to the magnet bore environment can be minimized by coil design. Similarly, RF field propagation and losses in the tissue load of the coil can be controlled and the specific absorption rate (SAR) and consequential RF heating can be minimized by proper RF coil design [12, 17, 18].

A popular means for signal transmission and reception at these high field strengths has been multi-element transverse electromagnetic (TEM) transceiver coils [19]. The multi-element volume coil comprises of an array of transmission lines operated as independent coil elements in typically multiple-element transmit and receive configurations. In these arrays, a popular TEM transmission line element is the microstrip resonator [20-22]. The capacitively terminated microstrip resonator must generate a magnetic field over a large enough volume to illuminate the sample, but unlike an antenna application, the conservative near magnetic field is required rather than the far field generated by the transmission line. This resonator,

along the transmission line, produces a transmit magnetic field (B_1^+) above the center of the coil, and decreasing B_1^+ towards the capacitive terminations.

However, depending on the MR application, the RF field profile (B_1^+) may need to be uniform along the length of the element or maximized for a specific region above the element; for either case, electrical fields (subsequently SAR) need to be minimized. To spatially transform the excited RF field, new TEM strategies need to be investigated for near field profile control. In this investigation, we aim to optimize the TEM multi-element volume array through novel coil element designs. The traditional microstrip line elements are modified into multi-section alternating impedance sections referred to as stepped impedance resonators (SIR) to vary the impedance of the transmission line and to localize the current at desired locations along the length of the transmission line. These new elements are combined in multi-element arrays and the transmit magnetic field produced in phantom and experimental studies from the two designs are seen to have promising results when compared to the microstrip resonator. The feasibility of this approach is numerically simulated, and then experimentally validated by imaging a phantom and the human head at 7 tesla.

II. THEORY

A. TEM (Microstrip) Resonators

The fundamental building blocks of all RF coils are simple series and parallel RLC circuits that have respective open- and short-circuit transmission line analogs. With either open- or short-circuit terminations at both ends of the strip, a microstrip transmission line [23]–[25] is resonant when it's half wavelength long. Short-circuited microstrip ends lead to a current maximum at the strip ends while the voltage maximum is seen at the center of the microstrip line. Alternately, open-circuit ends lead to a current maximum at the center of the strip and a voltage maximum at the strip ends. This configuration is ideal when the microstrip is used as an MRI transmission element, as the strip current generates the magnetic field that needs to be maximized, while the strip voltage generates an electric field that must be minimized by the coil.

At 298 MHz (operating frequency of proton at 7 tesla), the microstrip resonators' half-wavelength is approximately 0.5 m in air, which is impractically long as an MRI element. To shorten the electric length of the resonator, the microstrip open terminations may be replaced by its lumped element equivalent, the capacitor.

B. Stepped Impedance Resonators (SIRs)

Stepped impedance resonators (SIRs) have been found to be advantageous when designing microwave filters [26-28]. The SIR resonator has advantages over the microstrip due to the reduction of circuit size and independent control of the characteristic impedance of the resonator during the design procedure. The SIR is composed of alternating narrow and wide signal line sections where the narrow sections of the line are much greater in impedance than the wide sections of the line, $Z_2 \gg Z_1$, as illustrated by a 3-section SIR in Fig. 1. When compared to a traditional microstrip line, this leads to the high impedance section having greater current density along the outer edges of the narrow conductor section of line than the

wide section line for the same amount of total current. In other words, the current per unit length is the same in the narrow and wide sections of the 7-section SIR, but the narrow section has a much higher current density. By alternating sections of high- and low-characteristic impedance lines, the current density along the length of the line can be modulated; SIRs may be designed by adjusting the number of sections and the impedances of the individual sections to control current density magnitudes at specific locations along the transmission line. This concept will be explained more thoroughly in the upcoming discussions.

In SIRs, the terminating capacitors (C_t) shorten the physical length and allow for tuning of the resonator to the desired frequency (Fig. 2). The impact of adding tuning capacitors to the 7-section and 3-section SIR can be seen in the S_{11} parameters in Fig. 3. Each capacitively terminated resonant structure was tuned and matched to 298 MHz, while the non-terminated structure was matched at the input for minimum reflections. It is seen that the microstrip SIR open circuit acts as a low Q resonator at lower frequencies. By adding capacitors, the first mode of resonance is lowered to 298 MHz to act as a high Q microstrip resonator and the line now acts as a resonator at the MRI frequency of interest (in our case 298 MHz).

To understand the magnetic field generated from these non-uniform transmission lines, the current density distribution on the elements need to be examined (Fig. 4). In RF head coils at 298 MHz, element line lengths (~12-15 cm) are small compared to the guide wavelength (~60 cm); hence, the quasi-static approach may be used. In the microstrip line, the current on the section is concentrated at the edges of the signal line and is a function of dielectric constant and the width of the strip (Fig. 4a). In the SIR circuits, the quasi-static formulation suggests continuity of current at the junction of the wide and narrow line sections (neglecting the small discontinuity parasitic effects). Thus, the time average current density is greater in the narrow sections than in the wide sections close to the strip and therefore, the magnetic field is higher near the narrow sections when compared to the wide sections.

The free space electromagnetic field profiles for a single microstrip element, 3-section SIR and the 7-section SIR microstrip element for the same input power illustrates this effect (Ansoft's HFSS simulations) (Fig. 4a-f). In this example, the wide and narrow strip widths, as well as, the total coil length are same dimension for all three coil types. All the elements are tuned to 298 MHz with tuning capacitors. The current density plots of the SIR elements versus the straight microstrip illustrate the impact of varying the impedances along the line (Fig. 4 a, c, e). The microstrip element has a for-shortened sinusoidal current distribution with slight maximum amplitude at the center of the edge and minimum at the ends (Fig. 4a). In the 7-section SIR, the narrow sections have an uneven distribution of current density; the 3 narrow sections having significantly greater current density at their edges than the 4 wide sections (Fig. 4c). In the 3-section SIR design, there is greater current density in the narrow middle section than the wide sections (Fig. 4e). Fig. 4b, d, and f illustrate the equivalent H-field (magnetic field) in the central plane along the length of the coil. The sinusoidal current distribution in Fig. 4a equates to the sinusoidal field distribution in Fig. 4b with maximum H at the center. The strong current profiles in the 3 narrow sections of the 7-section SIR Fig. 4c lead to the combined 3 peak behavior (ripple appearance) in the H-field as seen in Fig. 4d.

The 3-section element indicates that the strong current density in the middle narrow section leads to stronger H-field in the center of the coil than the microstrip (Fig. 4f).

For the same input power, the modulation of current densities along the length of the transmission line coupled with the change in the H-field along the length of the coil in Figure 4 indicates that the greater amount of H-field can be directed towards the sample with the 3-section SIR and greater amount of H is available along the length of the coil for a 7-section SIR than the traditional microstrip element.

However, these elements cannot be evaluated just in a free-space medium. With a dielectric of air or Teflon ($\epsilon_r = 2.08$) along the transmission line, these coils operate in a quasi-static regime; however, this is not the case when the coils are coupled to a lossy sample. In the sample, at 298 MHz, the resonant wavelength is greatly reduced due to the high dielectric constant. The average human head at this frequency has a relative permittivity of ~ 60 reducing the free space wavelength ($\sim 100\text{cm}$) by a factor between 7 and 8. This leads to the resonant wavelength being close to the diameter of the head. At this frequency, the current along the length of the coils can be considered to be in the quasi-static regime because the wavelength is much longer than the coil structure, but the fields that are produced in the head must not be considered quasi-static. Instead of phase being constant in the head, phase-dependent constructive and destructive interferences can dominate a coil's performance. In the upcoming sections, an electromagnetic solver will illustrate the impact of these designs in the phantom and in the head.

III. SIMULATION AND EXPERIMENTAL METHODS

A. Electromagnetic Simulation

Single element plots for conduction currents in Fig. 4a-f were simulated with HFSS version 11. Agilent's Advanced Design Systems (ADS) was used to produce the S-parameters on a circuit level for the SIR and microstrip designs (Fig. 2 and 3). Numerical Maxwell solutions for phantom investigations of the 8-element transceiver arrays were calculated for a sucrose/saline phantom ($\epsilon_r = 58.1$, $\sigma = 0.539\text{ S/m}$) using XFDTD version 6.5 (Remcom Inc., State College, PA). Each element structure was simulated individually. In simulation, the microstrip, 7-section SIR, and 3-section SIR required tuning capacitors of 6.3 pF, 4.7 pF, and 3.3 pF, respectively, for resonance at 298 MHz. The magnetic B_1 field from the simulations was decomposed into its circularly polarized rotating counterclockwise component (B_1^+) with respect to the direction of the main magnetic field. The B_1^+ is denoted as the transmit component of the magnetic field can be extracted from the simulations by Eq. 1.

$$B_1^+ = \frac{B_x + iB_y}{2} \quad (1)$$

The B_1^+ shimming method [15] is used in order to maximize the constructive interferences at the ROI between complex B-fields of multiple transmit elements, thus maximizing the available B_1 magnitude. Once the fields of a single element were calculated, net fields produced by an eight element coil on the phantom was calculated in post-processing by

Matlab version 7.5 (MathWorks, Natick, MA). Because the phantom was rotationally spherical, the fields generated by a coil at any other angular position were equivalent to rotating the single channel field maps around the central axis. By adding $n \cdot 45^\circ$ of phase ($n = 0, 1, 2, \dots, 7$) to each of the synthetic coils and using the principle of superposition, the net fields of an eight channel coil phased for constructive interference in the center of the phantom was calculated. For head simulations, each element was simulated individually and combined in post-processing with B_1^+ shimming phases [29, 30] to produce a circularly polarized B_1^+ field in the center of the brain. All the coils were normalized to 1 Watt of input power for comparison purposes.

To estimate SAR in the head, full-Maxwell calculations of absorbed power were performed for a human head model in three different coils. The head model was created using the anatomical data from the National Library of Medicine Visible Male Project. A head model resolution of 2.0 mm in the left-right and anterior-posterior directions and 2.5 mm in the inferior-superior directions was used for calculations. The SAR values in the simulation tool (XFDTD) are calculated by Eq. 2.

$$SAR = \frac{\sigma_x}{2\rho_x} |E_x|^2 + \frac{\sigma_y}{2\rho_y} |E_y|^2 + \frac{\sigma_z}{2\rho_z} |E_z|^2 \quad (2)$$

B. RF Coil Design and Construction

Two TEM transceiver volume arrays consisting of 3-section and 7-section SIR elements and one TEM transceiver array consisting of uniform microstrip elements were built (Fig. 5). Each element was comprised of a low loss Teflon substrate ($\epsilon_r = 2.08$) with height and length of 1.2 cm and 14.0 cm, respectively. The elements were attached to a cylindrical plexi-glass shell 25.4 cm in diameter and 14.0 cm in length. The conductor widths were alternated with 1.2 cm wide copper tape and 16 gauge copper wire (0.13 cm diameter) for the 3-section elements (2 tape and 1 wire section) and 7-section elements (4 tape and 3 wire sections), while the standard straight microstrip element had only 1.2 cm copper foil (14 cm in length). Once in a multi-element volume configuration, each coil type was individually tuned to 298 MHz (7 T) and matched to a 50- Ω , coaxial cable for loaded conditions (sucrose/saline phantom and human head). Bench tests allowed for more than -15 dB of isolation between nearest-neighbor elements when loaded with a phantom and the human head for all three volume coils, hence, decoupling capacitors between nearest neighbors were not incorporated in the designs.

An 8-liter cylindrical phantom was designed to represent the electrical properties of the head (Fig. 6). The phantom was chosen to be roughly the same diameter as the head and long enough to represent the neck. The phantom was designed with the recipe given in [31] and consisted of sucrose (47.6%), distilled water (51.1%), and salt (1.3%). The 8-liter bottle was filled with 5071.7 g of sucrose, 4717.1 g of water, and 129.1 g of salt to attain the percent volume specifications of the recipe.

The equipment used to measure and validate the electrical properties of the solution was an Agilent E4991A RF Impedance Analyzer with an Agilent 85070D Dielectric Probe Kit. The

resulting dielectric permittivity (ϵ_r) and conductivity (σ) of the solution was 58.11 and 0.54 (S/m) respectively. Comparing the results to published works by Gabriel [32] at 298 MHz, it can be seen that the values obtained in this phantom are comparable to the values obtained at the same frequency for an average brain by Gabriel ($\epsilon_r=52.01$ and $\sigma = 0.55$ S/m) [33].

C. MRI Experiments

Experiments were performed in a 7 T ($\omega_0 = 298.0$ MHz) 90 cm bore magnet with Siemens console, head gradients, and 8 independent RF receivers (Magnex Scientific, UK) interfaced to a Siemens console (Siemens HealthCare, Germany). In vivo data were acquired on volunteers with signed consent under an Institutional Review Board approved protocol. Eight 1-kW RF amplifiers (CPC, Brentwood, NY) were utilized for transmission. Each of the eight RF outputs was wired to custom-built Transmit/Receive (T/R) switches allowing for both transmit and receive operations.

Mapping of the transmit B_1 magnitude was accomplished with a 2D double angle mapping method approach (60 and 120 degrees) for both the phantom and human experiments [34]. The sequence consisted of a Gradient Echo acquisition with the following parameters: Phantom (TR=1500 ms, TE=4 ms) and Human (TR=6000 ms, TE=4 ms), and slice thickness of 5 mm. For final image reconstruction, images from each receive element were simply combined as the root of the sum of squares. B_1^+ mapping and B_1^+ shimming calculations were performed in Matlab (MathWorks, Natick, MA). All acquisitions complied with FDA guidelines for specific absorption rates (SAR) and gradient duty cycle.

In phantom experiments, incremental geometric phase sets were used for B_1^+ mapping; Neighboring elements drive phases were incremented by 45 degrees. In head experiments, B_1^+ phase shimming was performed by determining a set of transmit phases that will maximize the B_1^+ phase coherence at the center of the coil, graphically prescribed in a custom GUI written in Matlab. These phases are calculated by averaging the relative transmit phase over the chosen ROI (center of the brain); B_1^+ shimming is performed by subtracting these phases from each transmit elements initial drive phase through the means of computer controlled phase shifters for subsequent acquisitions when transmitting with all elements simultaneously.

IV. RESULTS AND DISCUSSION

A. Phantom Results-Experimental and Simulation

Figures 7 and 8 illustrate the simulated FDTD and experimental B_1^+ results produced for the sucrose/saline phantom by the 8-element volume coils consisting of microstrip, 7-section SIR, and 3-section SIR elements at 7 tesla. Fig. 7 displays the central transaxial slice results while Fig. 8 illustrates the central sagittal slices of the sucrose/saline phantom. The sagittal slices acquired were centered over the microstrip, 3-section and 7-section SIR for both experimental and simulated results. The peak B_1^+ values are termed as B_1^+ values at the center pixel of the phantom. When normalized to the same input power (1 Watt), the B_1^+ XFDTD center slice axial simulations with the phantom show the 8-element, 3-section SIR ($0.310 \mu\text{T/W}^{0.5}$) coil performing at 45% greater in the center of the coil (Fig. 7c) than the TEM microstrip resonator ($0.214 \mu\text{T/W}^{0.5}$) (Fig. 7a), while 7-section SIR ($0.296 \mu\text{T/W}^{0.5}$)

has a B_1^+ of which has 38% greater B_1^+ in the center of the coil (Fig. 7b). Experimentally, the 3-section SIR coil performs at $(0.275 \mu\text{T}/\text{W}^{0.5})$ in the center of the phantom (Fig. 7f) which is 38% greater in the center of the coil compared to the microstrip $(0.199 \mu\text{T}/\text{W}^{0.5})$ element (Fig. 7d) while the 7-section SIR coil $(0.240 \mu\text{T}/\text{W}^{0.5})$ performs 24% greater in transaxial plane (Fig. 7e).

In the sagittal central slice investigation (Fig. 8a-f), experimental and simulated B_1^+ maps illustrate the pronounced circular excitation with higher central peak B_1^+ values for the 3-section SIR volume coil (Fig. 8c, f) while, the 7-section SIR illustrates the larger extent of usable B_1^+ intensity in the center plane of the phantom (Fig. 8b, e) when compared to the TEM microstrip and 3-section SIR volume coils (Fig. 8a, d).

B. Human Head Results-Experimental and Simulation

Figure 9 displays the experimental 7 T central transaxial and sagittal slice B_1^+ maps attained for all three volume coils, while, Figure 10 illustrates the simulated B_1^+ and SAR results from the same coils. The head in the coil was centered such that the nose of the patient was placed between coil elements in all experiments. The central sagittal slices were therefore acquired with the center of the images located in-between elements in all experiments. Table 1 summarizes the results from Fig. 9 and 10. The red dashed line in Fig. 9a indicates the regions in the central transaxial slice where B_1^+ values were averaged for all studies. Experimental results were normalized to 1 Watt of input power for all studies.

In the transaxial central slice in the human head (Fig. 9a-c), the 8-element volume coil consisting of 3-section element coil produced an averaged peak B_1^+ value of $0.397 \mu\text{T}/\text{W}^{0.5}$ in the center of the brain, while the 7-section element coil volume coil performed at $0.372 \mu\text{T}/\text{W}^{0.5}$ in the same region which is 20% and 12%, respectively, greater in the same slice than the TEM microstrip element $(0.332 \mu\text{T}/\text{W}^{0.5})$ (Fig. 9a-f). The B_1^+ XFDTD simulations of the head (Fig. 10a-c) illustrate the same constructive addition of fields as the experimental investigations and show the 8-element, 3-section SIR coil $(0.514 \mu\text{T}/\text{W}^{0.5})$ having 25% greater B_1^+ value in the center of the brain while the 7-section SIR $(0.490 \mu\text{T}/\text{W}^{0.5})$ has 19% greater B_1^+ in the center of the brain when compared to the microstrip $(0.409 \mu\text{T}/\text{W}^{0.5})$ indicating strong correlation between experimental and theoretical results. A summary of these results can be seen in Table 1. Note: Remcom's XFDTD software does not account for cable losses, impedance mismatches, hence the slight differences in B_1^+ values.

However, an increase in B_1^+ also leads to an increase in electrical fields and subsequent SAR values in the sample. The simulated SAR results (Fig. 10 d-f) comparing all the coils for 1 W input power (Table 1) indicates that the 8-element volume coil consisting of 7-section SIR elements (1.791 W/kg) has greater peak SAR than the 3-section SIR (1.573 W/kg) ; both coil types have a greater peak SAR values than the microstrip structure (1.466 W/kg) (Fig. 10 d-f). It is difficult to compare the three different volume coils on just B_1^+ and peak SAR values; these two benchmarks need to be combined to get a true indication of coil performance. This can be accomplished by investigating the RF efficiencies of the coils. This term is defined by the authors as the magnetic energy versus localized electric field energy per unit power in the sample or simply B_1^+ versus peak SAR.

In simulation studies (Fig. 10a-f), B_1^+ to peak SAR ratio ($\mu\text{T}/\text{W}^{0.5}/\text{W/kg}$) of the 3-section (0.3268) has a greater ratio than the 7-section SIR (0.2736) and microstrip TEM element (0.2790), which have similar ratios. In experimental B_1^+ versus peak simulated SAR values, once again, the 3-section SIR (0.253) has a greater ratio than the microstrip TEM structure (0.226). The 7-section SIR performs the worst with a ratio of 0.208. These results are summarized in Table 1.

The greater peak B_1^+ in the center of the phantom is also seen in the head experiments. However, the large extent of usable B_1^+ seen in sagittal plane for the phantom studies for the 7-section SIR is not seen in the same coil during head experiments. The non-uniform shape and electrical properties of the head lead to constructive interference patterns very similar to the 3-section SIR and the TEM microstrip structures. Also, the 3-section element coil has a greater amount of its field directed towards the middle of the head (Fig. 9f) which is illustrated by the lack of signal at the brain stem when compared to the microstrip and the 7-section SIR coils.

V. CONCLUSIONS

In this paper, TEM transceiver volume phased arrays consisting of stepped impedance resonators are designed and investigated for human head imaging for a 7 tesla MRI system. The modulations of current along the transmission line in these new structures lead to the control of transmit magnetic field profile on the signal line and in the sample. Two different stepped impedance resonators showed improvements in transmit peak B_1^+ in phantom and in the head during simulation and experimental investigations when compared to the volume phased array coil consisting of microstrip resonators. FDTD calculations of transmit magnetic field versus peak SAR in the head also showed benefits of using stepped impedance resonators when compared to the microstrip elements.

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Fig.1.
3-section SIR signal line illustration of the low impedance (Z_1) and high impedance (Z_2) sections.

Fig.2.

7-section and 3-section SIR circuits with and without terminating capacitors.

Fig.3.

ADS simulations of 7-section and 3-section SIRs with and without terminating capacitors illustrating the reduction of the circuit and resonance at 298 MHz (7 tesla).

Fig.4. Current density (J/m) profiles on signal line a), c), e) and H-field (A/m) along the center plane of the coil (sagittal plane) b), d), f), respectively, of capacitively terminated microstrip resonator, 7-section SIR and 3-section SIR (HFSS simulations)

Fig.5.

a) 8-element microstrip TEM coil b) 8-element 7-section SIR and c) 8-element 3-section SIR volume coil.

Fig.6.
Sucrose/saline phantom placed in 8-element 7 T coil phased array

Fig.7.

Central slice transaxial phantom B_1^+ maps ($\mu\text{Tesla}/W^{0.5}$) XFDTD simulation (a-c) and 7 T experimental (d-f) results for 8 element for a, d) microstrip b, e) 7-section SIR c, f) 3-section SIR.

Fig.8.

Central slice sagittal phantom B_1^+ maps ($\mu\text{Tesla}/\text{W}^{0.5}$) XFdtd simulation (a-c) and 7 T experimental (d-f) results for 8 element for a, d) microstrip b, e) 7-section SIR c, f) 3-section SIR.

Fig.9.

Central transaxial (a-c) and sagittal (d-f) experimental B_1^+ maps ($\mu\text{Tesla}/W^{0.5}$) of human head for 8-element volume coil consisting of a, d) microstrip, b, e) 7-section SIR c, f) 3-section SIR (the dashed circular ring in a) indicates the regions in the image where the B_1^+ values were averaged). The top edge of the 14 cm coil was aligned with the top of the patients' skull.

Fig.10.

Central sagittal B_1^+ ($\mu\text{T}/\text{W}^{0.5}$) (a-c) and SAR (W/kg) XFDTD simulations (d-f) results for 8 element for 8-element volume coil consisting of a, d) microstrip b, e) 7-section SIR c, f) 3-section SIR.

Table 1

B_1^+ and SAR Results for 8 element volume coils with human head - Simulation and Experimental

8 element coil	Peak B_1^+ (Simulated) ^a	Peak B_1^+ (Experimental) ^a	Peak. SAR (Simulated) ^b	Peak B_1^+ (simulated) /Peak SAR (Simulated) ^c	Peak B_1^+ (experimental) /Peak SAR (simulation) ^c
microstrip	0.409	0.332	1.466	0.2790	0.226
7-section SIR	0.490	0.372	1.791	0.2736	0.208
3-section SIR	0.514	0.397	1.573	0.3268	0.253

^aPeak B_1^+ values in $\mu T/(W^{0.5})$

^bPeak SAR values in W/kg

^cRatio of peak B_1^+ versus SAR in $\mu T/(W^{0.5})/W/kg$