A New Combination of Radio-Frequency Coil Configurations Using High-Permittivity Materials and Inductively Coupled Structures for Ultrahigh-Field Magnetic Resonance Imaging

Jeung-Hoon Seo 1, Young-Seung Jo 1,2, Chang-Hyun Oh 2,* and Jun-Young Chung 3,4,*

1 Neuroscience Research Institute, Gachon University, Incheon 21988, Republic of Korea
2 Department of Electronics and Information Engineering, Korea University, Sejong 30019, Republic of Korea
3 Department of Neuroscience, College of Medicine, Gachon University, Incheon 21565, Republic of Korea
* Correspondence: ohch@gachon.ac.kr (C.-H.O.); jychung@gachon.ac.kr (J.-Y.C.); Tel.: +82-32-822-5361 (J.-Y.C.)

Abstract: In ultrahigh-field (UHF) magnetic resonance imaging (MRI) system, the RF power required to excite the nuclei of the target object increases. As the strength of the main magnetic field (B₀ field) increases, the improvement of the RF transmit field (B₁⁺ field) efficiency and receive field (B₁⁻ field) sensitivity of radio-frequency (RF) coils is essential to reduce their specific absorption rate and power deposition in UHF MRI. To address these problems, we previously proposed a method to simultaneously improve the B₁⁺ field efficiency and B₁⁻ field sensitivity of 16-leg bandpass birdcage RF coils (BP-BC RF coils) by combining a multichannel wireless RF element (MCWE) and segmented cylindrical high-permittivity material (scHPM) comprising 16 elements in 7.0 T MRI. In this work, we further improved the performance of transmit/receive RF coils. A new combination of RF coil with wireless element and HPM was proposed by comparing the BP-BC RF coil with the MCWE and the scHPM proposed in the previous study and the multichannel RF coils with a birdcage RF coil-type wireless element (BCWE) and the scHPM proposed in this study. The proposed 16-ch RF coils with the BCWE and scHPM provided excellent B₁⁺ field efficiency and B₁⁻ field sensitivity improvement.

Keywords: inductively coupled wireless structure; high-permittivity material; finite-difference time domain method; multichannel RF coil; birdcage RF coil; 7.0 T MRI

1. Introduction

Advances in noninvasive 3-dimensional (3D) medical imaging such as ultrasound, computed tomography, and magnetic resonance imaging (MRI) have contributed to the early diagnosis of neurodegenerative diseases with enhanced accuracy. Among them, MRI provides a high signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR), as well as exquisite spatial resolution. In addition, MRI has been highly utilized in clinical practice, as it can provide both anatomical and functional images.

However, over the past 20 years, the main magnetic field (B₀ field) strength of MRI has remained limited to 3.0 T [1–4], and the clinical use of ultrahigh-field (UHF) MRI (B₀ field: 7.0 T or higher) has been delayed despite its advantages, such as high SNR, CNR, and spatial resolution, owing to the increased B₀ field strength.

The recent approval of 7.0 T MRI scanners by the United States Food and Drug Administration (FDA) can greatly contribute to the diagnosis of neurodegenerative diseases, such as Alzheimer’s disease [5–10]. In particular, 7.0 T MRI can help detect and distinguish subtle and potentially treatable lesions that cannot be detected or optimally evaluated at 3.0 T [11–17]. Nevertheless, 7.0 T MRI could not be widely used in clinical applications due to radio-frequency (RF) safety issues because RF coils require relatively high RF power higher than 3.0 T [18–25]. For instance, the use of the 7.0 T MRI system approved by the U.S. FDA is
subject to clinical permission to acquire only head and extremities (arms and legs) images. The 7.0 T MRI system requires a large amount of RF energy for whole-body imaging, causing tissue heating upon RF exposure [26–32] and increasing the specific absorption rate (SAR) depending on the electric field (E field) concentration [33–42].

The improvement of RF coil performance plays an essential role in overcoming the RF safety issues related to UHF MRI, which requires the acquisition of images using less RF power to secure RF safety as the B\textsubscript{0} field strength increases. For this reason, RF field (B\textsubscript{1} field) sensitivity and uniformity of the RF coil have become important design elements [43–47]. In the MRI system, the RF power required to excite the nuclei of the target object increases with the B\textsubscript{0} field strength. As the B\textsubscript{0} field intensity increases, the B\textsubscript{1} field sensitivity of the RF coil used in UHF MRI becomes particularly important because greater RF power is required to excite the proton (\textsuperscript{1}H) nuclei of the human body using RF pulses [48–58].

To improve B\textsubscript{1} field efficiency, the simultaneous improvement of the RF transmit field (B\textsubscript{1}+ field) efficiency and receive field (B\textsubscript{1}− field) sensitivity of the RF coil is required [59–64]. Previous studies have been conducted to improve the B\textsubscript{1}+ field efficiency and the B\textsubscript{1}− field sensitivity separately; however, UHF MRI requires the improvement of both the B\textsubscript{1}+ field efficiency [47,65–69] and the B\textsubscript{1}− field sensitivity [47,70–76] simultaneously.

Therefore, a method for simultaneously improving the B\textsubscript{1}+ field efficiency and the B\textsubscript{1}− field sensitivity was previously proposed to obtain a UHF MR image with a minimum SAR. The proposed RF coil configuration was based on the combination of high-permittivity materials and inductively coupled elements to improve the B\textsubscript{1}+ field efficiency and the B\textsubscript{1}− field sensitivity, determined by electromagnetic field (EM field) simulations [47].

To explain the RF coil configurations proposed in our previous work more specifically, the EM field simulation was performed using a bandpass-type birdcage (BP-BC) RF coil for RF transmission and reception because BP-BC RF coils provide the most uniform and the highest B\textsubscript{1} field efficiency in UHF MRI [77–80]. Additional structures were proposed using segmented cylindrical high-permittivity material (scHPM) to improve B\textsubscript{1}+ field efficiency and a multichannel wireless element (MCWE) to improve B\textsubscript{1}− field sensitivity. The suitability of their use in 7.0 T MRI has been confirmed in a previous study, but further investigation is required to obtain their combination to further improve the B\textsubscript{1}+ field efficiency and the B\textsubscript{1}− field sensitivity.

First, the disadvantage of the BP-BC RF coil combinations with scHPM and MCWE is that unwanted frequencies can be applied to scHPM by the extremely narrow mode space of the BP-BC RF coil in UHF MRI. The B\textsubscript{1}+ field generated around each leg of BC RF coil is drawn according to Maxwell’s right-hand rule. Each leg of the BC RF coil is driven by a sinusoidal current, but the peak current of each successive leg is delayed by 360°/16 (number of legs) = 22.5°. This is a homogeneous mode of the BC RF coil. In the resonance spectrum (in S-parameters) of the BC RF coil, the BC RF coil has various modes equal to half the number of legs of the BC RF coil, except for the end-ring modes. However, the modes of these BC RF coils have an extremely narrow mode space in the UHF MRI. For this reason, a decrease in B\textsubscript{1}+ field efficiency was expected; thus, the BP-BC RF coil used as the transmit/receive (Tx/Rx) RF coil was replaced with the 16 channel (16-ch) RF coil. We also improved the B\textsubscript{1}− field sensitivity of the 16-element MCWE using large-volume BC RF coils as the wireless element (WE) owing to an increase in the inductively coupled area. The dimensions of the MCWE and scHPM were defined as the size between the legs of the BP-BC RF coil. The B\textsubscript{1} field distribution of the BP-BC RF coil is generated in the vertical direction of the closed loop between the legs. If the MCWE and scHPM dimensions exceed the size between the legs of the BP-BC RF coil, the RF wave may be distorted horizontally, and signal sensitivity may be reduced, so the dimensions and number of MCWE and scHPM were set to a size that can minimize RF wave distortion.

Therefore, in this study, we propose a new combination of RF coil configuration that provides superior B\textsubscript{1}+ field efficiency and B\textsubscript{1}− field sensitivity at 7.0 T MRI compared to the previously proposed combinations of the scHPM and the MCWE. To this end, EM field simulations were performed by alternating the role of the BP-BC RF coil used as a Tx/Rx
RF coil and the MCWE used as a WE to improve $B_1^-$ field sensitivity, except for scHPM (which has already been verified to improve $B_1^+$ field efficiency).

Moreover, we compared the 16-ch RF coil with or without the BCWE (w/wo-BCWE) and the BP-BC RF coil with or without the MCWE (w/wo-MCWE) using EM field simulations. Thus, the optimal configuration of the BP-BC RF coil with MCWE and scHPM combinations (BP-BC RF coil + scHPM – w/wo-MCWE) and the 16-ch RF coil with BCWE and scHPM combinations (16-ch RF coil + scHPM – w/wo-BCWE) was identified with enhanced $B_1^+$ field efficiency and $B_1^-$ field sensitivity. Through the alternating use of the Tx/Rx RF coil and the WE, it was possible to determine which cases provided further improved $B_1^+$ field efficiency and $B_1^-$ field sensitivity. The performance of each RF coil combination was compared using EM field analysis under unnormalized ($|B_1^+|$, $|B_1^-|$, and $|E|$ fields) and normalized ($|B_1^+|$ field and SAR) conditions. The EM field simulations confirmed that the advanced form of the 16-ch RF coil + scHPM – w/wo-BCWE configuration provided enhanced $B_1^+$ field efficiency and $B_1^-$ field sensitivity compared to the BP-BC RF coil + scHPM – w/wo-MCWE at 7.0 T.

2. Materials and Methods

2.1. EM Field Simulation Setup

To evaluate the performance of the BP-BC RF coil + scHPM – w/wo-MCWE configuration and the 16-ch RF coil + scHPM – w/wo-BCWE configuration, 3D modeling and EM field calculations were performed using the FDTD method based on xFDTD simulation software version 6.6 (Remcom, Inc., State College, PA, USA) [81].

The components of the EM field simulation model were divided into three types: the Tx/Rx RF coil (BP-BC RF coil and the 16-ch RF coil) for RF transmission and reception, the WE (MCWE and BCWE) for $B_1^-$ field sensitivity improvement, and the scHPM for $B_1^+$ field efficiency improvement.

As Tx/Rx RF coils for human head MR imaging, the BP-BC RF coil and the 16-ch RF coil were designed to have the same diameter (330 mm) and length (150 mm). The BP-BC RF coil consists of 16 legs with a BP-BC type structural design, and the 16-ch RF coil consists of 16 surface elements. The BP-BC RF coil and the 16-ch RF coil were operated in the Tx/Rx mode, and they provided uniform RF distribution to the conductor using current sources (1 A) with a sinusoidal RF pulse. Current RF sources are used assuming an ideal current distribution, and RF coupling between RF elements is neglected [82–90]. The BP-BC RF coil consists of 16 legs with 80 current RF sources, and the 16-ch RF coil consists of 128 current RF sources (8 current RF sources per single channel). For ideal current driving condition through the micro-strip line of the transmission RF coil, multiple current RF sources were applied instead of the capacitors used for tuning and matching the RF coil. In addition, the micro-stripe lines constituting the RF coil were composed of a perfect electric conductor (PEC) instead of copper. Where the current intensity and geometrical phase of RF coils are defined at each current driving point, the EM field distribution can be calculated in an actual MRI experiment assuming target resonance RF frequency tuning and impedance matching with a tuning and matching capacitor. For $B_1^-$ field sensitivity improvement, the WE was designed and compared to the MCWE composed of 16 elements and the BCWE composed of the BP-BC element. The dimensions of the MCWE were 280 mm $\times$ 150 mm (diameter $\times$ length), and it consists of 16 elements, each with dimensions of 40 mm $\times$ 150 mm, which were placed between the legs of the BP-BC RF coil. The dimensions of the BCWE were identical to those of the MCWE, and the BCWE was designed with a 16-leg BP-BC structure. The MCWE and the BCWE were tuned to 300 MHz using only passive elements without RF sources. The MCWE was configured with a 6.26 pF tuning capacitor, and the BCWE was configured with a 3.84 pF tuning capacitor. All wireless elements used in MCWE were tuned to a frequency of 300 MHz using tuning capacitors by applying an RF source, and only the RF source was removed. BCWE, like MCWE, was used by tuning to a frequency of 300 MHz in the presence of an RF source and then removing only the RF source. Since each capacitor applied to MCWE and BCWE was used by applying geometric
phase information, the MCWE and BCWE were operated under circular polarization mode. For EM field simulations under ideal conditions, the Tx/Rx RF coils and WEs were made from perfect electric conductors.

The scHPM consists of a segmented cylinder with an outer diameter of 315 mm, an inner diameter of 295 mm, and a length of 150 mm. The scHPM was divided into 16 elements of the same size as that of the MCWE. The relative permittivity and loss tangent of the scHPM were 300 and 0.05, respectively. The width of each RF coil element and the scHPM was set to 40 mm so that the MCWE could be placed using the gap between the legs of the BP-BC RF coil. The scHPM was also located between the Tx/Rx RF coil and the WE.

2.2. EM Field Analysis

To compare the EM field sensitivities and distributions generated by RF coil configurations under ideal conditions, an oil-based cylindrical phantom model with a diameter of 224 mm and a length of 150 mm was used. To analyze the EM field effects of RF coil configurations on the human body, EM field simulations were performed using a human head model (a HIFI head model with 17 tissue characteristics provided by Remcom, Inc., State College, PA, USA).

In general, the uniform phantoms used in the EM field simulation of MRI mainly use oil-based, water-based, and average phantoms (mean value of dielectric properties with white matter and gray matter in the human head). In the 7.0 T MRI simulation, the oil-based phantom consists primarily of dielectric properties with a conductivity $0 \, \text{S} \cdot \text{m}^{-1}$ and a permittivity of 4. Furthermore, the water-based phantom using distilled water consists of dielectric properties with a conductivity $5 \times 10^{-5} \, \text{S} \cdot \text{m}^{-1}$ and a permittivity of 76.7. Moreover, the average phantom uses the average value of conductivity and permittivity of the white matter and gray matter in the human brain at a target frequency.

In this study, an oil-based cylindrical phantom was used, and EM field simulation using an oil-based cylindrical phantom was performed to evaluate the quantitative performance of the RF coil and verify the electromagnetic field distribution under ideal conditions. This was to minimize inhomogeneities, such as standing wave effects with shortened RF wavelength lengths in UHF MRI, distortions in peripheral region of the phantom due to high dielectric properties, and distortions of RF transmission and reception by shifted $B_1^+ / B_1^-$ fields [91,92].

The EM field simulation calculations of the oil-based cylindrical phantom were performed for a total of 26,292,960 voxels ($372 \times 372 \times 190$ cells), while the EM field simulation calculations of the human head model were performed for a total of 66,022,560 voxels ($522 \times 372 \times 340$ cells). The voxel resolution of the EM field simulations was set to 1 mm$^3$. For numerical analysis, EM field sensitivities and distributions composed of complex data matrices were analyzed in terms of the $|B_1^+|$, $|B_1^-|$, and $|E|$ fields, and the SAR map. The FDTD complex data matrices of the EM field results were computed using MATLAB (version 2020a, Mathworks, Inc., Natick, MA, USA) for data analysis of the $|B_1^+|$, $|B_1^-|$, and $|E|$ fields, and the SAR map.

EM field calculations were categorized into two groups: the unnormalized ($|B_1^+|$, $|B_1^-|$, and $|E|$ fields) and normalized ($|B_1^+|$ field and SAR map) cases. In the unnormalized case, we compared the EM field sensitivity changes for each combination of the Tx/Rx RF coil with the WE and the scHPM under the same conditions, and the normalized case was used to predict the actual experiments when a $90^\circ$ ($\pi/2$) RF pulse was induced. The total current applied to two types of RF coils in unnormalized condition was 1 A, and the equal current was applied to the combination of the two types of RF coils, as it was directly applied to the PEC without a capacitor at the position of the capacitor used to manufacture the actual RF coil. This method was applied to compare two types of RF coils under equal input RF power conditions, and the EM field generated by the equal amount of current applied to the conductor could be analyzed.

The unnormalized EM fields were compared using sensitivity-related factors and SD value-related factors as the relative differences, change rates, and capacity rates. The
relative sensitivity differences and relative standard deviation (SD) value differences were calculated by comparing the 16-ch RF coil combination and the BP-BC RF coil combination. The sensitivity change ratio and SD value change ratio were calculated by comparing the Tx/Rx RF coil alone result with the result of applying the WE and scHPM combinations. The capacity rate was used as a result of dividing the maximum EM field sensitivity value or the SD value of the 16-ch RF coil combinations by the maximum EM field sensitivity value or the SD value of the BP-BC RF coil combinations. The capability rate was a reference indicating how much the 16-ch RF coil combinations were relatively improved in terms of signal sensitivity and SD value compared to the BP-BC RF coil combination.

To calculate the normalized EM field, the efficiency of the unnormalized $|B_1^+|$ field generated by the RF coil configurations was measured at the 3D center point values of the RF coil, and the calculated normalization coefficient (Norm-COEF) was applied to the unnormalized EM field. Specifically, the 3D center point values of the RF coil configurations in the normalized $|B_1^+|$ field maps were calculated by assuming a flip angle of $\pi/2$; thus, the RF pulse was normalized at 1.957 $\mu$T [93,94]. By applying this calculated Norm-COEF to the unnormalized $|B_1^+|$ field and the SAR map, normalized $|B_1^+|$ field and SAR map analyses were performed assuming an actual 7.0 T MRI experiment.

The $|B_1|$ field is composed of two circularly polarized components: the $|B_1^+|$ and $|B_1^-|$ fields. The components of the $|B_1^+|$ and $|B_1^-|$ fields are defined as $B_1^+$ and $B_1^-$ in the rotating frame, respectively. $B_1^+$ and $B_1^-$ can be defined as follows:

$$B_1^+ = \left( \frac{(B_{1x} + iB_{1y})}{\sqrt{2}} \right), B_1^- = \left( \frac{(B_{1x} - iB_{1y})}{\sqrt{2}} \right)$$

Here, $B_{1x}$ and $B_{1y}$ are the $B_1$ components along the x- and y-axes, respectively.

The values of the unnormalized SAR map can be calculated as follows:

$$SAR(r) = \frac{\sigma}{2\rho} E^2 \propto \frac{dT}{dT}$$

where $E^2 = E\cdot E^*$ denotes the squared magnitude of the induced E field, $\rho = \rho(r)$ is the mass density ($\text{kg/m}^3$), and $T$ is the temperature ($\circ\text{K}$).

The whole-averaged SAR (mean SAR) and maximum SAR (max SAR) values were compared using the unaveraged SAR instead of the averaged SAR (1 g averaged SAR and 10 g averaged SAR) in normalized SAR maps. The mean SAR and max SAR values were calculated using mean and maximum values of unaveraged SAR in the entire tissue of the human head model, respectively. In addition, the unaveraged SAR was calculated by applying only the unaveraged SAR value of the entire human tissue region, excluding the massless region and the free space part of the human head model used.

3. Results and Discussion

To verify the effects of the scHPM and the WE on the EM field generated by the BP-BC RF coil, we performed EM field simulations (Figure 1) using the BP-BC RF coil and the 16-ch RF coil with the scHPM (w-scHPM) or without the scHPM (wo-scHPM). Next, to evaluate the impact of the simultaneous use of the WE and the scHPM, we compared the above configurations of the BP-BC RF coil with the WE (w-WE) or without the WE (wo-WE). The EM field results were simulated under unnormalized (Figures 2–4) and normalized (Figures 5 and 6) conditions.
Figure 1. Configurations of Tx/Rx RF coil combinations for EM field simulations using the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – w-scHPM – w-MCWE; (b,j) BP-BC RF coil – w-scHPM – w-MCWE; (c,k) 16-ch RF coil – w-scHPM – w-MCWE; (d,j) 16-ch RF coil – w-scHPM – w-BCWE; (e,m) BP-BC RF coil – w-scHPM – w-MCWE; (f,l) BP-BC RF coil – w-scHPM – w-MCWE; (g,o) 16-ch RF coil – w-scHPM – w-BCWE; (h,p) 16-ch RF coil – w-scHPM – w-BCWE.

Figure 1 shows the configurations of the numerical EM field simulation models. The EM field simulation models for the oil-based cylindrical phantom w-WE involved the BP-BC RF coil alone (BP-BC RF coil – w-scHPM – w-MCWE) (Figure 1a), the BP-BC RF coil with the scHPM (BP-BC RF coil – w-scHPM – w-MCWE) (Figure 1b), the 16-ch RF coil alone (16-ch RF coil – w-scHPM – w-BCWE) (Figure 1c), and the 16-ch RF coil with the scHPM (16-ch RF coil – w-scHPM – w-BCWE) (Figure 1d), while those w-WE involved the BP-BC RF coil with the WE (BP-BC RF coil – w-scHPM – w-MCWE) (Figure 1e), the BP-BC RF coil with the scHPM and the WE (BP-BC RF coil – w-scHPM – w-MCWE) (Figure 1f), the 16-ch RF coil with the WE (16-ch RF coil – w-scHPM – w-BCWE) (Figure 1g), and the 16-ch RF coil with the scHPM and the WE (16-ch RF coil – w-scHPM – w-BCWE) (Figure 1h).
The EM fields were calculated numerically, and the sensitivity values were compared using the 3D center point values of the RF coil, as shown in Figure 1. For the quantitative analysis of the numerically calculated EM fields, we compared the values for the $|B_1^+|$, $|B_1^-|$, and $|E|$ fields under the unnormalized condition. The 3D center point values (or max values) and standard deviation (SD) values for the unnormalized $|B_1^+|$, $|B_1^-|$, and $|E|$ field sensitivities are shown in Table 1. We validated the Tx/Rx configurations by comparing the EM field sensitivities under the unnormalized conditions and compared the oil-based cylindrical phantom and human head model.
Figure 3. Unnormalized $|\mathbf{B}_1^-|$ field distributions using the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – wo-scHPM – wo-MCWE; (b,j) BP-BC RF coil – w-scHPM – wo-MCWE; (c,k) 16-ch RF coil – wo-scHPM – wo-BCWE; (d,l) 16-ch RF coil – w-scHPM – w-BCWE; (e,m) BP-BC RF coil – wo-scHPM – wo-BCWE; (f,l) BP-BC RF coil – w-scHPM – w-MCWE; (g,o) 16-ch RF coil – wo-scHPM – w-BCWE; (h,p) 16-ch RF coil – w-scHPM – w-BCWE.

In the presence of the WE, not only the unnormalized $|\mathbf{B}_1^-|$ field improvement but also the unnormalized $|\mathbf{B}_1^+|$ field improvement was achieved, and in the 16-ch RF coil configuration with the BCWE, the unnormalized with the field improvement effect was much higher than that of the BP-BC RF coil configuration with the MCWE. This means that the proposed BCWE structure provided higher efficiency than the MCWE structure. In addition, the 16-ch RF coil configuration using scHPM showed an excellent $|\mathbf{B}_1^+|$ efficiency improvement effect compared to the BP-BC RF coil configuration using scHPM. This means that scHPM applied to 16-ch RF coils can improve $|\mathbf{B}_1^+|$ efficiency more effectively (Figure 2d,l).
The relative sensitivity difference was compared between the BP-BC RF coil and the 16-ch RF coil alone (16-ch RF coil – HPM – WE) and the 16-ch RF coil with the WE (16-ch RF coil – HPM – WE, 16-ch RF coil – w-scHPM – w-BCWE). The unnormalized EM field simulation models for the oil-based cylindrical phantom were analyzed by calculating the unnormalized EM field sensitivity and SD values (listed in Table 1) of each Tx/Rx RF coil configuration. To evaluate their field efficiency and field sensitivity, the relative difference between the maximum sensitivities was calculated (listed in Table 1). The relative sensitivity difference was compared between the BP-BC RF coil and the 16-ch RF coil configurations (Case 1: comparing the BP-BC RF coil – w-scHPM – w-BCWE and the 16-ch RF coil – w-scHPM – w-BCWE; Case 2: comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE; Case 3: comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE; Case 4: comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE).

In addition, to define the Tx/Rx RF coil combination that can provide optimal performance in terms of sensitivity and SD change ratios (listed in Tables 2 and 3), we compared Tx/Rx RF coils – w-scHPM – w-WE with Tx/Rx RF coil – w-scHPM – w-WE, Tx/Rx RF coil – w-scHPM – w-WE, and Tx/Rx RF coil – w-scHPM – w-WE.
Figure 5. Normalized $|B^*_1|$ field distributions using the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – wo-scHPM – wo-MCWE; (b,j) BP-BC RF coil – w-scHPM – wo-MCWE; (c,k) 16-ch RF coil – wo-scHPM – wo-BCE; (d,l) 16-ch RF coil – w-scHPM – wo-BCE; (e,m) BP-BC RF coil – wo-scHPM – w-MCWE; (f,l) BP-BC RF coil – w-scHPM – w-MCWE; (g,o) 16-ch RF coil – wo-scHPM – w-BCWE; (h,p) 16-ch RF coil – w-scHPM – w-BCWE.

Figure 2 (see also Figures S1 and S2) shows the unnormalized $|B^*_1|$ field distribution results using the oil-based cylindrical phantom and human head model. In the unnormalized $|B^*_1|$ field results, the unnormalized $|B^*_1|$ field distribution provided higher efficiency compared to the 16-ch RF coil alone. In particular, the sensitivity-related factors in all EM fields showed similar trends (listed in Tables 1–3).

The relative sensitivity differences in the unnormalized $|B^*_1|$ field were evaluated using the oil-based cylindrical phantom and the human head model. In the oil-based cylindrical phantom results, the relative sensitivity differences in the unnormalized $|B^*_1|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 3.004 times, 8.048 times, 9.894 times, and 7.072 times, respectively, under wo-WE/wo-scHPM (Figure 2a,c), wo-WE/w-scHPM (Figure 2b,d), w-WE/wo-scHPM (Figure 2e,g), and w-WE/w-scHPM (Figure 2f,h) conditions. In the human head model results, the relative sensitivity differences in the unnormalized $|B^*_1|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 0.942 times, 1.754 times, 1.746 times, and 3.056 times, respectively, under wo-WE/wo-
scHPM (Figure 2i,k), wo-WE/w-scHPM (Figure 2j,l), w-WE/w-scHPM (Figure 2m,o), and w-WE/w-scHPM (Figure 2n,p) conditions.

Figure 6. Normalized SAR maps constructed using Norm-COEF values: (a) BP-BC RF coil – w-scHPM – w-WE; (b) BP-BC RF coil – w-scHPM – w-WE; (c) 16-ch RF coil – w-scHPM – w-WE; (d) 16-ch RF coil – w-scHPM – w-WE; (e) BP-BC RF coil – w-scHPM – w-WE; (f) BP-BC RF coil – w-scHPM – w-WE; (g) 16-ch RF coil – w-scHPM – w-WE; (h) 16-ch RF coil – w-scHPM – w-WE.

The sensitivity change ratios of the unnormalized \( |B_i^T| \) field using the oil-based cylindrical phantom were calculated as approximately 1.178 times (for BP-BC RF coil configurations comparing Figures 2a and 2b), 1.150 times (for BP-BC RF coil configurations comparing Figures 2a and 2e), respectively. 3.156 times (for 16-ch RF coil configurations comparing Figures 2c and 2d), 3.087 times (for 16-ch RF coil configurations comparing Figures 2c and 2g), and 4.224 times (for 16-ch RF coil configurations comparing Figures 2c and 2h). In the human head model results, the sensitivity change ratios of the unnormalized \( |B_i^T| \) field were calculated as approximately 1.036 times (for BP-BC RF coil configurations comparing Figures 2i and 2j), respectively. 1.083 times (for BP-BC RF coil configurations comparing Figures 2i and 2m), 1.328 times (for BP-BC RF coil configurations comparing Figures 2i and 2n), 1.928 times (for 16-ch RF coil configurations comparing Figures 2k and 2l), 2.007 times (for 16-ch RF coil configurations comparing Figures 2k and 2o), and 4.309 times (for 16-ch RF coil configurations comparing Figures 2k and 2p).

The sensitivity capability rates of the unnormalized \( |B_i^T| \) field using the oil-based cylindrical phantom were calculated as 267.909% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE), 329.368% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE), and 235.423% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE). The sensitivity capability rates of the unnormalized \( |B_i^T| \) field using the human head model were calculated as 186.132% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE), 185.282% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE), and 324.391% (comparing the BP-BC RF coil – w-scHPM – w-WE and the 16-ch RF coil – w-scHPM – w-WE).
Table 1. Maximum and SD values of unnormalized EM field simulation results ($|B_1^+|$, $|B_1^-|$, and $|E|$ fields) using the oil-based cylindrical phantom and human head model.

|                               | BP-BC RF Coil | BP-BC RF Coil | 16-ch RF Coil | 16-ch RF Coil |
|-------------------------------|---------------|---------------|---------------|---------------|
|                               |               |               |               |               |
|                               | wo-scHPM      | w-scHPM       | wo-scHPM      | w-scHPM       |
| Max values of the unnormalized $|B_1^+|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| w-WE                          | 0.247         | 0.291         | 0.742         | 2.342         |
| w-WE                          | 0.284         | 0.443         | 2.810         | 3.133         |
| wo-WE                         | 0.588         | 0.609         | 0.554         | 1.068         |
| w-WE                          | 0.637         | 0.781         | 1.112         | 2.387         |
| Human head model              |               |               |               |               |
| w-WE                          | 0.004         | 0.004         | 0.007         | 0.009         |
| w-WE                          | 0.003         | 0.011         | 0.058         | 0.071         |
| wo-WE                         | 0.093         | 0.094         | 0.086         | 0.165         |
| w-WE                          | 0.100         | 0.120         | 0.173         | 0.371         |
| SD values of the unnormalized $|B_1^-|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| w-WE                          | 0.050         | 0.071         | 0.156         | 0.537         |
| w-WE                          | 0.058         | 0.128         | 0.659         | 0.784         |
| wo-WE                         | 0.271         | 0.273         | 0.256         | 0.487         |
| w-WE                          | 0.292         | 0.346         | 0.508         | 1.081         |
| Human head model              |               |               |               |               |
| w-WE                          | 0.014         | 0.016         | 0.041         | 0.137         |
| w-WE                          | 0.014         | 0.025         | 0.154         | 0.170         |
| wo-WE                         | 0.048         | 0.049         | 0.046         | 0.088         |
| w-WE                          | 0.052         | 0.060         | 0.091         | 0.191         |
| Max values of the unnormalized $|E|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| w-WE                          | 260.257       | 202.869       | 644.104       | 2163.678      |
| w-WE                          | 224.388       | 296.424       | 1998.820      | 1900.393      |
| wo-WE                         | 727.420       | 711.303       | 603.219       | 1186.803      |
| w-WE                          | 740.012       | 918.380       | 1230.528      | 2725.248      |
| Human head model              |               |               |               |               |
| w-WE                          | 60.182        | 44.438        | 154.113       | 516.632       |
| w-WE                          | 50.242        | 64.141        | 463.342       | 422.434       |
| wo-WE                         | 62.877        | 63.044        | 57.199        | 110.145       |
| w-WE                          | 66.373        | 79.834        | 115.689       | 246.113       |

The sensitivity-related factors of the unnormalized $|B_1^+|$ field results showed remarkably similar tendencies, indicating improved sensitivity in the 16-ch RF coil configurations compared to the BP-BC RF coil configurations. In particular, the sensitivity change ratio and the sensitivity capability rate showed extremely similar tendencies.

The SD value relative differences in the unnormalized $|B_1^+|$ field were evaluated using the oil-based cylindrical phantom and human head model. In the oil-based cylindrical phantom results, the relative sensitivity differences between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 1.750 times, 2.250 times, 19.333 times, and 6.455 times, respectively, under wo-WE/wo-scHPM (Figure 2a,c), wo-WE/w-scHPM (Figure 2b,d), w-WE/wo-scHPM (Figure 2e,g), and w-WE/w-scHPM (Figure 2f,h) conditions. In the human head model results, the relative SD value differences in the unnormalized $|B_1^+|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 0.925 times, 1.755 times, 1.730 times, and 3.092 times, respectively, under wo-WE/wo-scHPM (Figure 2i,k), wo-WE/w-scHPM (Figure 2j,l), w-WE/wo-scHPM (Figure 2m,o), and w-WE/w-scHPM (Figure 2n,p) conditions.
Table 2. Unnormalized EM field ($|B_1^+|$, $|B_1^-|$, $|B^-_1|$, and $|E|$ fields) sensitivity change ratios for scHPM – wo-WE, wo-WE, and wo-scHPM – wo-WE compared to the Tx/Rx RF coil alone using oil-based cylindrical phantom and human head model.

|                         | BP-BC RF Coil               | 16-ch RF Coil               | Capability Rate |
|-------------------------|-----------------------------|-----------------------------|-----------------|
|                         | w-scHPM – wo-WE            | w-scHPM – wo-WE             |                 |
| Sensitivity change ratio of the unnormalized $|B_1^+|$ field (%) | 117.814                     | 315.633                    | 267.909         |
|                         | 114.980                     | 378.706                    | 329.368         |
|                         | 132.823                     | 430.866                    | 324.391         |
| Oil-based cylindrical phantom |                              |                             |                 |
|                         | w-scHPM – wo-WE            | w-scHPM – wo-WE             |                 |
|                         | 103.571                     | 192.780                    | 186.132         |
|                         | 108.333                     | 200.722                    | 182.282         |
|                         | 132.823                     | 430.866                    | 324.391         |
| Human head model        | w-scHPM – wo-WE            | w-scHPM – wo-WE             |                 |
| Sensitivity change ratio of the unnormalized $|B^-_1|$ field (%) | 142.000                     | 344.231                    | 242.146         |
|                         | 116.000                     | 422.436                    | 364.169         |
|                         | 256.000                     | 502.564                    | 196.314         |
|                         | 100.738                     | 190.234                    | 188.841         |
|                         | 107.749                     | 198.438                    | 184.166         |
|                         | 127.675                     | 422.266                    | 330.734         |
| Sensitivity change ratio of the unnormalized $|E|$ field (%) | 77.950                      | 335.921                    | 430.947         |
|                         | 86.218                      | 310.326                    | 359.932         |
|                         | 113.897                     | 295.044                    | 259.046         |
| Oil-based cylindrical phantom | 97.784                      | 196.745                    | 201.203         |
|                         | 101.731                     | 203.994                    | 200.522         |
|                         | 126.252                     | 451.784                    | 357.844         |
| Human head model        | w-scHPM – wo-WE            | w-scHPM – wo-WE             |                 |

The SD value change ratios of the unnormalized $|B_1^+|$ field using the oil-based cylindrical phantom were calculated as approximately 1.000 times (for BP-BC RF coil configurations comparing Figures 2a and 2b), 0.750 times (for BP-BC RF coil configurations comparing Figures 2a and 2e), 2.750 times (for BP-BC RF coil configurations comparing Figures 2a and 2f), 1.286 times (for 16-ch RF coil configurations comparing Figures 2c and 2d), 8.286 times (for 16-ch RF coil configurations comparing Figures 2c and 2g), and 10.143 times (for 16-ch RF coil configurations comparing Figures 2c and 2h). In the human head model results, the SD value change ratios of the unnormalized $|B_1^+|$ field were calculated as approximately 1.011 times (for BP-BC RF coil configurations comparing Figures 2i and 2j), 1.075 times (for BP-BC RF coil configurations comparing Figures 2i and 2m), 1.290 times (for BP-BC RF coil configurations comparing Figures 2i and 2n), 1.919 times (for 16-ch RF coil configurations comparing Figures 2k and 2l), 2.012 times (for 16-ch RF coil configurations comparing Figures 2k and 2o), and 4.314 times (for 16-ch RF coil configurations comparing Figures 2k and 2p).

The SD value capability rates of the unnormalized $|B_1^+|$ field using the oil-based cylindrical phantom were calculated as 128.571% (comparing the BP-BC RF coil – wo-scHPM – wo-MCWE and the 16-ch RF coil – wo-scHPM – wo-BCWE), 1104.762% (comparing the BP-BC RF coil – wo-scHPM – wo-MCWE and the 16-ch RF coil – wo-scHPM – wo-BCWE), 368.312% (comparing the BP-BC RF coil – wo-scHPM – wo-MCWE and the 16-ch RF coil – wo-scHPM – wo-BCWE). The sensitivity capability rates using the human head model were
calculated as 189.819% (comparing the BP-BC RF coil – \textit{w-scHPM – wo-MCWE} and the 16-ch RF coil – \textit{w-scHPM – wo-BCWE}), 187.081% (comparing the BP-BC RF coil – \textit{w-scHPM – wo-MCWE} and the 16-ch RF coil – \textit{w-scHPM – w-BCWE}), and 334.331% (comparing the BP-BC RF coil – \textit{w-scHPM – w-MCWE} and the 16-ch RF coil – \textit{w-scHPM – w-BCWE}).

Table 3. SD change ratios of the unnormalized EM field (|B$_1^+$|, |B$_1^-$|, and |E| fields) for scHPM – \textit{wo-WE}, \textit{w-WE}, and \textit{scHPM – w-WE} compared to the Tx/Rx RF coil alone using the oil-based cylindrical phantom and human head model.

|                   | BP-BC RF Coil | 16-ch RF Coil | Capability Rate |
|-------------------|---------------|---------------|-----------------|
|                   | \textit{w-scHPM} – \textit{wo-WE} | \textit{w-scHPM} – \textit{w-WE} | \textit{wo-MCWE} – \textit{w-BCWE} | \textit{wo-MCWE} – \textit{w-BCWE} |
| **SD change ratio of the unnormalized |B$_1^+$| field (%)** |
| **Oil-based cylindrical phantom** | 100.000 | 128.571 | 128.571 | 1104.762 |
| \textit{w-scHPM} – \textit{wo-WE} | 100.000 | 128.571 | 128.571 | 1104.762 |
| \textit{w-scHPM} – \textit{w-WE} | 100.000 | 128.571 | 128.571 | 1104.762 |
| **SD change ratio of the unnormalized |B$_1^-$| field (%)** |
| **Oil-based cylindrical phantom** | 114.286 | 334.146 | 292.378 | 292.378 |
| \textit{w-scHPM} – \textit{wo-WE} | 100.000 | 375.601 | 375.610 | 375.610 |
| **SD change ratio of the unnormalized |E| field (%)** |
| **Oil-based cylindrical phantom** | 73.839 | 335.229 | 453.999 | 453.999 |
| \textit{w-scHPM} – \textit{wo-WE} | 83.484 | 300.650 | 360.129 | 360.129 |

As can be seen from the unnormalized |B$_1^+$| field results, the sensitivity value, sensitivity change ratio, and SD value change ratio of the unnormalized |B$_1^+$| field were greatly influenced by Tx/Rx RF coil configurations with the scHPM and the WE. In terms of the SD value change ratio of the unnormalized |B$_1^+$| field, the SD value change ratio of the 16-ch RF coil alone – \textit{wo-BCWE} compared with those of 16-ch RF coil + scHPM, the 16-ch RF coil alone – \textit{w-BCWE}, and the 16-ch RF coil + scHPM – \textit{w-BCWE} using the oil-based cylindrical phantom showed very contradictory results compared to the SD value change ratios of other EM fields. As shown in Figure 2, in the unnormalized |B$_1^+$| field results, the relative differences between the oil-based cylindrical phantom and the human head model results proved that the scHPM \textit{wo-BCWE} applied to the 16-ch RF coil was more sensitive to dielectric properties. The extremely different SD value change ratio between the |B$_1^+$| field results of the oil-based cylindrical phantom and the human head model was that the oil-based cylindrical phantom was relatively closer to the 16-RF coil configuration applied scHPM and BCWE than the human head model. In addition, compared to the human head
model, there was a large difference in the rate of change in SD values due to the dielectric properties of the oil-based phantom. However, in the 16-ch RF coil configurations, the SD values of the unnormalized $|B_1^w|$ fields changed relatively significantly, while the SD values of unnormalized $|B_1^{sc}|$ and $|E|$ fields showed a change similar to the sensitivity change ratio of the unnormalized $|B_1^{sc}|$ and $|E|$ field.

However, unlike the sensitivity-related factors of the unnormalized $|B_1^w|$ field results, the SD value-related factors of the unnormalized $|B_1^w|$ field results increased dramatically due to the WE in the oil-based cylindrical phantom results. This rapid increase in the SD value-related factors was expected to intensify due to the BCWE, which consists of volume shapes that adhere to the cylindrical phantom, and its low dielectric properties. On the other hand, in the human head model results, it was confirmed that the sensitivity-related and SD value-related factors of the unnormalized $|B_1^w|$ field showed a similar tendency. A sharp increase in the SD value of the BCWE was observed using the oil-based cylindrical phantom model, but a sharp change in the SD value-related factors could not be observed in the human head model. Based on the oil-based cylindrical phantom results, a rapid change in the SD value-related factors was observed in the 16-ch RF coil configuration with the BCWE, but the advantage of the sensitivity-related factors increased in terms of RF power. Thus, it is worth recommending the use of the 16-ch RF coil – $\omega$-scHPM – $\omega$-BCWE and the 16-ch RF coil – $\omega$-scHPM – $\omega$-BCWE.

Figure 3 (see also Figures S3 and S4) shows the unnormalized $|B_1^w|$ field distribution results using the oil-based cylindrical phantom and human head model. From the unnormalized $|B_1^w|$ field results shown in Figure 3, the combinations of $\omega$-scHPM and $\omega$-WE/\(\omega\)-BCWE (Figure 3d–h) allowed us to observe extreme changes in $|B_1^w|$ field sensitivity in the periphery region of the oil-based cylindrical phantom, as it was located close enough to the BCWE and the scHPM. Similar abnormal unnormalized $|B_1^w|$ field patterns in the periphery region were not observed in the human head model results, but the unnormalized $|B_1^w|$ field penetration into the deep region of the human head model was confirmed. The unnormalized $|B_1^w|$ field distribution also provided higher sensitivity compared to the 16-ch RF coil alone, depending on how WE and scHPM combinations were applied.

The relative sensitivity differences in the unnormalized $|B_1^w|$ field were evaluated using the oil-based cylindrical phantom and human head model. In the oil-based cylindrical phantom results, the relative sensitivity differences in the unnormalized $|B_1^w|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 3.120 times, 7.563 times, 11.362 times, and 6.125 times, respectively, under $\omega$-WE/$\omega$-BCWE (Figure 3a,c), $\omega$-WE/$\omega$-scHPM (Figure 3b,d), $\omega$-WE/$\omega$-scHPM (Figure 3e,g), and $\omega$-WE/$\omega$-scHPM (Figure 3f,h) conditions. In the human head model results, the relative sensitivity differences in the unnormalized $|B_1^w|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 0.945 times, 1.784 times, 1.740 times, and 3.124 times, respectively, under $\omega$-WE/$\omega$-scHPM (Figure 3i,k), $\omega$-WE/$\omega$-scHPM (Figure 3j,l), $\omega$-WE/$\omega$-scHPM (Figure 3m,o), and $\omega$-WE/$\omega$-scHPM (Figure 3n,p) conditions.

The sensitivity change ratios of the unnormalized $|B_1^w|$ field using the oil-based cylindrical phantom were calculated as approximately 1.420 times (BP-BC RF coil configurations—comparison of Figures 3a and 3b), 1.160 times (BP-BC RF coil configurations—comparison of Figures 3a and 3e), 2.560 times (BP-BC RF coil configurations—comparison of Figures 3a and 3f), 3.442 times (16-ch RF coil configurations—comparison of Figures 3c and 3d), 4.224 times (16-ch RF coil configurations—comparison of Figures 3c and 3g), and 5.026 times (16-ch RF coil configurations—comparison of Figures 3c and 3h). In the human head model results, the sensitivity change ratios of the unnormalized $|B_1^w|$ field were calculated as approximately 1.007 times (BP-BC RF coil configurations—comparison of Figures 3i and 3j), 1.077 times (BP-BC RF coil configurations—comparison of Figures 3i and 3m), 1.277 times (BP-BC RF coil configurations—comparison of Figures 3i and 3n), 1.902 times (16-ch RF coil configurations—comparison of Figures 3k and 3l), 1.984 times (16-ch RF coil configurations—comparison of Figures 3k and 3o), and 4.223 times (16-ch RF coil configurations—comparison of Figures 3k and 3p).
The sensitivity capability rates of the unnormalized $|B_{\text{u}}|_1$ field using the oil-based cylindrical phantom were calculated as 242.416% (comparing the BP-BC RF coil – w-scHPM – w-SCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 364.169% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 196.314% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE). The sensitivity capability rates of the unnormalized $|B_{\text{u}}|_1$ field using the human head model were calculated as 188.841% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 184.166% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 330.734% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE).

The sensitivity-related factors of the unnormalized $|B_{\text{u}}|_1$ field results indicated significantly improved sensitivity in the 16-ch RF coil configurations compared to the BP-BC RF coil configurations. Particularly, in the oil-based cylindrical phantom results, sensitivity-related factors were detected at the position where the 16-ch RF coil and the scHPM structure were disposed in the periphery region of the oil phantom. As shown in Figure 3g, the unnormalized $|B_{\text{u}}|_1$ field sensitivity rapidly increased due to the influence of the BCWE located inside the 16-ch RF coil. These results confirmed that the unnormalized $|B_{\text{u}}|_1$ field improvement effect of the 16-ch RF coil with the BCWE was significantly better than that of the BP-BC RF coil with the MCWE, and that the BCWE structure was much more effective than the MCWE structure when the WE was applied to the Tx/Rx RF coil.

The relative SD value differences of the unnormalized $|B_{\text{u}}|_1$ field were evaluated using the oil-based cylindrical phantom and human head model. In the oil-based cylindrical phantom results, the relative sensitivity differences between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 2.929 times, 8.563 times, 11.000 times, and 6.800 times, respectively, under wo-WE/wo-scHPM (Figure 3a,c), wo-WE/w-scHPM (Figure 3b,d), wo-WE/wo-scHPM (Figure 3e,g), and wo-WE/w-scHPM (Figure 3f,h) conditions. In the human head model results, the relative SD value differences of the unnormalized $|B_{\text{u}}|_1$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 0.959 times, 1.796 times, 1.750 times, and 3.183 times, respectively, under wo-WE/wo-scHPM (Figure 3i,k), wo-WE/w-scHPM (Figure 3j,l), wo-WE/wo-scHPM (Figure 3m,o), and wo-WE/w-scHPM (Figure 3n,p) conditions.

The SD value change ratios of the unnormalized $|B_{\text{u}}|_1$ field using the oil-based cylindrical phantom were calculated as approximately 1.143 times (BP-BC RF coil configurations—comparison of Figures 3a and 3b), 1.000 times (BP-BC RF coil configurations—comparison of Figures 3a and 3e), 1.786 times (BP-BC RF coil configurations—comparison of Figures 3a and 3f), 3.341 times (16-ch RF coil configurations—comparison of Figures 3c and 3d), 3.756 times (16-ch RF coil configurations—comparison of Figures 3c and 3g), and 4.146 times (16-ch RF coil configurations—comparison of Figures 3c and 3h). In the human head model results, the SD value change ratios of the unnormalized $|B_{\text{u}}|_1$ field were calculated as approximately 1.021 times (BP-BC RF coil configurations—comparison of Figures 3i and 3j), 1.083 times (BP-BC RF coil configurations—comparison of Figures 3i and 3m), 1.250 times (BP-BC RF coil configurations—comparison of Figures 3i and 3n), 1.913 times (16-ch RF coil configurations—comparison of Figures 3k and 3l), 1.978 times (16-ch RF coil configurations—comparison of Figures 3k and 3o), and 4.152 times (16-ch RF coil configurations—comparison of Figures 3k and 3p).

The SD value capability rates of the unnormalized $|B_{\text{u}}|_1$ field using the oil-based cylindrical phantom was calculated as 128.571% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 1104.762% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 368.312% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE). The sensitivity capability rates using the human head model were calculated as 189.819% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 187.081% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 330.734% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE).
was confirmed that the unnormalized |E| field was more concentrated in the Tx/Rx RF (BP-BC RF coil configurations—comparison of Figures 4i and 4j), 1.017 times (BP-BC RF coil conditions. In the oil-based cylindrical phantom results, the relative sensitivity differences in the unnormalized |E| field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated as 201.203% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 259.046% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE). The sensitivity capability rates of the unnormalized |B1+| field using the human head model were calculated as 201.203% (comparing the BP-BC RF coil –
The sensitivity-related factors of the unnormalized $|E|$ field results tended to contradict the results of other unnormalized EM fields ($|B_1^+|$ and $|B_1^-|$ fields), except for the unnormalized $|E|$ field in the 16-ch RF coil configuration in the oil-based cylindrical phantom results. In particular, in the oil-based cylindrical phantom results, unlike the human head model results, the sensitivity change ratio of the unnormalized $|E|$ field was the highest for 16-ch RF coil – w-scHPM – w-BCWE and the lowest for 16-ch RF coil – w-scHPM – w-BCWE. These results showed that uniform phantoms with low dielectric properties, located very uniformly and close to the scHPM, such as the oil-based cylindrical phantom, can increase the concentration of the unnormalized $|E|$ field by the scHPM. In addition, it was confirmed that the unnormalized $|E|$ field concentration was higher in the case of the BP-BC RF coil configuration with the WE (BP-BC RF coil – w-scHPM – w-MCWE), whereas the unnormalized $|E|$ field concentration was higher in the periphery region of the oil-based cylindrical phantom owing to the scHPM than in the 16-ch RF coil configuration (16-ch RF coil – w-scHPM – w-BCWE).

The relative SD value differences in the unnormalized $|E|$ field were evaluated using the oil-based cylindrical phantom and human head model. In the oil-based cylindrical phantom results, the relative SD value differences between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 2.561 times, 11.626 times, 0.910 times, and 1.747 times, respectively, under $w$-WE/$w$-scHPM (Figure 4a,c), $w$-WE/$w$-scHPM (Figure 4b,d), $w$-WE/$w$-scHPM (Figure 4e,g), and $w$-WE/$w$-scHPM (Figure 4f,h) conditions. In the human head model results, the relative SD value differences in the unnormalized $|E|$ field between the BP-BC RF coil and the 16-ch RF coil configurations were calculated to be approximately 0.910 times, 1.747 times, 1.747 times, and 3.083 times, respectively, under $w$-WE/$w$-scHPM (Figure 4i,k), $w$-WE/$w$-scHPM (Figure 4j,l), $w$-WE/$w$-scHPM (Figure 4m,o), and $w$-WE/$w$-scHPM (Figure 4n,p) conditions.

The SD value change ratios of the unnormalized $|E|$ field using the oil-based cylindrical phantom were calculated as approximately 0.738 times (BP-BC RF coil configurations—comparison of Figures 4a and 4b), 0.835 times (BP-BC RF coil configurations—comparison of Figures 4a and 4e), 1.066 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f), 3.352 times (16-ch RF coil configurations—comparison of Figures 4c and 4d), 3.007 times (16-ch RF coil configurations—comparison of Figures 4a and 4e), 1.066 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f), 0.835 times (BP-BC RF coil configurations—comparison of Figures 4a and 4b), and 0.738 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f). The SD value change ratios of the unnormalized $|E|$ field were calculated as approximately 1.003 times (BP-BC RF coil configurations—comparison of Figures 4a and 4b), 0.835 times (BP-BC RF coil configurations—comparison of Figures 4a and 4e), 1.066 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f), 3.352 times (16-ch RF coil configurations—comparison of Figures 4c and 4d), 3.007 times (16-ch RF coil configurations—comparison of Figures 4a and 4e), 1.066 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f), 0.835 times (BP-BC RF coil configurations—comparison of Figures 4a and 4b), and 0.738 times (BP-BC RF coil configurations—comparison of Figures 4a and 4f).

The SD value capability rates of the unnormalized $|E|$ field using the oil-based cylindrical phantom were calculated as 453.999% (comparing the BP-BC RF coil – w-scHPM – w-BCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 360.129% (comparing the BP-BC RF coil – w-scHPM – w-BCWE), and 257.186% (comparing the BP-BC RF coil – w-scHPM – w-MCWE). The sensitivity capability rates using the human head model were calculated as 192.056% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), 191.606% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE), and 338.888% (comparing the BP-BC RF coil – w-scHPM – w-MCWE and the 16-ch RF coil – w-scHPM – w-BCWE).
was calculated using the SD values (listed in Table 1) and mean values (listed in Table 4) of the 16-ch RF coil – w HPM and BP-BC RF coil – w HPM configurations were calculated as 0.383 (16-ch RF coil – w HPM) and 0.273 (BP-BC RF coil – w HPM) configurations were calculated as 0.362 (BP-BC RF coil – w HPM) and 0.076 (16-ch RF coil – w HPM) configurations were calculated as 0.264 (16-ch RF coil – w HPM) and 0.521 (BP-BC RF coil – w HPM) configurations were calculated as 0.793 (BP-BC RF coil – w HPM – w-MCWE), 0.192 (16-ch RF coil – w HPM – w-MCWE), 0.085 (16-ch RF coil – w HPM – w-BCWE), and 0.076 (16-ch RF coil – w HPM – w-BCWE). The Norm-COEF values of the 16-ch RF coil configurations were calculated as 0.689 (BP-BC RF coil – w HPM – w-MCWE), and 0.521 (BP-BC RF coil – w HPM – w-MCWE). The Norm-COEF values of the 16-ch RF coil configurations were calculated as 0.264 (16-ch RF coil – w HPM – w-BCWE), 0.071 (16-ch RF coil – w HPM – w-BCWE), 0.085 (16-ch RF coil – w HPM – w-BCWE), and 0.076 (16-ch RF coil – w HPM – w-BCWE). The Norm-COEF values of the 16-ch RF coil configurations were calculated as 0.383 (16-ch RF coil – w HPM – w-BCWE), 0.087 (16-ch RF coil – w HPM – w-BCWE), 0.192 (16-ch RF coil – w HPM – w-BCWE), and 0.199 (16-ch RF coil – w HPM – w-BCWE).
Table 4. Coefficient of variation (SD/mean) in unnormalized EM field simulation results ($|B_1^+|$, $|B_1^-|$, and $|E|$ fields) using the oil-based cylindrical phantom and human head model.

|                      | BP-BC RF Coil | BP-BC RF Coil | 16-ch RF Coil | 16-ch RF Coil |
|----------------------|---------------|---------------|---------------|---------------|
|                      | wo-scHPM      | w-scHPM       | wo-scHPM      | w-scHPM       |
| Mean values of the unnormalized $|B_1^+|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 0.237 | 0.271 | 0.722 | 2.827 |
| w-WE | 0.276 | 0.383 | 2.312 | 2.656 |
| Human head model |               |               |               |               |
| wo-WE | 0.251 | 0.269 | 0.251 | 0.499 |
| w-WE | 0.280 | 0.350 | 0.512 | 1.112 |
| Mean values of the unnormalized $|B_1^-|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 0.026 | 0.024 | 0.063 | 0.248 |
| w-WE | 0.023 | 0.038 | 0.234 | 0.262 |
| Human head model |               |               |               |               |
| wo-WE | 0.137 | 0.131 | 0.118 | 0.220 |
| w-WE | 0.141 | 0.164 | 0.229 | 0.480 |
| Mean values of the unnormalized $|E|$ field |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 170.457 | 135.949 | 141.255 | 140.482 |
| w-WE | 152.245 | 189.240 | 150.249 | 180.131 |
| Human head model |               |               |               |               |
| wo-WE | 37.067 | 34.905 | 34.263 | 33.066 |
| w-WE | 35.753 | 34.325 | 33.789 | 33.363 |
| CV (SD/mean) of the unnormalized $|B_1^+|$ field (%) |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 1.688 | 1.4777 | 0.9695 | 0.3184 |
| w-WE | 1.086 | 2.8728 | 2.5087 | 2.6732 |
| Human head model |               |               |               |               |
| wo-WE | 3.067 | 34.905 | 34.263 | 33.066 |
| w-WE | 35.753 | 34.325 | 33.789 | 33.363 |
| CV (SD/mean) of the unnormalized $|B_1^-|$ field (%) |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 53.640 | 65.871 | 65.600 | 55.153 |
| w-WE | 60.450 | 65.036 | 65.840 | 64.886 |
| Human head model |               |               |               |               |
| wo-WE | 35.139 | 37.433 | 39.082 | 40.073 |
| w-WE | 36.932 | 36.563 | 39.808 | 39.800 |
| CV (SD/mean) of the unnormalized $|E|$ field (%) |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| wo-WE | 35.306 | 32.687 | 109.103 | 367.758 |
| w-WE | 33.001 | 33.894 | 308.384 | 234.515 |
| Human head model |               |               |               |               |
| wo-WE | 44.513 | 44.877 | 43.574 | 43.648 |
| w-WE | 44.175 | 44.320 | 43.325 | 43.452 |

Table 5. Normalization coefficient (Norm-COEF).

|                      | BP-BC RF Coil | BP-BC RF Coil | 16-ch RF Coil | 16-ch RF Coil |
|----------------------|---------------|---------------|---------------|---------------|
|                      | wo-scHPM      | w-scHPM       | wo-scHPM      | w-scHPM       |
| Norm-COEF (Oil-based cylindrical phantom) |               |               |               |               |
| wo-WE | 0.793 | 0.723 | 0.264 | 0.085 |
| w-WE | 0.689 | 0.521 | 0.076 | 0.071 |
| Norm-COEF (Human head model) |               |               |               |               |
| wo-WE | 0.362 | 0.352 | 0.383 | 0.199 |
| w-WE | 0.334 | 0.273 | 0.192 | 0.087 |

The normalized $|B_1^+|$ fields after the application of the Norm-COEFS (listed in Table 5) were calculated, as shown in Figure 5 (also see Figures S7 and S8). The Norm-COEF of the 16-ch RF coil with the scHPM or WE was significantly lower compared to the BP-BC RF coil with the scHPM or WE. In other words, applying the scHPM or WE to the 16-ch RF coil...
excited an $^1$H by a 90° RF pulse at the center point of the target object with less RF power than applying the scHPM or WE to the BP-BC RF coil. In terms of Norm-COEF, the 16-ch RF coil – $w$-scHPM – $w$-BCWE provided 3.718 and 4.402 times lower RF power requirements in the oil-based cylindrical phantom and human head model, respectively, compared to 16-ch RF coil – $wo$-scHPM – $wo$-BCWE. The 16-ch RF coil – $w$-scHPM – $w$-BCWE also provided 11.169 and 4.161 times lower RF power requirements in the oil-based cylindrical phantom and human head model, respectively, compared to the BP-BC RF – $wo$-scHPM – $wo$-MCWE.

In the normalized $|B_1^+|$ field results using oil-based cylindrical phantom and human head models, the center values of the BP-BC RF coil and the 16-ch RF coil were both normalized to values when a 90° RF pulse was applied. For the comparison of uniformity of 16-ch RF coil combinations and BP-BC RF coil combinations, the mean value, SD value, and CV of the normalized $|B_1^+|$ field were evaluated as shown in Figure 5 and Table 6. The 16-ch RF coil configurations produced high unnormalized $|B_1^+|$ field efficiency (as listed in Table 5) without degrading the uniformity of the normalized $|B_1^+|$ field distribution compared to the BP-BC RF coil configurations (as listed in Table 6). These results were equally observed in the human head model results, and it was confirmed that the 16-ch RF coil configuration provided a more sensitive and uniform normalized $|B_1^+|$ field distribution.

Table 6. Mean values, SD values, and CV (SD/mean) in normalized $|B_1^+|$ field simulation results using the oil-based cylindrical phantom and human head model.

|                      | BP-BC RF Coil | BP-BC RF Coil | 16-ch RF Coil | 16-ch RF Coil |
|----------------------|---------------|---------------|---------------|---------------|
|                      | $wo$-scHPM    | $w$-scHPM     | $wo$-scHPM    | $w$-scHPM     |
| Oil-based cylindrical phantom |               |               |               |               |
| $wo$-WE              | 0.191         | 0.200         | 0.188         | 0.196         |
| $w$-WE               | 0.196         | 0.203         | 0.190         | 0.200         |
| Human head model     |               |               |               |               |
| $wo$-WE              | 0.096         | 0.100         | 0.091         | 0.095         |
| $w$-WE               | 0.098         | 0.099         | 0.094         | 0.096         |
| SD values of the normalized $|B_1^+|$ field $\times 10^{-5}$ (µT) |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| $wo$-WE              | 0.002         | 0.005         | 0.003         | 0.003         |
| $w$-WE               | 0.001         | 0.004         | 0.002         | 0.006         |
| Human head model     |               |               |               |               |
| $wo$-WE              | 0.033         | 0.033         | 0.034         | 0.033         |
| $w$-WE               | 0.033         | 0.033         | 0.034         | 0.033         |
| CV (SD/mean)         |               |               |               |               |
| Oil-based cylindrical phantom |               |               |               |               |
| $wo$-WE              | 1.011         | 2.527         | 1.569         | 1.430         |
| $w$-WE               | 0.409         | 2.183         | 1.022         | 2.902         |
| Human head model     |               |               |               |               |
| $wo$-WE              | 34.304        | 33.084        | 37.142        | 35.043        |
| $w$-WE               | 33.745        | 33.256        | 35.866        | 34.429        |

The SAR maps after the application of the Norm-COEFS were calculated, as shown in Figure 6 (also see Figures S9 and S10) and Table 7. Since the unnormalized SAR maps involve the SAR results without the consideration of the transmission RF power, more quantitative SAR maps were analyzed by calculating the normalized SAR maps by applying a 90° RF pulse. The mean and max SAR values are listed in Table 7. In the SAR maps of the BP-BC RF coil and the 16-ch RF coil configurations, it was not possible to confirm the rapid change in the SAR map distribution, as shown in Figure 6.
Table 7. Whole-average and max SAR values of normalized EM field simulation results.

|                      | BP-BC RF Coil | BP-BC RF Coil | 16-ch RF Coil | 16-ch RF Coil |
|----------------------|---------------|---------------|---------------|---------------|
|                      | wo-scHPM      | w-scHPM       | wo-scHPM      | w-scHPM       |
| Whole-averaged SAR   | 0.209 (w-WE)  | 0.215 (w-WE)  | 0.249 (w-WE)  | 0.238 (w-WE)  |
|                      | 0.216 (w-WE)  | 0.211 (w-WE)  | 0.243 (w-WE)  | 0.227 (w-WE)  |
| Max SAR values       | 8.203 (w-WE)  | 8.461 (w-WE)  | 8.496 (w-WE)  | 8.472 (w-WE)  |
|                      | 8.376 (w-WE)  | 8.248 (w-WE)  | 8.284 (w-WE)  | 8.552 (w-WE)  |

In the BP-BC RF coil configuration results, the mean SAR values were calculated as 0.209 W/kg (BP-BC RF coil – wo-scHPM – wo-MCWE), 0.215 W/kg (BP-BC RF coil – w-scHPM – wo-MCWE), 0.216 W/kg (BP-BC RF coil – w-scHPM – w-MCWE). The mean SAR values of the 16-ch RF coil configurations were calculated as 0.249 W/kg (16-ch RF coil – w-scHPM – w-MCWE), 0.238 W/kg (16-ch RF coil – w-scHPM – w-BCWE), 0.243 W/kg (16-ch RF coil – wo-scHPM – w-BCWE), and 0.227 W/kg (16-ch RF coil – wo-scHPM – w-BCWE). The max SAR values of the BP-BC RF coil configurations were calculated as 8.203 W/kg (BP-BC RF coil – wo-scHPM – w-MCWE), 8.461 W/kg (BP-BC RF coil – w-scHPM – wo-MCWE), 8.376 W/kg (BP-BC RF coil – w-scHPM – w-MCWE), and the max SAR values of the 16-ch RF coil configurations were calculated as 8.496 W/kg (16-ch RF coil – w-scHPM – w-BCWE), 8.472 W/kg (16-ch RF coil – w-scHPM – w-BCWE), 8.284 W/kg (16-ch RF coil – wo-scHPM – w-BCWE), and 8.552 W/kg (16-ch RF coil – wo-scHPM – w-BCWE).

Ironically, the results of the normalized SAR maps and SAR values showed no significant changes. Since the 16-ch RF coil with the scHPM and WE provided higher $|B_{1+}|$ field efficiency, the distribution of SAR maps was almost unchanged compared to those without the scHPM and the WE, as the RF coil could be excited by 90° using low RF power.

To summarize the unnormalized EM field simulation results, the unnormalized $|B_{1+}|$ field sensitivity improvement rate between the 16-ch RF coil – w-scHPM – w-MCWE (improved 422.237% in the oil-based cylindrical phantom and 430.866% in the human head model compared to the 16-ch RF coil alone) and the BP-BC RF coil – w-scHPM – w-MCWE (improved 179.352% in the oil-based cylindrical phantom and 132.823% in the human head model compared to the BP-BC RF coil alone) was 235.424% based on the oil-based cylindrical phantom result and 324.391% based on the human head model.

Similarly, the unnormalized $|B_{1-}|$ field sensitivity improvement rate between the 16-ch RF coil – w-scHPM – w-MCWE (improved 502.564% in the oil-based cylindrical phantom and 422.266% in the human head model compared to the 16-ch RF coil alone) and the BP-BC RF coil – w-scHPM – w-MCWE (improved 256.000% in oil-based cylindrical phantom and 127.675% in the human head model compared to the BP-BC RF coil alone) was 196.314% and 330.735% based on the oil-based cylindrical phantom and the human head model, respectively.

Moreover, the unnormalized $|E|$ field sensitivity improvement rate between the 16-ch RF coil – w-scHPM – w-MCWE (improved 502.564% in the oil-based cylindrical phantom and 422.266% in the human head model compared to the 16-ch RF coil alone) and the BP-BC RF coil – w-scHPM – w-MCWE (improved 256.000% in oil-based cylindrical phantom and 127.675% in the human head model compared to the BP-BC RF coil alone) was found to be 110.848% and 153.124%, respectively, using the oil-based cylindrical phantom and the human head model.

In the unnormalized EM field simulation results, the unnormalized $|B_{1+}|$ and $|B_{1-}|$ field sensitivity improvement rates of the 16-ch RF coil with the scHPM and the BCWE were substantial and showed surprisingly similar tendencies. In particular, the 16-ch RF
coil consisting of the same dimensions as the scHPM was used as a Tx/Rx RF coil to significantly improve the $|B^+_{1}|$ field sensitivity. In addition, it was confirmed that the sensitivity of the $|B^-_{1}|$ field was greatly improved by applying the BCWE, which provided a volume structure, compared to the MCWE.

Moreover, the 16-ch RF coil with the scHPM and the BCWE showed little change in the unnormalized $|E|$ field sensitivity improvement rate. The $|E|$ field sensitivity improvement rate was calculated under unnormalized conditions, but when applying the Norm-COEFS to an $|E|$ field distribution, the 16-ch RF coil with the scHPM and the BCWE could exhibit a lower $|E|$ field distribution. In addition, these results were also applied to the SAR maps, and the normalized SAR maps were also measured with a 16-ch RF coil without any difference.

In the normalized EM field simulation results, the normalized $|B^+_{1}|$ field in the 16-ch RF coil with the scHPM and the BCWE showed a high unnormalized $|B^+_{1}|$ field efficiency without degrading the uniformity of the normalized $|B^+_{1}|$ field distribution compared to the BP-BC RF coil with the scHPM and the MCWE. In particular, the results of the human head model confirmed improved $|B^+_{1}|$ field efficiency in a wider area from the 16-ch RF coil with the scHPM and the BCWE to the frontal lobe. As a result, it was confirmed that the 16-ch RF coil with the scHPM and the BCWE had better $|B^+_{1}|$ field efficiency and uniformity than the BP-BC RF coil with the scHPM and the MCWE and could significantly improve $|B^+_{1}|$ field efficiency without degrading $|B^+_{1}|$ field uniformity.

For the SAR map with the applied Norm-COEFS, the 16-ch RF coil combination with the BCWE and the scHPM provided RF safety in terms of sufficiently reduced SAR due to the relatively low RF power applied to the 16-ch RF coil. Under the normalized condition, both the $|B^+_{1}|$ field efficiency and $|B^-_{1}|$ field sensitivity of the 16-ch RF coil – scHPM – w-BCWE increased dramatically compared to the BP-BC RF coil – w-scHPM – w-MCWE, providing SAR distributions similar to those of the BP-BC RC coil or the 16-ch RF coil alone.

The unnormalized ($|B^+_{1}|$, $|B^-_{1}|$, and $|E|$ fields) and normalized EM field results (normalized $|B^+_{1}|$ field and SAR maps) of the 16-ch RF coil with the scHPM and the BCWE and the BP-BC RF coil with scHPM and the MCWE were compared in the $x$–$z$ and $y$–$z$ planes, as well as in the $x$–$y$ plane, and the results are provided in the Supplementary Materials (Figures S1–S10).

Here, we describe the results of the EM field simulations of the 16-ch RF coil with the scHPM and the BCWE, which was intended to improve $|B^+_{1}|$ field efficiency and $|B^-_{1}|$ field sensitivity. The scHPM structure has proved to be optimal and has clear limitations in further structure improvement [47]. The BP-BC RF coil proposed in our previous work has been widely used in MRI because it provides a highly uniform $B_0$ field distribution, but a decrease in RF wavelength after increasing the $B_0$ field strength reduces the $B_1$ field uniformity, which is a critical weakness for volume coils, such as BP-BC RF coils. In addition, the MCWE used as the WE was configured as a multichannel considering interference with the scHPM, but there were limitations in providing sufficient $|B^+_{1}|$ and $|B^-_{1}|$ field improvement capabilities.

To address these problems, we decided to change the role of the BP-BC RF coil and the MCWE. The 16-ch RF coil, which provides a more uniform $|B^+_{1}|$ and $|B^-_{1}|$ field distribution with a higher sensitivity than the BC RF coil in UHF MRI, was changed and applied to a Tx/Rx coil, and the BP-BC RF coil was used to improve the $|B^-_{1}|$ field sensitivity in the MCWE.

As a result, the 16-ch RF coil consisting of the same dimensions as the scHPM was able to dramatically improve the $|B^+_{1}|$ field efficiency, and the BCWE consisting of the BC RF coil was also able to significantly improve the $|B^-_{1}|$ field sensitivity. Thus, to improve the limited $B_1$ field sensitivity due to the reduced RF transceiver efficiency and sensitivity in UHF MRI, the proposed combination of RF coil configurations can be adopted using a BCWE for improving $|B^-_{1}|$ field sensitivity and an scHPM for enhancing $|B^+_{1}|$ field efficiency.
In this study, the $|B_1^+|$ field efficiency of the 16-ch RF coil with the scHPM was significantly improved compared to the BP-BC RF coil with the scHPM by configuring the scHPM structure to fit the size of the 16-ch RF coil and increasing the energy density without distorting the RF transmit direction. Furthermore, by replacing the MCWE with the BCWE, higher $|B_1^{-}|$ field efficiency was ensured.

The synergy between the BCWE and scHPM configurations allowed the $|B_1^+|$ and $|B_1^{-}|$ fields in the 16-ch RF coil to dramatically enhance the Tx/Rx efficiency and sensitivity compared to BP-BC RF coil – $w$-scHPM – $w$-MCWE. Improvements in Tx/Rx efficiency and sensitivity could consequently lead to $|B_1|$ field improvement, which can be seen in Figures S11–S13.

Our numerical EM field calculations also showed that the proposed combination of the 16-ch RF coil with the scHPM and the BCWE provided enhanced $|B_1^+|$ field efficiency and $|B_1^{-}|$ field sensitivity in the human head model. The SAR values of the proposed combination indicate its suitability for UHF MRI.

To discuss a single limitation in this study, we were unable to obtain MR images for the human head using the proposed 16-ch RF coil – $w$-scHPM – $w$-BCWE. The MRI system currently in possession is a 7.0 T MRI scanner (Siemens, Magnetom, Germany), which is currently operated jointly in hospitals and laboratories. For the 7.0 T MRI system currently in operation, it is essential to obtain an Institutional Review Board (IRB) for human application with research equipment permit, not clinical-only equipment.

Acquisition of an Instrumental Review Board (IRB) is essential for human imaging using 7.0 T MRI systems for clinical and research purposes. However, the 7.0 T MRI used for clinical and research purposes is strictly prohibited for obtaining human images with artifacts such as scHPM and BCWE inserted inside the main magnet bore. For this reason, it may be difficult to obtain human head images of the 16-ch RF coil – $w$-scHPM – $w$-BCWE using 7.0 T MRI systems for future human imaging, so we plan to implement the 16-ch RF coil – $w$-scHPM – $w$-BCWE using a pre-clinical MRI system.

4. Conclusions

The proposed new combination of the Tx/Rx RF coil configuration using an scHPM and a WE was tried to simultaneously improve $|B_1^+|$ field efficiency and $|B_1^{-}|$ field sensitivity in UHF MRI. In our previous study, we proposed an RF coil combination for the 7.0 T MRI system that provided improved $|B_1^+|$ field efficiency and $|B_1^{-}|$ field sensitivity over the BP-BC RF coil combination with the scHPM and the MCWE.

Since the scHPM structure has already been determined to be the optimal structure, a method for simultaneously improving $|B_1^+|$ field efficiency and $|B_1^{-}|$ field sensitivity by optimizing the structures of RF coils and the WE was developed. As a result, the propagation direction of the RF transmission power through the scHPM was further clarified using a 16-ch RF coil as a Tx/Rx coil, and a volumetric WE (BCWE) was proposed to maximize the efficiency of $|B_1^{-}|$ field sensitivity. Through these processes, the 16-ch RF coil with the scHPM and the BCWE was proposed as a new RF coil combination to simultaneously improve the $B_1^+$ field efficiency and the $B_1^-$ field sensitivity.

The EM field simulations were performed to verify the effects of the proposed configurations using numerical calculations. The configuration of the 16-ch RF coil combination with the scHPM and the BCWE was modified to switch the roles of the BP-BC RF coil and the MCWE. The 16-ch RF coil combination with the scHPM and the BCWE was compared to the BP-BC RF coil combination with the scHPM and the MCWE. The structures of the scHPM and the WE (BCWE and MCWE) were designed to match the propagation direction of the RF wave generated by the Tx/Rx RF coils (16-ch RF coil and BP-BC RF coil) and the positions of both the scHPM and the WE (BCWE and MCWE). Specifically, the design and dimensions of both the scHPM and the WE (BCWE and MCWE) were such that they could be located between the elements of the 16-ch RF coil and the legs of the BC coil.

The proposed 16-ch RF coil combination with the scHPM and the BCWE dramatically improved both the $|B_1^+|$ field efficiency and $|B_1^-|$ field sensitivity of the 16-ch RF coil in
UHF MRI. The $|B^+|$ field efficiency and $|B^-|$ field sensitivity of 16-ch RF coil combination with the scHPM and the BCWE provided significant performance improvements over the BP-BC RF coil combination with the scHPM and the MCWE proposed in the previous work. According to these results, the proposed RF coil configuration can improve the performance of 16-ch RF coils limited by low RF power and SAR issues in UHF MRI (above 7.0 T).

Nevertheless, the application of the 16-ch RF coil with the scHPM and the BCWE to the actual clinical 7.0 T MRI failed, and we could not perform it experimentally to verify with the EM field simulation results. MR image acquisition using unauthorized devices is strictly prohibited. Due to strict MRI safety regulations, actual experiments using the scHPM and the BCWE could not be performed. However, through the findings demonstrated, we confirmed the possibility of scHPM and BCWE use to simultaneously enhance $|B^+|$ field efficiency and $|B^-|$ field sensitivity.

In our future work, a study on the combination of the 16-ch RF coil with the scHPM and the MCWE will be conducted using numerical EM field simulations. To conduct a study on the optimized MCWE location, we will study the effects of the alignment and tilt angle of 16-ch RF coil, scHPM, and MCWE.

**Supplementary Materials:** The following supporting information can be downloaded at: https://www.mdpi.com/article/10.3390/s22228986/s1: Figure S1. Unnormalized $|B^+|$ field distribution (x–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S2. Unnormalized $|B^-|$ field distribution (y–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S3. Unnormalized $|B^+|$ field distribution (x–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S4. Unnormalized $|B^-|$ field distribution (y–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S5. Unnormalized $|E|$ field distribution (x–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S6. Unnormalized $|E|$ field distribution (y–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S7. Normalized $|B^+|$ field distribution (x–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE. Figure S8. Normalized $|B^-|$ field distribution (y–z plane) in the oil-based cylindrical phantom (a–h) and human head model (i–p): (a,i) BP-BC RF coil – scHPM – MCWE; (b,j) BP-BC RF coil – scHPM – MCWE; (c,k) 16-ch RF coil – scHPM – BCWE; (d,j) 16-ch RF coil – scHPM – BCWE; (e,m) BP-BC RF coil – scHPM – MCWE; (f,l) BP-BC RF coil – scHPM – MCWE; (g,o) 16-ch RF coil – scHPM – BCWE; (h,p) 16-ch RF coil – scHPM – BCWE.
Sensors 2022, 22, 8968

3. Frayne, R.; Goodyear, B.G.; Dickhoff, P.; Lauzon, M.L.; Sevick, R.J. Magnetic resonance imaging at 3.0 Tesla: Challenges and
6. Kerchner, G.A. Ultra-high field 7T MRI: A new tool for studying Alzheimer’s disease. J. Alzheimer’s Dis. 2011, 26, 91–95. [CrossRef]

References
1. Thulborn, K.R. Clinical rationale for very-high field (3.0 Tesla) functional magnetic resonance imaging. Top. Magn. Reson. Imaging 1999, 10, 37–50. [CrossRef] [PubMed]
2. Yacoub, E.; Shmuel, A.; Pfeuffer, J.; Van De Moortele, P.F.; Adriany, G.; Andersen, P.; Vaughan, J.T.; Merkle, H.; Ugurbil, K.; Hu, X. Imaging brain function in humans at 7 tesla. Magn. Reson. Med. 2001, 45, 588–594. [CrossRef] [PubMed]
3. Frayne, R.; Goodyear, B.G.; Dickhoff, P.; Lauzon, M.L.; Sevick, R.J. Magnetic resonance imaging at 3.0 Tesla: Challenges and advantages in clinical neurological imaging. Investig. Radiol. 2003, 38, 385–402. [CrossRef] [PubMed]
4. Lu, H.; Nagae-Poetscher, L.M.; Golay, X.; Lin, D.; Pomper, M.; van Zijl, P.C.M. Routine clinical brain MRI sequences for use at 3.0 Tesla. J. Magn. Reson. Imaging 2005, 22, 13–22. [CrossRef] [PubMed]
5. FDA. Clears First 7T Magnetic Resonance Imaging Device. 2017. Available online: https://www.fda.gov/NewsEvents/Newsroom/PressAnnouncements/ucm580154.htm (accessed on 27 June 2019).

Author Contributions: J.-H.S. and Y.-S.J. designed the EM field simulation models for RF coil configurations (16-ch RF coil, sCHPM, and BCWE) and analyzed EM field simulation models using FDTD methods. J.-H.S. and Y.-S.J. conducted the simulation calculations. J.-H.S. and Y.-S.J. performed the writing and formatting of this manuscript and contributed equally as the first co-authors. C.-H.O. contributed equally as the corresponding authors. All authors have read and agreed to the published version of the manuscript.

Funding: This work was supported by a National Research Foundation of Korea (NRF) grant funded by the Korean government (MSIT) (No. NRF-2022R1A2C2010363) and a grant of the Korea Health Technology R&D Project through the Korea Health Industry Development Institute (KHIDI), funded by the Ministry of Health & Welfare, Republic of Korea (grant number: HR14C0002).

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: Not applicable.

Conflicts of Interest: The authors declare no conflict of interest.

[CrossRef] [PubMed] [CrossRef] [PubMed] [CrossRef] [PubMed] [CrossRef] [PubMed] [CrossRef] [PubMed]
32. Hong, S.-E.; Oh, S.; Choi, H.-D. RF exposure assessment for various poses of patient assistant in open MRI environment. *Appl. Sci.* 2021, 11, 4967. [CrossRef]
33. Bottomley, P.A.; Andrew, E.R. RF magnetic field penetration, phase shift and power dissipation in biological tissue: Implications for NMR imaging. *Phys. Med. Biol.* 1978, 23, 630–643. [CrossRef]
34. Bachli, R.; Saner, M.; Meier, D.; Boskamp, E.B.; Boesiger, P. Increased RF power absorption in MR imaging due to RF coupling between body coil and surface coil. *Magn. Reson. Med.* 1989, 9, 105–112. [CrossRef]
35. Grandolfo, M.; Vecchia, P.; Gandhi, O.P. Magnetic resonance imaging: Calculation of rates of energy absorption by a human-torso model. *Bioelectromagnetics* 1990, 11, 117–128. [CrossRef]
36. Simunic, D.; Wach, P.; Renhart, W.; Stollberger, R. Spatial distribution of high-frequency electromagnetic energy in human head during MRI: Numerical results and measurements. *IEEE Trans. Biomed. Eng.* 1996, 43, 88–94. [CrossRef]
37. Jin, J.; Chen, J. On the SAR and field inhomogeneity of birdcage coils loaded with the human head. *Magn. Reson. Med.* 1997, 38, 953–963. [CrossRef]
38. Qian, D.; El-Sharkawy, A.-M.M.; Bottomley, P.A.; Edelstein, W.A. An RF dosimeter for independent SAR measurement in MRI scanners. *Med. Phys.* 2013, 40, 122303. [CrossRef] [PubMed]
39. Yetisir, F.; Turk, E.A.; Guerin, B.; Gagoski, B.A.; Grant, P.E.; Adalsteinsson, E.; Wald, L.L. Safety and imaging performance of two-channel RF shimming for fetal MRI at 3T. *Magn. Reson. Med.* 2021, 86, 2810–2821. [CrossRef]
40. Noetscher, G.M.; Serano, P.; Wartman, W.A.; Fujimoto, K.; Makarov, S.N. Visible Human Project® female surface based computational phantom (Nelly) for radio-frequency safety evaluation in MRI coils. *PloS ONE* 2016, 11, e026922. [CrossRef]
41. Seo, J.-H.; Ryu, Y.; Chung, J.-Y. Simulation study of radio frequency safety and the optimal size of a single-channel surface radio frequency coil for mice at 9.4 T magnetic resonance imaging. *Sensors* 2022, 22, 4274. [CrossRef]
42. Tarasek, M.R.; Shu, Y.; Kang, D.; Tao, S.; Gray, E.; Huston, J.; Hua, Y.; Yeo, D.T.B.; Bernstein, M.A.; Foo, T.K. Average SAR prediction, validation, and evaluation for a compact MR scanner head-sized RF coil. *Magn. Reson. Imaging* 2022, 85, 168–176. [CrossRef]
43. Pang, Y.; Xie, Z.; Li, Y.; Xu, D.; Vigneron, D.; Zhang, X. Resonant mode reduction in radiofrequency volume coils for ultra-high field magnetic resonance imaging. *Sensors* 2020, 20, 122303. [CrossRef] [PubMed]
44. Santarelli, M.F.; Giovannetti, G.; Hartwig, V.; Celi, S.; Positano, V.; Landini, L. The core of medical imaging: State of the art and perspectives on the detectors. *Magn. Reson. Imaging* 2022, 85, 1642. [CrossRef]
45. Yoon, J.-S.; Kim, J.-M.; Chung, H.-J.; Jeong, Y.-J.; Jeong, G.-W.; Park, I.; Kim, G.-W.; Oh, C.-H. Development of a proton-frequency-transparent birdcage radiofrequency coil for in vivo 13C MRS/MRSI study in a 3.0 T MRI system. *Appl. Sci.* 2021, 11, 11445. [CrossRef]
46. Seo, J.-H.; Chung, J.-Y. A preliminary study for reference RF coil at 11.7 T MRI: Based on electromagnetic field simulation of hybrid-BC RF coil according to diameter and length at 3.0, 7.0 and 11.7 T. *Sensors* 2022, 22, 1512. [CrossRef] [PubMed]
47. Seo, J.-H.; Han, Y.; Chung, J.-Y. A comparative study of birdcage RF coil configurations for ultra-high field magnetic resonance imaging. *Sensors* 2022, 22, 1741. [CrossRef]
48. Bottomley, P.A.; Redington, R.W.; Edelstein, W.A.; Schenck, J.F. Estimating radiofrequency power deposition in body NMR imaging. *Magn. Reson. Med.* 1985, 2, 336–349. [CrossRef]
49. Chen, C.N.; Sank, V.J.; Cohen, S.M.; Houlit, D.I. The field dependence of NMR imaging. I. Laboratory assessment of signal-to-noise ratio and power deposition. *Magn. Reson. Med.* 1986, 3, 722–729. [CrossRef]
50. Robitaille, P.M. On RF power and dielectric resonances in UHF MRI. *NMR Biomed.* 1999, 12, 318–319. [CrossRef]
51. Houlit, D.I.; Phil, D. Sensitivity and power deposition in a high field imaging experiment. *J. Magn. Reson. Imaging* 2000, 12, 46–67. [CrossRef]
52. Barberi, E.A.; Gati, J.S.; Rutt, B.K.; Menon, R.S. A transmit-only/receive-only (TORO) RF system for high field MRI/MRS applications. *Magn. Reson. Med.* 2000, 43, 284–289. [CrossRef]
53. Collins, C.M.; Smith, M.B. Signal-to-noise ratio and absorbed power as functions of main magnetic field strength, and definition of “90 degrees” RF pulse for the head in the birdcage coil. *Magn. Reson. Med.* 2001, 45, 684–691. [CrossRef]
54. Yang, Q.X.; Wang, J.; Zhang, X.; Collins, C.M.; Smith, M.B.; Liu, H.; Zhu, X.-H.; Vaughan, J.T.; Ugurbil, K.; Chen, W. Analysis of wave behavior in lossy dielectric samples at high field. *Prog. Electromagn. Res. Lett.* 2020, 93, 143–151. [CrossRef]
55. Ugurbil, K. Magnetic resonance imaging at ultrahigh fields. *IEEE Trans. Biomed. Eng.* 2014, 61, 1364–1379. [CrossRef]
56. Malik, S.J.; Hand, J.W.; Sattarine, R.; Price, A.N.; Hajnal, J.V. Specific absorption rate and temperature in neonate models resulting from exposure to a 7T head coil. *Magn. Reson. Med.* 2021, 86, 1299–1313. [CrossRef]
57. Malik, S.J.; Hand, J.W.; Carmichael, D.W.; Hajnal, J.V. Evaluation of specific absorption rate and heating in children exposed to a 7T MRI head coil. *Magn. Reson. Med.* 2022, 88, 1434–1449. [CrossRef]
58. Edelstein, W.A.; Glover, G.H.; Hardy, C.J.; Redington, R.W. The intrinsic signal-to-noise ratio in NMR imaging. *Magn. Reson. Med.* 1986, 3, 604–618. [CrossRef]
59. De Zwart, J.A.; Ledden, P.J.; Gelderen, P.V.; Bodurka, J.; Chu, R.; Duyn, J.H. Signal-to-noise ratio and parallel imaging performance of a 16-channel receive-only brain coil array at 3.0 Tesla. *Magn. Reson. Med.* 2004, 51, 22–26. [CrossRef]
61. Kim, K.-N.; Seo, J.-H.; Han, S.-D.; Heo, P.; Im, G.H.; Lee, J.H. Development of double-layer coupled coil for improving S/N in 7 T small-animal MRI. *Scanning* **2015**, *37*, 361–371. [CrossRef]

62. Kim, K.-N.; Ryu, Y.; Seo, J.-H.; Kim, Y.-B. Magnetic field sensitivity at 7-T using dual-helmholtz transmit-only coil and 12-channel receive-only bended coil. *Scanning* **2016**, *38*, 515–524. [CrossRef]

63. Kim, K.-N.; Hernandez, D.; Seo, J.-H.; Noh, Y.; Han, Y.; Ryu, Y.C.; Chung, J.-Y. Quantitative assessment of phased array coils with different numbers of receiving channels in terms of signal-to-noise ratio and spatial noise variation in magnetic resonance imaging. *PLoS ONE* **2019**, *14*, e0219407. [CrossRef] [PubMed]

64. Giovannetti, G.; Fiori, A.; Martini, N.; Franceschello, R.; Aquaro, G.D.; Pingitore, A.; Frijia, F. Sodium radiofrequency coils for magnetic resonance: From design to applications. *Electronics* **2021**, *10*, 1788. [CrossRef]

65. Kell, R.C.; Greenham, A.C.; Olds, G.C.E. High-permittivity temperature-stable ceramic dielectrics with low microwave loss. *Am. Ceram. Soc.* **1973**, *56*, 352–354. [CrossRef]

66. Dang, Z.-M.; Yuan, J.-K.; Yao, S.-H.; Liao, R.-J. Flexible nanodielectric materials with high permittivity for power energy storage. *Adv. Mater.* **2013**, *25*, 6334–6365. [CrossRef] [PubMed]

67. Lee, B.-Y.; Zhu, X.-H.; Rupprecht, S.; Lanagan, M.T.; Yang, Q.X.; Chen, W. Large improvement of RF transmission efficiency and reception sensitivity for human in vivo 31P MRS imaging using ultrahigh dielectric constant materials at 7 T. *Magn. Reson. Imaging* **2017**, *42*, 158–163. [CrossRef]

68. Zivkovic, I.; Teeuwisse, W.; Slobozhanyuk, A.; Nenasheva, E.; Webb, A. High permittivity ceramics improve the transmit field and receive efficiency of a commercial extremity coil at 1.5 Tesla. *J. Magn. Reson.* **2019**, *299*, 59–65. [CrossRef]

69. Vorobyev, V.; Shchelokova, A.; Zivkovic, I.; Slobozhanyuk, A.; Baena, J.D.; Del Risco, J.P.; Abdeldaim, R.; Webb, A.; Glybovski, S. An artificial dielectric slab for ultrahigh field MRI: Proof of concept. *J. Magn. Reson.* **2020**, *320*, 106835. [CrossRef]

70. Hoult, D.I.; Tomanek, B. Use of mutually inductive coupling in probe design. *Concepts Magn. Reson. Part B Magn. Reson. Eng.* **2002**, *15*, 262–285. [CrossRef]

71. Wang, T.; Ciobanu, L.; Zhang, X.; Webb, A. Inductively coupled RF coil design for simultaneous microimaging of multiple samples. *Concepts Magn. Reson. B* **2008**, *33B*, 236–243. [CrossRef]

72. Bulumulla, S.B.; Fiveland, E.; Park, K.J.; Foo, T.K.; Hardy, C.J. Inductively coupled wireless RF coil arrays. *Magn. Reson. Imaging* **2015**, *33*, 351–357. [CrossRef]

73. Mett, R.R.; Sidabras, J.W.; Hyde, J.S. MRI surface-coil pair with strong inductive coupling. *Rev. Sci. Instrum.* **2016**, *87*, 124704. [CrossRef]

74. Byun, J.-D.; Seo, J.-H.; Kang, T.; Ryu, Y.; Kim, K.-N. Birdcage coil with inductively coupled RF coil array for improving |B1| field sensitivity in 7-T MRI. *J. Magn. Reson.* **2017**, *227*, 378–381. [CrossRef]

75. Seo, J.-H.; Lee, J.J.; Kim, K.-N. Surface coil with an inductively coupled wireless surface and volume coil for improving the magnetic field sensitivity at 400-MHz MRI. *J. Magn. Reson.* **2018**, *23*, 192–195. [CrossRef]

76. Mahmood, M.F.; Gharhan, S.K.; Mohammed, S.L.; Al-Naji, A.; Chahl, J. Design of powering wireless medical sensor based on spiral-spider coils. *Designs* **2021**, *5*, 59. [CrossRef]

77. Teeuwisse, W.M.; Brink, W.M.; Haines, K.N.; Webb, A.G. Simulations of high permittivity materials for 7 T neuroimaging and evaluation of a new barium titanate-based dielectric. *Magn. Reson. Med.* **2012**, *67*, 912–918. [CrossRef]

78. Seo, J.-H.; Ryu, Y.; Han, S.-D.; Song, H.; Kim, H.-K.; Kim, K.-N. Influence of biological subject, shielding cage, and resonance frequency on radio wave propagation in a birdcage coil. *Electron. Lett.* **2016**, *52*, 801–803. [CrossRef]

79. Ahmad, S.F.; Kim, Y.C.; Choi, I.C.; Kim, H.D. Recent progress in birdcage RF coil technology for MRI system. *Diagnostics* **2020**, *10*, 1017. [CrossRef]

80. Kim, Y.C.; Kim, H.D.; Yun, B.-J.; Ahmad, S.F. A simple analytical solution for the designing of the birdcage RF coil used in NMR imaging applications. *Appl. Sci.* **2020**, *10*, 2242. [CrossRef]

81. Yee, K.S. Numerical solution of initial boundary value problems involving Maxwell’s equations in isotropic media. *IEEE Trans. Antennas Propag.* **1996**, *14*, 302–307.

82. Haemer, G.G.; Vaidya, M.; Collins, C.M.; Sodickson, D.K.; Wiggins, G.C.; Lattanzi, R. Approaching ultimate intrinsic specific absorption rate in radiofrequency shimming using high-permittivity materials at 7 Tesla. *Magn. Reson. Med.* **2018**, *80*, 391–399. [CrossRef]

83. Liu, W.; Kao, C.-P.; Collins, C.M.; Smith, M.B.; Yang, Q.X. On consideration of radiated power in RF field simulations for MRI. *Magn. Reson. Med.* **2013**, *69*, 290–294. [CrossRef] [PubMed]

84. Yang, Q.X.; Rupprecht, S.; Luo, W.; Sica, C.; Herze, Z.; Wang, J.; Cao, Z.; Vesek, J.; Lanagan, M.T.; Carlucco, G.; et al. Radiofrequency field enhancement with high dielectric constant (HDC) pads in a receive array coil at 3.0 T. *J. Magn. Reson. Imaging* **2013**, *38*, 435–440. [CrossRef] [PubMed]

85. Cao, Z.; Park, J.; Cho, Z.-H.; Collins, C.M. Numerical evaluation of image homogeneity, signal-to-noise ratio, and specific absorption rate for human brain imaging at 1.5, 3, 7, 10.5, and 14 T in an 8-channel transmit/receive array. *J. Magn. Reson. Imaging* **2015**, *41*, 1432–1439. [CrossRef]

86. Alon, L.; Deniz, C.M.; Carlucco, G.; Brown, R.; Sodickson, D.K.; Collins, C.M. Effects of anatomical differences on electromagnetic fields, SAR, and temperature change. *Concepts Magn. Reson. Part B Magn. Reson. Eng.* **2016**, *46*, 8–18. [CrossRef]

87. Vaidya, M.V.; Collins, C.M.; Sodickson, D.K.; Brown, R.; Wiggins, G.C.; Lattanzi, R. Dependence of $B_1^+$ and $B_1^-$ field patterns of surface coils on the electrical properties of the sample and the MR operating frequency. *Concepts Magn. Reson. Part B Magn. Reson. Eng.* **2016**, *46*, 25–40. [CrossRef] [PubMed]
88. Alon, L.; Lattanzi, R.; Lakshmanan, K.; Brown, R.; Deniz, C.M.; Sodickson, D.K.; Collins, C.M. Transverse slot antennas for high field MRI. *Magn. Reson. Med.* **2018**, *80*, 1233–1242. [CrossRef]

89. Vaidya, M.V.; Deniz, C.M.; Collins, C.M.; Sodickson, D.K.; Lattanzi, R. Manipulating transmit and receive sensitivities of radiofrequency surface coils using shielded and unshielded high-permittivity materials. *Magn. Reson. Mater. Phys. Biol. Med.* **2018**, *31*, 355–366. [CrossRef]

90. Vaidya, M.V.; Lazar, M.; Deniz, C.M.; Haemer, G.G.; Chen, G.; Bruno, M.; Sodickson, D.K.; Lattanzi, R.; Collins, C.M. Improved detection of fMRI activation in the cerebellum at 7 T with dielectric pads extending the imaging region of a commercial head coil. *J. Magn. Reson. Imaging* **2018**, *48*, 431–440. [CrossRef]

91. Hoult, D.I. The principle of reciprocity in signal strength calculations—A mathematical guide. *Concepts Magn. Reson.* **2000**, *12*, 173–187. [CrossRef]

92. Im, G.H.; Seo, J.-H.; Kim, K.-N.; Heo, P.; Chung, C.C.; Jang, M.S.; Lee, J.H.; Kim, S.I. Effective arrangement of separated transmit-only/receive-only RF coil for improvement of $B_1$ homogeneity at 7 Tesla. *J. Korean Phys. Soc.* **2014**, *65*, 616–624. [CrossRef]

93. Tang, L.; Hue, Y.K.; Ibrahim, T.S. Studies of RF shimming techniques with minimization of RF power deposition and their associated temperature changes. *Concepts Magn. Reson. Part B Magn. Reson. Eng.* **2011**, *39B*, 11–25. [CrossRef]

94. Herrmann, T.; Liebig, T.; Mallow, J.; Bruns, C.; Stadler, J.; Mylius, J.; Brosch, M.; Svedja, J.T.; Chen, Z.; Rennings, A.; et al. Metamaterial-based transmit and receive system for whole-body magnetic resonance imaging at ultra-high magnetic fields. *PLoS ONE* **2018**, *13*, e0191719. [CrossRef]