Folded-end dipole transceiver array for human whole-brain imaging at 7 T

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The advancement of clinical applications of ultrahigh field (UHF) MRI depends heavily on advances in technology, including the development of new radiofrequency (RF) coil designs. Currently, the number of commercially available 7 T head RF coils is rather limited, implying a need to develop novel RF head coil designs that offer superior transmit and receive performance. RF coils to be used for clinical applications must be robust and reliable. In particular, for transmit arrays, if a transmit channel fails the local specific absorption rate may increase, significantly increasing local tissue heating. Recently, dipole antennas have been proposed and used to design UHF head transmit and receive arrays. The dipole provides a unique simplicity while offering comparable transmit efficiency and signal-to-noise ratio with the conventional loop design. Recently, we developed a novel array design in our laboratory using a folded-end dipole antenna. In this work, we developed, constructed and evaluated an eight-element transceiver bent folded-end dipole array for human head imaging at 7 T. Driven in the quadrature circularly polarized mode, the array demonstrated more than 20% higher transmit efficiency and significantly better whole-brain coverage than that provided by a widely used commercial array. In addition, we evaluated passive dipole antennas for decoupling the proposed array. We demonstrated that in contrast to the common unfolded dipole array, the passive dipoles moved away from the sample not only minimize coupling between the adjacent folded-end active dipoles but also produce practically no destructive interference with the quadrature mode of the array.

**KEYWORDS**
array optimization, decoupling, folded-end dipole, human head imaging, transceiver array, ultrahigh field MRI
1 | INTRODUCTION

MRI at ultrahigh field (UHF, ≥ 7 T) offers undisputable benefits of superior signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) over imaging at field strengths more commonly used in a clinical setting (3 T and below). The US Food and Drug Administration (FDA) has recently cleared two commercially available 7 T MRI scanners for diagnostic imaging, and the European Union has also certified one of those machines for clinical use. The advancement of clinical applications of UHF MRI depends heavily on advances in technology, including the development of new RF coil designs. Currently, the number of commercially available 7 T head RF coils is rather limited, implying a need to develop novel RF head coil designs that offer superior transmit (Tx) and receive (Rx) performance. New coils to be used for clinical applications must be very robust and reliable, as one of the major problems in UHF imaging is safety due to a significant increase of local tissue heating. In particular, for transmit arrays, if a transmit channel fails then the local tissue heating may significantly increase. The latter is often assessed by calculating the peak of local specific absorption rate (SAR). At UHF, the pSAR can be very high due to significant shortening of the wavelength (below 15 cm at 300 MHz).

Due to a lack of a body Tx coil, UHF RF coils usually require a local Tx coil. Further, to mitigate problems caused by the short RF wavelength at UHF, Tx arrays are widely used, which makes the design of the RF coil quite complex. The most common UHF head coil designs use the ToRo-arrangement, consisting of an outer Transmit-only coil (array) and an inner Receive-only array. State-of-the-art UHF RF head coils include a larger 8- or 16-element Tx array and a tight-fit 32- and 64-element Rx array. The development of novel, simpler and more robust designs that simultaneously offer comparable or even improved Tx and Rx performance is therefore in great demand.

Most 7 T Tx and Rx array designs use surface loop elements. Recently, dipole antennas have been proposed and used to design UHF head Tx\textsuperscript{5,6,8–13} and Rx\textsuperscript{14,15} arrays. The dipole provides a unique simplicity while offering comparable Tx efficiency and SNR with the conventional loop design. In addition, recent developments in ultimate intrinsic SNR (UISNR) theory suggest that dipoles and loops must be combined to achieve the maximum possible SNR in relatively large samples (e.g. the human body, head) at UHF.\textsuperscript{16–19} A UHF array that consists only of loops does not offer a maximum possible SNR. Thus, further optimization of the TxRx dipole antenna design is important. For imaging of a human head, the physical size of dipole antennas has to be reduced considerably from the naturally resonant half-wavelength, \(\lambda/2\) (50 cm in free space at 300 MHz). Reducing the dipole antenna size leads to lower loading, which results in a narrower resonance line width and suboptimal Tx and Rx performance. Various methods have been proposed for reducing the physical size of dipoles while maintaining their electrical length.\textsuperscript{8,10,11,20}

Recently, we developed a new type of dipole antenna in our laboratory, a folded-end dipole,\textsuperscript{15,21–23} which we used to design and construct transceiver (TxRx)\textsuperscript{21–23} and Rx\textsuperscript{15} arrays for imaging the human head at 9.4 T. In this design, both ends of the dipole, which carry higher voltage and lower current relative to the dipole center, are bent away from the sample and folded. This reduces the voltage along the “loaded” portion of the folded-end dipole and makes the current distribution along it more uniform. By the “loaded” portion of the dipole we imply the part of the folded-end antenna that is located near the sample and produces most of the RF field inside it. As a result of such alteration of the antenna design, the distribution of the Tx field, \(B_1^+\), is extended in the longitudinal direction (along the magnetic axis) and local SAR on the surface of the human head (ears, nose) is reduced.\textsuperscript{23} The Tx efficiency, evaluated as \(B_1^+ / \sqrt{\text{pSAR}}\), where pSAR is peak local SAR, is also improved.\textsuperscript{23} In addition, the shift of the resonance frequency due to variation in the head size\textsuperscript{15,24} is reduced.\textsuperscript{15} The latter is important for tight-fit arrays including TxRx and Rx arrays.

Another important aspect when designing Tx (or TxRx) head arrays is decoupling of adjacent dipoles. Due to the short length (<\(\lambda/2\)) of the dipoles and consequent lower loading, adjacent dipoles are usually coupled to the level of −10 dB or stronger.\textsuperscript{8,11,12,20} This degree of coupling may be excessive for optimal Tx performance. A recently proposed decoupling technique using passive dipoles placed symmetrically between adjacent active Tx dipoles is appealing because no galvanic connection is required.\textsuperscript{12,23,25–27} At the same time, as we have shown previously,\textsuperscript{23} passive dipoles of similar size to active dipoles can interfere destructively with the RF field of the Tx array. Therefore, the design of passive decoupling dipoles always needs to be thoroughly assessed.

In this work, we used folded-end dipole elements to construct an 8-element TxRx human head array for use at 7 T. To improve Tx efficiency and longitudinal coverage, we optimized the shape and geometry of dipole elements. We assessed the array performance both numerically and experimentally, including evaluation of two passive dipole designs for decoupling of adjacent dipoles. Finally, we compared the Tx performance of the array with that of the commercial RF array coil (8-channel Tx/32-channel Rx array, Nova Medical).

2 | METHODS

2.1 | Electromagnetic simulations

Before constructing the array, we evaluated and optimized the new array design using numerical electromagnetic (EM) simulations. These were performed using CST Studio Suite 2019 (Dassault Systèmes, Vélizy-Villacoublay, France) and the time-domain solver based on the finite-integration technique. We loaded the simulated arrays with two multitissue voxel models (“Duke” and “Elia”) from the Virtual Family,\textsuperscript{28} cropped at the chest level, with an isotropic resolution of 2 mm.
Following our previous works on designing the 9.4 T folded-end dipole array,\textsuperscript{21–23} we positioned eight dipole elements equidistantly around the head. All dipoles were modeled using annealed copper wire of 1.5-mm diameter. To accommodate the increase in the wavelength at 7 T, we increased the length of the dipoles. All folded-end dipoles were 250 mm in the z-direction (along the axis of the magnet). The total length of wire was 460 mm, which is close to $\lambda/2$ in the free space (500 mm at 300 MHz). The array was placed on a plastic holder with dimensions similar to those in the design of 9.4 T dipole arrays.\textsuperscript{21–23} The 3-mm-thick holder was 200 mm in width (left–right) and 230 mm in height (anterior–posterior). To better fit a human head, the holder was tapered at its superior location, where it was 155 mm in width and 185 mm in height. No simulated arrays had an RF shield. Figure 1A shows the geometry of the folded-end dipole array. The holder is hidden for better visualization. As seen in Figure 1A, each folded-end dipole is bent at the top to better fit the head geometry. This antenna design is referred to throughout the text as a bent folded-end dipole.

A large copper cylinder (640 mm diameter, 1600 mm long), to mimic the RF shield of the gradient coil, was included in all simulations. All matching and tuning circuits were calculated during postprocessing. Distributions (maps) of the Tx $B_1^+$ field and local SAR$_{10g}$ (averaged over 10 g of tissue) were calculated for 1 W input power at the coil input. $B_1^+$ and SAR$_{10g}$ maps were simulated for the quadrature circularly polarized (CP) mode of the array, which corresponds to a 45° phase shift between adjacent dipole antennas. We evaluated the Tx performance as $<B_1^+>/\sqrt{P}$, where $P$ is an RF power at the coil input, and $<B_1^+>/\sqrt{pSAR_{10g}}$ is the safety excitation efficiency (SEE).\textsuperscript{27} The $<B_1^+>$ value was averaged over a 130-mm transverse slab that includes the majority of the human brain. Homogeneity was also evaluated as the standard deviation (SD) of the $B_1^+$ value over the same volume.

During optimization of the array Tx performance, we considered the following alterations of the dipole array design. First, we moved the array relative to the head along the z-axis extending the edge of the array above the head, as shown in Figure 1A. As demonstrated previously,\textsuperscript{30,31} such an extension can enhance the RF field in the superior area of the head and, thus, improves the longitudinal coverage. We tested three positions of the array shown in Figure 1A, that is, the array aligned with head and moved by 20 and 40 mm out (these will be referred to as the “0-mm”, “20-mm” and “40-mm” positions). To further improve the $B_1^+$ field at the superior head area,\textsuperscript{21–23,32} we added a flat local elliptical RF shield (175 mm x 140 mm) placed near the head (Figure 1A). We also checked whether moving the local shield closer to the head has an effect on the RF field distribution (the last two pictures in Figure 1A). In addition, we evaluated the performance of two larger bent folded-end dipole arrays obtained by moving each dipole by 10 and 20 mm away from the head. Both array designs were extended by 40 mm above the head (40-mm position) and included the local superior RF shield. Also, for a more general comparison, we calculated pSAR$_{10g}$, $B_1^+$ efficiency, and SEE for 500 random shims (sets of eight driving amplitudes and phases) for two bent folded-end arrays (i.e. the 40-mm array and the same array with all dipoles moved by 20 mm away from the head). In both evaluations we used the Duke voxel model and performed averaging over 130-cm slab (Figure 1B).

**FIGURE 1** Electromagnetic (EM) simulation models (A) and corresponding $B_1^+$ maps (B) obtained using various altered versions of the bent folded-end dipole array all loaded by the Duke voxel model. Position “0-mm” corresponds to the edge of the array aligned with head. Positions “20-mm” and “40-mm” correspond to the array moved by 20 and 40 mm in the superior direction, respectively. The local superior shield is moved by 20 mm in the last picture of Figure 1. The 130-mm averaging transverse slab is shown in Figure 1B by a dashed line.
Finally, we compared the Tx performance of four different dipole arrays (Figure 2A). These include the bent folded-end dipole array, bent (unfolded) dipole array, straight folded-end dipole array, and straight (unfolded) dipole array. All four array designs were extended by 40 mm above the head and included the local superior shield (Figure 2A).

In the next step, we added passive decoupling dipoles to the TxRx-array design. Figure 3 shows placement of passive dipoles relative to active TxRx dipoles. We simulated two types of passive dipoles (Figure 3), that is, straight parallel passive dipoles located at the same distance from the phantom as the active dipoles (decoupling 1), and straight parallel passive dipoles moved farther away from the phantom to the level of the folded portion (decoupling 2). Decoupling 2 (i.e. the straight passive dipole moved away from the sample) was suggested previously as a method to optimize decoupling and minimize destructive interference. All passive dipoles (decoupling methods 1 and 2) were 250 mm long. In addition, we tested both decoupling methods for an array of bent (unfolded) dipoles (Figure 3, last column). All passive and active dipoles were modeled using 1.5-mm annealed copper wire. Because the presence of the passive dipole near the nose caused very strong local SAR in this area (described further below), it was removed, and all simulated decoupled dipole array designs (Figure 3) included only seven passive dipoles.

### 2.2 Array construction

After EM modeling, we constructed an 8-element TxRx dipole array. The geometry of the array holder and the dipole antennas was the same as in EM modeling. Figure 4A–C show the EM model and photographs of the final array design. The cover was removed in Figure 4B for better visualization of the components. Figure 4C shows the inner surface of the cover where seven straight passive dipoles were attached. The TxRx active dipoles were 35 mm in height and 250 mm in length in the z-direction. The total length of the dipole antennas were 460 mm. The passive dipoles were 250 mm in length. All dipoles were constructed using 1.5-mm tinned copper wire. Matching of the dipoles was realized using a π-circuit consisting of two inductors and a variable capacitor (Johanson Corp., Boonton, NJ, USA). Based on EM simulation data, we used variable capacitors of 20 pF. Values of tuning inductors were also estimated from simulations and measured from 10 to 20 nH. Such small values of the inductors indicate that the total dipole length was close to $\lambda/2$. To bring the resonance frequency of the passive dipoles down closer to 297 MHz, inductors (~150 nH) were connected in series with each passive dipole (Figure 4C). During construction, these inductors were adjusted to

**FIGURE 2** Electromagnetic (EM) simulation models (A) and corresponding $B_1^+$ maps (B) obtained using four different dipole arrays (bent folded-end dipole array, bent dipole array, straight folded-end dipole array, and straight dipole array), all loaded by the Duke voxel model. All four array designs were extended by 40 mm (position “40-mm”) above the head and included the local superior shield.
minimize coupling between adjacent active dipoles. All inductors were handmade using 1.2-mm tinned copper wire. A shielded cable trap was introduced at the input of each TxRx dipole element to cancel the common mode excited on the outer surface of the cable braid. To connect the coil to an MRI scanner, we constructed an 8-channel interface, which included a homebuilt 8-way splitter and eight T/R switchboards with integrated preamplifiers (WMA series, WanTcom, Chanhassen, MN, USA). The interface was placed in a separate box outside of the array holder. The array was connected to the interface box using an 8-channel multimodular coaxial connector (ODU, Mueldorf, Germany). As we did not plan on using fast pulse sequences, the local superior shield was not slotted and was constructed using a kapton laminate (AKAFLEX, Krempel GmbH, Germany) with relatively thin 25-μm copper foil. During transmission, the array was driven in the CP mode, corresponding to a 45° phase shift between adjacent dipoles.

### 2.3 Experimental evaluation

Before in vivo measurements, the dipole array was evaluated on the bench, in the scanner and in simulation, in accordance with the safety procedure developed in our laboratory. 

Bench evaluation of the array included measurements of the entire 8 x 8 S-matrix using a network analyzer (E5071C, Agilent Technology, Santa Clara, CA, USA). The final version of the array coil was tuned and matched on a human head. No retuning was required when the coil was tested on several subjects with different head sizes. Human subjects participated in the study after giving signed informed consent according to procedures approved by the local institutional ethics committee. All data were acquired on a Siemens MAGNETOM (Erlangen, Germany) 7 T whole-body human MRI scanner. The developed dipole array Tx performance was compared with that of the commercial 8 Tx/32 Rx ToRo-array (Nova Medical, Wilmington, MA, USA). $B_1^+$ maps were obtained using the 3D actual flip angle imaging (AFI) sequence (field of view [FoV]: 244 x 244 x 100 mm³, voxel size: 1.8 x 1.8 x 5 mm³, TR₁/TR₂: 20/100 ms, TE: 4 ms, flip angle [FA]: 60°). $B_1^+$ maps were normalized to the RF power level at the coil input. Images were obtained with a 3D gradient echo sequence (TR/TE: 18/2 ms, FA: 8°, voxel size 2 mm isotropic, FoV: 240 x 240 x 192 mm). To evaluate the performance of the new dipole array in a typical UHF application, we also performed MP2RAGE measurements for the dipole array and Nova array coil, acquired with 1-mm isotropic resolution in approximately 11 min (other parameters: TRₑₒᵣᵢₙ = 6/6000 ms, FA₁/₂ = 5°/9°, TI₁/₂ 800/2000 ms, 6/8 partial Fourier in partition direction, 10 ms tr-FOCI inversion).
3  |  RESULTS

3.1  |  EM simulations: A comparison of the dipole arrays without decoupling

First, we tested various alterations of the design of the folded-end dipole array shown in Figure 1A. The results of the evaluation are presented in Figure 1B and Table 1. As can be seen from the first picture in Figure 1A, dipoles aligned with the top of the head produced a very low RF field in the superior area of the head. Also, an addition of the local RF shield to this design, as suggested by Avdievich et al., did not lead to any significant changes in the $B_1^+$ field distribution (not shown). However, extending the dipoles above the head increased $B_1^+$ both in the superior head area as well as near the center of the head. Overall, the 40-mm position provided superior results in comparison with both the 0-mm and 20-mm positions. For example, $<B_1^+>/\sqrt{P}$ and SEE values measured for the 40-mm position were 28% and 35% higher than those obtained for the 0-mm position. Homogeneity was also significantly improved (Table 1). Therefore, we used the design with the 40-mm extension for further evaluations.

Also, as can be seen in the last two pictures of Figure 1A, the addition of the superior shield to the 40-mm array design further increased $B_1^+$ values over the entire brain. Also, $<B_1^+>/\sqrt{P}$ and SEE were increased by 21% and 8%, respectively. Moving the shield closer to the head did not substantially change the $B_1^+$ distribution, but it did decrease $<B_1^+>/\sqrt{P}$ and SEE by 5.1% and 1.4%, respectively (Table 1); it also reduced homogeneity (SD/$<B_1^+>$) by approximately 10%.

Increasing the diameter of the 40-mm array design by 20 and 40 mm caused a decrease in $<B_1^+>/\sqrt{P}$ of 4.3% and 9.9%, respectively (Table 1). At the same time, SEE values were very similar (within 2%) for all three designs, including the original 40-mm as well as both larger
arrays. Figure S1 shows the results of RF shimming using 500 random shim values. The average of the local pSAR values was 0.725 and 0.549 W/kg, while the average $B_1^{\text{eff}}$ efficiency was 0.324 and 0.273 $\mu$T/$\sqrt{W}$ for the original and the largest arrays, respectively. Finally, the average SEE was 0.385 and 0.377 ($\mu$T/$\sqrt{kg})/\sqrt{W}$ for the small and large arrays, respectively.

Figure 2 compares results obtained for four different dipole array designs all shifted by 40 mm and loaded by the Duke voxel model. All four designs also include the superior shield (Figure 2A). Table 1 and Figure 2B present data obtained for dipole arrays shown in Figure 2A. As can be seen in Table 1, both folded-end and unfolded bent dipole arrays provided very similar (within 1%) $<B_1^{+}>/\sqrt{P}$ and SEE. Both straight dipole arrays demonstrated lower $<B_1^{+}>/\sqrt{P}$ (i.e. by 7.3% and 12.6%) but similar SEE. Homogeneity was very similar for all four arrays.

As evaluation of local heating is often assessed by calculating SAR10g distributions; Figure 5A shows examples of sagittal SAR10g maps calculated for bent folded-end and unfolded arrays. Figure 5B shows transverse SAR10g maps all cut through the ears. Locations of the transverse slices, cut through positions of pSAR10g, are shown in Figure 5A. In addition, to demonstrate the effect of dipole length on local SAR values, we included data calculated for the shorter 170-mm bent unfolded dipole array. This is the same length as we used previously in designing the 9.4 T dipole arrays.21–23 The ends of the shorter array were aligned with the top of the head, and the superior shield was positioned at a 30-mm distance from the head.21–23 The shorter array produced the highest level of SAR10g (0.66 W/kg) in the ears. Both 250-mm-long dipole arrays (i.e. unfolded and folded-end) produced substantially lower SAR10g values in this area (Figure 5B).

As demonstrated previously,15,24 high voltage (electric field) at the ends of dipole antennas positioned close to the sample, as found in TxRx or Rx tight-fit arrays, may produce a substantial shift of the resonance frequency due to a variation in the head size. In the folded-end dipole design, this effect is reduced15 as a consequence of moving both ends of the dipole away from the sample. To evaluate this effect, we tuned both bent dipole arrays (folded-end and unfolded) on the Duke voxel model and checked the changes of tuning and matching by placing the smaller Ella voxel model inside without any further adjustments. Figure S2 shows $S_{11}$ frequency dependences for all eight elements of both arrays loaded by the Ella voxel model, demonstrating that frequency shifts measured for the unfolded bent dipole array were substantially higher, and as a result reflection from some elements increased by up to ~5 dB.

### Table 1: Simulated and experimental results (circularly polarized mode)

| Voxel model | Array | Position, decoupling | $<B_1^{+}>/\sqrt{P}$, $\mu$T/$\sqrt{W}$ | $<B_1^{+}>/\sqrt{P}$, ratio | SD/$<B_1^{+}>$ | pSAR10g, W/kg | SEE, $\left(\mu$T$/kg\right)/\sqrt{W}$ | SEE, ratio | $P_T^{d}$, W |
|-------------|-------|----------------------|---------------------------------|-------------------|----------------|----------------|---------------------------------|-----------|---------|
| Duke        | Bent fold.-end | 0-mm | 0.264 | 0.35 | 0.430 | 0.403 | 1 | 0.685 |
|             |       | 20-mm | 0.313 | 1.19 | 0.27 | 0.434 | 0.475 | 1.18 | 0.658 |
|             |       | 40-mm | 0.338 | 1.28 | 0.24 | 0.388 | 0.543 | 1.35 | 0.618 |
| Bent fold.-end (local shield) | | 20-mm | 0.363 | 1.38 | 0.24 | 0.466 | 0.532 | 1.32 | 0.731 |
|             |       | 40-mm | 0.410 | 1.55 | 0.25 | 0.487 | 0.587 | 1.46 | 0.764 |
|             |       | 40-mm | 0.391 | 1.48 | 0.24 | 0.446 | 0.586 | 1.45 | 0.726 |
|             |       | (20-mm larger)$^b$ | | | | | | |
|             |       | (40-mm larger)$^c$ | | | | | |
|             |       | 40-mm | 0.373 | 1.41 | 0.24 | 0.417 | 0.577 | 1.42 | 0.695 |
|             |       | 40-mm | 0.390 | 1.48 | 0.28 | 0.482 | 0.579 | 1.44 | 0.731 |
|             |       | 40-mm | 0.399 | 1.51 | 0.29 | 0.509 | 0.559 | 1.39 | 0.795 |
|             |       | 40-mm | 0.407 | 1.54 | 0.26 | 0.494 | 0.579 | 1.44 | 0.731 |
| Bent (local shield) | | 40-mm | 0.408 | 1.55 | 0.25 | 0.476 | 0.592 | 1.47 | 0.75 |
|             |       | 40-mm | 0.374 | 1.42 | 0.27 | 0.457 | 0.553 | 1.37 | 0.588 |
|             |       | 40-mm | 0.320 | 1.21 | 0.26 | 0.390 | 0.512 | 1.27 | 0.423 |
| Straight (local shield) | | 40-mm | 0.382 | 1.46 | 0.24 | 0.437 | 0.578 | 1.42 | 0.692 |
|             |       | 40-mm | 0.364 | 1.38 | 0.24 | 0.402 | 0.573 | 1.41 | 0.666 |
| Straight fold.-end (local shield) | | 0-mm | 0.333 | 1.26 | 0.34 | 0.660 | 0.411 | 1.02 | 0.609 |
|             |       | 40-mm | 0.428 | - | 0.25 | 0.506 | 0.602 | - | 0.738 |

$^a$Calculated for 1 W of RF power at the array input; $<B_1^{+}>$ is averaged over 130-mm transverse slab.

$^b$The array size was increased by 20 mm by moving each dipole element 10 mm away.

$^c$The array size was increased by 40 mm by moving each dipole element 20 mm away.

$^d$Power absorbed in the tissue per 1 W of RF stimulated power at the array input.

$^e$The superior shield is moved by 20 mm closer to the head (Figure 1A, the last picture).
3.2 | EM simulations: Decoupling of the array

In our previous work at 9.4 T, we demonstrated that passive decoupling dipoles can produce strong destructive interference with the dipole array CP mode. As a result, the $B_{1+}$ field was substantially reduced and peripheral voids occurred in the $B_{1+}$ distribution. To evaluate whether at 7 T the addition of passive decoupling dipoles causes similar destructive interference with the Tx field of the array, we numerically simulated both bent folded-end and unfolded dipole arrays decoupled using the two methods shown in Figure 3. Figure 6A depicts central sagittal $B_{1+}$ maps obtained using the decoupled bent folded-end dipole array as well as the same array without any decoupling. Figure 6B shows results of the similar evaluation procedure performed for the bent (unfolded) array. In all cases arrays were loaded using the Duke voxel model. In addition, Figures S3 and S4 show similar transversal and coronal slices obtained using the bent folded-end dipole array and bent (unfolded) array. Figure 7 shows ratios of $B_{1+}$ maps obtained using decoupled bent folded-end and unfolded arrays to those obtained using the same arrays without decoupling. Ratio maps are evaluated for central transverse, coronal and sagittal slices. Table 1 presents quantitative evaluation of the Tx performance of the decoupled arrays. As can be seen from Figures 6, 7, S3 and S4 and Table 1, the Tx performance of the bent folded-end dipole array is only slightly reduced because of the addition of passive dipoles. $<B_{1+}>/\sqrt{P}$ is reduced by 2.7% and 0.7% for decoupling 1 and 2, respectively; SEE is reduced by 5% and 1.4%, respectively. Thus, the reduction is negligible for decoupling method 2. At the same time, peripheral voids in the $B_{1+}$ maps are seen in the transverse slices (Figures 7A and S3A). For decoupling 2, however, the voids are substantially reduced and barely noticeable. The effect of destructive interference is significantly stronger in the case of the bent unfolded dipole array and is especially pronounced for decoupling method 2. $<B_{1+}>/\sqrt{P}$ is reduced by 9% and 27.5% for decoupling methods 1 and 2, respectively; SEE is reduced by 7% and 15.6%, respectively. Peripheral voids in the $B_{1+}$ maps are also clearly seen for both transverse slices, as shown in Figures 7D and S3B.

It is interesting that decoupling method 2 caused such a strong reduction of $<B_{1+}>/\sqrt{P}$ in the case of unfolded dipoles. This indicates that a stronger current is induced in the passive dipoles. As demonstrated previously, the amplitude of the induced current is $\sim |X_{C}| X$, where $X_{C}$ is the absolute value of the impedance of the mutual capacitance between adjacent active dipoles (i.e. $-jX_{C}$), and $X$ is the absolute value of the impedance of the mutual capacitance between active and passive dipoles (i.e. $-jX$). We numerically calculated the reactive mutual impedance between a pair of folded-end and unfolded active dipoles and an active and passive dipole loaded by the Duke voxel model (Figure 4E, active dipoles 1 and 8). The mutual reactive impedance between two active dipoles was $-j4.6$ and $-j6.1$ ohms for folded-end and unfolded dipoles, respectively. The reactive mutual impedance between an active and passive dipole was $-j19.5$ and $-j17.4$ ohms for decoupling methods 1 and 2, respectively, in the
FIGURE 6  Electromagnetic (EM) simulated central sagittal $B_1^+$ maps obtained using bent folded-end (A) and bent (unfolded) (B) dipole arrays without decoupling and decoupled using methods 1 and 2. All arrays are loaded by the Duke voxel model.

FIGURE 7  Transverse (A), sagittal (B) and coronal (C) maps of the ratios of $B_1^+$ field obtained using decoupled bent folded-end dipole array to that obtained by the array without decoupling. Figure 7B shows the position of the transverse slice. The sagittal and coronal slices are taken approximately through the head center. Transverse (D), sagittal (E) and coronal (F) maps of the ratios of $B_1^+$ field obtained using decoupled bent (unfolded) dipole array to that obtained by the array without decoupling.
case of bent folded-end dipoles. In the case of bent (unfolded) dipoles, the impedance of the mutual capacitance between an active and a passive dipole was -j19.9 and -j10.4 ohms for decoupling methods 1 and 2, respectively. Thus, moving the passive dipoles away from the sample caused a substantial reduction of $X$ in the case of the unfolded dipoles and, correspondently, the higher current induced in the passive dipoles compared with the case of the folded-end dipoles. Finally, moving the passive dipoles away from the sample caused the current in the passive dipoles to increase by factors of 1.12 and 1.90 for the bent folded-end and unfolded dipoles, respectively.

As discussed above in the Methods section, in all decoupled array designs we only used seven passive dipoles, with the passive dipole near the nose removed. Depending on the decoupling method (method 1 or 2), head size (Duke or Ella), and head position, the addition of the dipole in this location caused up to a threefold increase in the local SAR (i.e. in the range of 1.5 W/kg and higher).

### 3.3 Experimental evaluation

After numerical evaluation of the array performance and its safety assessment, we constructed the dipole array. The final array design included eight unshielded bent folded-end dipoles uniformly surrounding the head and seven straight passive dipoles attached at the array cover (Figure 4A–C). Figure 4D shows an 8 x 8 S-matrix measured using the bent folded-end dipole array loaded with a male head (570 mm in circumference). S-matrices were measured for the coupled and decoupled versions of the array. The average coupling measured between neighboring channels was −18.3 and −12.1 dB for the decoupled and coupled array versions, respectively. The average coupling measured between next-neighboring channels (i.e. 1 and 3, 2 and 4, etc.) was −18.1 and −20.2 dB for the decoupled and coupled arrays, respectively.

Figure 8A shows in vivo data including central sagittal, coronal and transverse images and corresponding $B_{1+}$ maps obtained for the same male subject (570 mm in circumference) using the constructed 7 T bent folded-end dipole array. For comparison, Figure 8C shows similar $B_{1+}$ maps obtained using the commercial array. Both arrays were driven in the CP mode. All $B_{1+}$ maps are normalized to the coil input for comparison with simulated data. Averaged over a 130-mm transverse slab, which includes the majority of the brain, $<B_{1+}>$ was 9.70 and 7.96 $\mu$T/W (or 0.31 and 0.25 $\mu$T/W) for the constructed dipole array and the commercial array, respectively. As can be seen from Figure 8, the dipole array provided significantly better coverage, especially in the brain stem area. In addition, Figure 8B shows sets of in vivo transversal $B_{1+}$ maps spanning the majority of the brain that were obtained using the constructed bent folded-end dipole array and commercial array for the same subject as in Figure 8. Positions of the slices are marked in Figure 8C. To directly compare the level of $B_{1+}$ field inhomogeneity, all maps are scaled to corresponding $B_{1+}$ maximums.

Finally, Figure 9A shows examples of sagittal, coronal and transverse isotropic MP2RAGE images acquired using the unshielded bent folded-end dipole array decoupled by method 2 and the same subject as in Figure 8. For comparison, Figure 9B shows similar slices obtained using the commercial array and the same subject. Driven in the CP mode, the new dipole array demonstrates excellent coverage and contrast without any additional RF shimming of the Tx field. By contrast, the commercial array shows a region with poor contrast, due to significantly lower values of the $B_{1+}$ field.

#### 4 DISCUSSION

We developed and evaluated the TxRx bent folded-end 8-element dipole array for human head imaging at 7 T. This type of dipole antenna was successfully used previously in designing a 9.4 T head array, which demonstrated an improvement in longitudinal $B_{1+}$ coverage (along the magnetic axis) and a reduction in pSAR compared with the unfolded dipole array design. This improvement in Tx performance was achieved due to a more uniform current distribution along the loaded portion of the folded-end dipole compared with the unfolded dipole of similar physical length. Alteration to the design (i.e. extension of the dipole elements above the head and addition of the local superior shield) resulted in significant improvement in Tx performance, which was evaluated both as $<B_{1+}> / \sqrt{IP}$ and $<B_{1+}> / \sqrt{P \text{SAR}_{10g}}$ (SEE). As seen in Table 1, $<B_{1+}> / \sqrt{IP}$ and SEE increased by 55% and 46%, respectively, due to both the 40-mm extension and local shield addition. The homogeneity of the $B_{1+}$ distribution also improved significantly. The addition of only the superior shield led to an increase in $<B_{1+}> / \sqrt{IP}$ of approximately 20%. Similar modification of the design at 9.4 T (400 MHz) improved the $B_{1+}$ field at the superior head area but did not cause any significant change in the $B_{1+}$ near the head center or average $<B_{1+}>$. This implies that at 300 MHz, the RF field reflected by the local shield interferes constructively with the RF field produced by the array at the head center due to the phase shift accumulated by the RF field propagating from the edge to the center of the head. At the same time, moving the shield closer to the head did not cause much change. This can be explained by the fact that the phase shift is mainly determined by the RF field propagation inside the head, which has a significantly higher dielectric constant than air. Similar to previously published data for an 8-element TxRx 9.4 T loop array, increasing the size of the dipole array by moving the dipoles away from the head caused a corresponding decrease of $<B_{1+}> / \sqrt{IP}$ and pSAR$_{10g}$ value (Figure 5E and Table 1). At the same time, SEE almost (within 2%) did not change.

Comparison of the Tx performance of the constructed dipole array with that of the commercial ToRo-array revealed significant improvement in the longitudinal coverage and Tx efficiency. Driven in the CP mode, the new bent folded-end dipole array provides more uniform $B_{1+}$ distribution in the majority of the brain (Figure 5E). The higher Tx efficiency of the dipole array is also explained by its smaller size compared with the commercial loop Tx array. However, the size of the dipole Tx array does not need be significantly increased when combined with additional Rx-only loops. Commonly,
the ToRo-combination of a Tx-only loop array with an Rx-only loop array requires two separate housings for both the Rx and Tx arrays. Therefore, the size of the loop Tx array must be sufficiently large to accommodate the Rx array inside it. The simplicity of the dipole array design (i.e. eight wires, no distributed capacitors, etc.) enables combining dipoles and Rx loops within a single layer, or slightly elevated above the Rx loops\textsuperscript{15,37,38}.

The bent folded-end dipole array provided better Tx performance than both the straight folded-end and unfolded arrays of similar length and diameter (Table 1). However, the bent unfolded array of similar geometry demonstrated very similar $B_{1+}$/$\sqrt{P}$, SEE and homogeneity values (Table 1). Thus, in contrast to 9.4 T, the more uniform current distribution of the folded-end dipole array did not cause a significant change in $B_{1+}$ distribution. This is most likely due to the extension of the dipoles above the head. For example, as seen in Figure 5 and Table 1, the shorter 170-mm bent (unfolded) dipole array produces substantially higher local SAR in the ears, which in turn causes a decrease in the SEE value. By increasing the length of the unfolded dipoles and extending them above the head, the level of local SAR in the ears was substantially decreased. SAR\textsubscript{10g} in the ears was further reduced to a negligible level for the bent folded-end dipole array (Figure 5B).

An additional benefit of the folded-end design is a reduction in the frequency change due to variation in head size. As shown in Figure S2, the folded-end dipole array demonstrated significantly less change in element resonance frequencies when both arrays were first tuned on the Duke voxel model and then loaded by the smaller Ella voxel model without readjustment. This effect is explained by the lower voltage along the dipole portion positioned near the head in the case of the folded-end dipoles. Thus, for optimal performance, a tight-fitting unfolded dipole array needs to be tuned individually on each head, which is not feasible in a clinical environment. Alternatively, the dipole elements can be moved farther away from the sample, and the diameter of the array increased. However, this is suboptimal when the array is used in TxRx mode.

In addition to modification of the geometry of the dipole elements, we tested two decoupling methods using passive dipole antennas\textsuperscript{12,23,25–27} shown in Figure 3. These methods allow coupling between adjacent active dipoles to be minimized without any electrical connection. This feature is
a critical part of any dipole array design because individual antennas are located at relatively large distances from each other, and common
decoupling techniques, which require electrical connections (i.e. capacitive, inductive), cannot be easily used. Alternatively, the dipoles must
be brought closer to each other, at least at one end, which limits the flexibility in choosing the geometry of the dipole antennas. In the final version
of the bent folded-end dipole array decoupled by method 2, the average coupling between adjacent elements was reduced by approximately 6 dB.
It is noteworthy that even although we did not introduce a passive dipole between active dipoles 4 and 5, the corresponding $S_{45}$ value was still
reduced by more than 3 dB (Figure 4D). This occurred due to coupling between adjacent passive dipoles. This interaction also led to a small (about) increase in coupling between the next-neighboring (i.e. 1 and 3, 2 and 4, etc.) active elements (Figure 4D). However, the degradation is sufficiently
small to preserve a reasonable level of the average coupling (i.e. approximately $-18$ dB).

Because relatively large passive dipoles may produce significant destructive interference with the CP mode of the array, we numerically
evaluated the Tx performance of both the bent folded-end and unfolded decoupled dipole arrays (Figure 3). Our assessment of the proposed bent
folded-end array design decoupled by passive dipoles revealed only a small reduction in the $\langle B_1^+ \rangle / \sqrt{P}$ and SEE values (Table 1), which was almost
negligible ($\sim 1\%$) for decoupling method 2. It is interesting that at 9.4 T, the presence of passive dipoles (method 2) produced substantially higher
destructive interference and caused approximately 15% reduction of $B_1^+$ near the center. Also, in comparison with 9.4 T, where decoupling
method 2 caused significant distortion of the $B_1^+$ distribution at the periphery, at 7 T the peripheral voids (Figure 7A) were reduced to an acceptable
level. In contrast to the case of the bent folded-end array, the $\langle B_1^+ \rangle / \sqrt{P}$ and SEE values obtained for the decoupled bent (unfolded) array
were substantially reduced, especially in the case of method 2 (Table 1). Also, the peripheral voids were clearly seen in the $B_1^+$ distribution pro-
duced by the decoupled bent unfolded dipole array (Figures 7D and S3B). It is interesting that moving the passive dipoles farther away from the sample, which intuitively implies a decrease in the interference between the RF fields of the passive and active dipoles, leads to an opposite result
in the case of the unfolded array. Previously, we observed a similar effect at 9.4 T. As shown above in the Methods section, moving the passive
dipoles away from the sample (method 2) caused a substantial reduction in the absolute value of the mutual capacitive impedance between active
and passive dipoles, $X$, in the case of the unfolded active dipoles. This also resulted in a significant increase (about 1.7-fold) in the current in the
passive dipoles. At the same time, as demonstrated previously, the RF field produced by the passive dipoles does not decrease at the same rate.

FIGURE 9 Examples of sagittal, coronal and transverse in vivo 1-mm isotropic whole-brain MP2RAGE images obtained using the constructed
bent folded-end dipole array (A) and commercial array (B) for the same subject as in Figure 8. The area of the reduced contrast in each image
obtained by the commercial array is marked by a red dashed line.
when the dipoles are moved away from the sample. As a result, the largest destructive interference level was observed in the case of the unfolded dipole array decoupled by method 2. These results provide an important message that not only the physical size of passive dipoles must be reduced to produce minimal interference with the RF field of the active dipoles, but also the coupling between the active and passive dipoles must be maximized to reduce the current induced in the passive dipoles. These two criteria must be satisfied simultaneously, which, unfortunately, may not be an easy task.

The developed array consists of a relatively small number of TxRx dipole elements, and obviously does not provide optimal SNR comparable with state-of-the-art head coils equipped with a multielement (e.g. 32 loops) Rx array. Further improvement of SNR can be obtained by combining the developed TxRx dipole array with multiple Rx-only surface loops, as suggested previously. Simultaneous reception using loops and dipoles in a human head array was also demonstrated by Avdievich et al. at 9.4 T: to minimize interaction between loops and dipoles, the dipole antennas were placed in the center of the loops. As a result, this design provided significant improvement (~30%) in central SNR.

5 | CONCLUSION

We developed, constructed and evaluated an 8-element, TxRx bent folded-end dipole array for human head imaging at 7 T. Driven in the quadrature CP mode, the array demonstrated more than 20% higher transmit efficiency and significantly better whole-brain coverage than that provided by a widely used commercial array. In addition, we evaluated passive dipole antennas for decoupling the proposed array. We demonstrated that, in contrast to the common unfolded dipole array, the passive dipoles moved away from the sample (decoupling method 2) not only minimize coupling between the adjacent folded-end active dipoles, but also produce practically no destructive interference with the Tx field of the array.

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DATA AVAILABILITY STATEMENT

Data are available upon request from the authors.

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SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section at the end of this article.

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