Decoupling of folded-end dipole antenna elements of a 9.4 T human head array using an RF shield

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1INTRODUCTION

Dipole antennas have recently been introduced to the field of MRI and successfully used, mostly as elements of ultra-high field (UHF, ≥ 7 T) human body arrays. Usage of dipole antennas for UHF human head transmit (Tx) arrays is still under development. Due to the substantially smaller size of the sample, dipoles must be made significantly shorter than in the body array. Additionally, head Tx arrays are commonly placed on the surface of rigid helmets made sufficiently large to accommodate tight-fit receive arrays. As a result, dipoles are not well loaded and are often poorly decoupled, which compromises Tx efficiency. Commonly, adjacent array elements are decoupled by circuits electrically connected to them. Placement of such circuits between distantly located dipoles is difficult. Alternatively, decoupling is provided by placing passive antennas between adjacent dipole elements. This method only works when these additional components are sufficiently small (compared with the size of active dipoles). Otherwise, RF fields produced by passive elements interfere destructively with the RF field of the array itself, and previously reported designs have used passive dipoles of about the size of array dipoles. In this work, we developed a novel method of decoupling for adjacent dipole antennas, and used this technique while constructing a 9.4 T human head eight-element transceiver array. Decoupling is provided without any additional circuits by simply folding the dipoles and using an RF shield located close to the folded portion of the dipoles. The array reported in this work demonstrates good decoupling and whole-brain coverage.

KEYWORDS
array optimization, decoupling of dipole antennas, human head transceiver array, transmit performance, ultra-high field MRI, whole-brain coverage

1 | INTRODUCTION

Dipole antennas have recently been introduced to the field of MRI and successfully used, mostly as transmit (Tx) or transceiver (TxRx) elements of ultra-high field (UHF, ≥ 7 T) human body arrays.1-3 It is important that elements of a body array can be placed at a fixed distance very close to
the tissue, and as a result heavily loaded antennas are very well decoupled, which is critical for optimal Tx performance. Usage of dipole antennas as elements of UHF human head arrays is still under development.4–9 Due to the substantially smaller size of the sample (the human head), dipoles must be made physically shorter than in body arrays, which decreases loading. Human head Tx arrays are also commonly placed on the surface of rigid helmets, which must be constructed large enough to accommodate tight-fit receive (Rx) arrays inside. As a result, adjacent dipoles are not sufficiently loaded and are often poorly decoupled.7,9,10 This restricts the choice of array geometries and may cause a decrease in the performance. Another option is to place additional passive decoupling antennas between adjacent dipole elements of a Tx (or TxRx) array.8,16 This method has certain drawbacks and works properly only in the case when additional decoupling components are small in comparison with the distance to the sample and size of major elements of the array. Otherwise, RF fields produced by passive elements interfere destructively with the RF field of the array itself.16 Since previously reported passive decoupling dipoles8,18 are about the same size as array dipoles, they inevitably produce a sufficiently strong RF magnetic field, \( B_1 \), within the sample. Unfortunately, authors have yet to report a comparison of the RF magnetic field distribution created by arrays with and without passive decoupling elements. In addition, these decoupling dipole antennas can strongly interact with Rx elements when a smaller tight-fit Rx array is placed inside of the Tx dipole array, ie a so-called transmit-only/receive-only (ToRo) system.19 For example, in Reference 10 the authors described a ToRo array design consisting of a Tx dipole array and a smaller 32-loop Rx array. The Tx array has essentially the same geometry as the TxRx array reported in Reference 8 with the difference that the authors of Reference 10 had to remove passive decoupling dipoles since “they could not be detuned, and therefore, interfered with the receive array adjustments.” As a result, the Tx dipole array was poorly decoupled.10 Thus, in our opinion, decoupling of the dipole elements of human head Tx (TxRx) arrays remains a very important design issue, and has yet to be solved.

In this work, we develop a novel decoupling method for adjacent dipole antenna elements of aTxRx human head array at 9.4 T (\( 1^H \) frequency of 399.72 MHz). Decoupling is provided without any additional electronic circuits by simply folding the dipoles and then using an RF shield located closely to the folded portion of the dipoles. Finally, we construct and evaluate a single-row array consisting of eight bent folded-end shielded dipole elements.

2 | METHODS

2.1 | Decoupling method

In comparison with adjacent surface loops, which are coupled inductively (a mutual inductance), a pair of loaded dipole antennas is coupled capacitively,20 which implies that imaginary parts of off-diagonal elements of the mutual impedance matrix change their signs from positive (inductive coupling) to negative (capacitive coupling). Therefore, to decouple adjacent dipole elements, we need to create additional inductive coupling between them. Figure 1A demonstrates how this inductive coupling can be produced by folding the ends of the dipoles and placing an RF shield near to their folded portions. The presence of the shield is electrically equivalent to placing a mirror image of the dipole carrying a current of the opposite sign (Figure 1A) at a distance double the distance to the shield. Two folded portions of both dipoles (a real dipole and its mirror image) produce a capacitive bridge, which depends on the length of the folded portion and the distance to the RF shield. Replacing both folded portions with an equivalent capacitor creates a loop. As a result, two shielded adjacent folded-end dipoles are coupled inductively.

![Schematic representation of the decoupling effect caused by folding the dipoles and placing the RF shield close to the folded portion. The effect is demonstrated for straight folded-end dipoles (A), bent folded-end dipoles (B), straight dipoles (C), and bent dipoles (D).](image-url)
Changing the effective capacitance value, $C_{\text{eff}}$, affects the current, and therefore the inductive coupling. As a result, by adjusting the distance to the shield and length of the folded portion, we can vary this mutual inductance to compensate the intrinsic capacitive coupling between two dipoles. In the text below we will refer to the folded-end dipoles shown in Figure 1A as straight folded-end dipoles. In addition to the dipoles shown in Figure 1A, we also evaluated bent folded-end dipoles (Figure 1B), straight dipoles (Figure 1C), and bent dipoles (Figure 1D). As demonstrated previously, folding and bending the dipoles improves the $B_1^+$ and specific absorption rate (SAR) distribution due to coupling to the intrinsic transverse electric (TE) mode of the human head.

2.2 | Electromagnetic (EM) simulations

EM simulations were performed using CST Studio Suite 2017 (CST, Darmstadt, Germany) and the time-domain solver based on the finite-integration technique. First, we numerically evaluated the $S_{12}$ coupling coefficient between pairs of the four different types of dipole shown in Figure 1. All dipoles were placed at the posterior location of a head-and-shoulder (HS) phantom. Figure 2A shows an example of the CST model for a pair of straight folded-end dipole antennas loaded by the HS phantom. The RF shield is hidden for better visualization. All dipoles had a length of 175 mm and were constructed using 1.5-mm copper annealed wire. Simulations of different wires (1 mm, 1.5 mm, 2 mm) show a negligible effect on the $B_1^+$ field value for a single element. The choice of 1.5-mm wire was mostly determined by its mechanical properties, which implies that the wire is sufficiently thick and stable but still flexible. For future comparison, the length of the dipoles was chosen to be same as the total length of the previously constructed 16-element double-row ($2 \times 8$) surface loop TxRx array,\cite{14,16,22,23} ie 175 mm. All matching and tuning circuits were taken into account in the CST Design Studio module. The RF copper shield was placed at 40-mm distance from the straight portion of the dipole elements. The distance to the shield was chosen based on previously published data for UHF human head size TxRx loop arrays.\cite{14,16,22,23} To increase $B_1^+$ field at the head superior location,\cite{24,25} we also added a passive elliptical RF shield (175 mm $\times$ 140 mm) placed 30 mm away from the head. Figure 2B shows the CST model of the eight-element ($1 \times 8$) bent folded-end dipole array loaded by the “Duke” human head voxel model.\cite{26} A larger copper cylinder (640 mm in diameter and 1600 mm in length) was also included in the model to mimic the RF shield of the gradient coil. To evaluate the dependence of coupling on loading, which is determined by the distance to the sample, we calculated $S_{12}$ values for the same pair of dipoles loaded by a smaller phantom, obtained by scaling the HS phantom by a factor of 0.9. This corresponds approximately to the variation in human head sizes. Before constructing the final version of the array, we also evaluated the Tx $B_1^+$ field and maximum local SAR values for an array of eight dipole antennas circumscribing a head. The final geometry of the folded-end dipole elements was chosen based on the best decoupling value and optimal Tx performance. To further understand the mechanism of inductive coupling between adjacent shielded folded-end dipoles, we also evaluated the imaginary part of the mutual impedance, $\text{Im}(Z_{12})$, for different pairs of dipoles loaded by the HS phantom. In this case the dipoles were not matched but tuned using inductors connected in series.

Two voxel models were used in simulations: the HS phantom, which was constructed to match tissue properties ($\varepsilon = 58.6, \sigma = 0.64 \text{ S/m at } 400 \text{ MHz}$),\cite{27} and a virtual family multi-tissue model, “Duke”,\cite{26} cropped at the chest level. For both voxel models, we used a resolution of 2 mm. $B_1^+$ field profiles and local SAR$_{10g}$ (averaged over 10 g of the tissue) maps were calculated for RF power of 1 W at the array input and then

**FIGURE 2**  (A) CST EM simulation model of a pair of straight folded-end dipoles loaded by the HS phantom. (B) Eight-element ($1 \times 8$) bent folded-end dipole array loaded by the Duke voxel model. In A, the RF shield is hidden for better visualization.
compared with experimentally measured data. Averaging of SAR was performed using the CST Legacy method. Additionally, we evaluated the Tx performance, both as \( \frac{\langle B_1^+ \rangle}{P} \), where \( P \) is the RF power measured at the array input, and as \( \frac{\langle B_1^+ \rangle}{\max \text{SAR}^{10g}} \), ie the safety excitation efficiency (SEE). The \( B_1^+ \) value was averaged over a 130-mm central transversal slab, which includes the majority of the human brain.

2.3 | Array construction

After numerical optimization of decoupling and the Tx performance of the array, we constructed the final eight-element dipole array. The geometry of the array holder and all elements were the same as in EM modeling. Figure 3 shows photographs and a schematic diagram of the array. The array consisted of eight TxRx bent folded-end dipoles (Figure 2B) with the length of the folded portion equal to 30 mm. We will refer to this element design as the 30-mm bent folded-end dipole in the text below. Dipoles were constructed using 1.5-mm tinned copper wire and measured 33 mm in height and 175 mm in length. The distance from the folded portion to the RF shield measured 7 mm. All inductors were hand made of 1.0-mm tinned copper wire and adjusted to bring the resonance frequency to 399.75 MHz. Increasing the diameter of the wire increases the \( Q \)-factor of the inductors. T/R switches were connected to each dipole element and located inside the array holder to minimize losses. Low-noise preamplifiers (WanTcom, Chanhassen, MN, USA) were incorporated into the T/R switch units. To prevent wave propagation along the cable, a shielded cable trap was introduced at the input of each dipole element. The array was shielded, with the RF shield located at 40-mm distance from the array holder. The RF shield measured 280 mm in length and was constructed using a double-sided Kapton (25 \( \mu \)m thickness) copper clad laminate (AKAFLEX, Krempel, Germany) as previously described. Sixteen capacitive segments \( \approx 7 \) nF were produced by overlapping inner and outer copper layers. Comparison with the unshielded array coil revealed no visible artifacts from eddy currents induced in the RF shield by fast switching sequences. During transmission, the array was driven in circularly polarized (CP) mode, which in our case corresponds to a 45° phase shift between adjacent dipoles. For this purpose, we constructed an eight-way Wilkinson splitter with corresponding fixed phase shifters incorporated into the splitter box. Phase shifters were fabricated using 50 \( \Omega \) cables.

2.4 | Experimental evaluation of the array performance

Before making in vivo measurements, the new phased array was evaluated on a bench and in the scanner and numerically simulated according to the safety procedure developed in our laboratory. The human subjects participated in the study after giving signed informed consent according to procedures approved by the local institutional review board committee. Bench evaluation of the array included measurements of the entire \( S_{12} \) matrix (8 \( \times \) 8) using a network analyzer (E5071C, Agilent Technology, Santa Clara, CA, USA). The \( S_{12} \) matrix was evaluated both in vivo and using the HS phantom.
All data were acquired on a Siemens Magnetom (Erlangen, Germany) 9.4 T human imaging system. \(B_1^+\) maps were acquired using the 3D actual flip angle imaging sequence\(^{31}\) (field of view = \(244 \times 244 \times 100\) mm\(^3\), voxel size = \(1.8 \times 1.8 \times 5\) mm\(^3\), TR\(_1)/TR\(_2\) = 20/100 ms, TE = 4 ms, flip angle (FA) = 60°). To evaluate the performance of the new dipole array in a typical UHF application, we also performed in-vivo MP2RAGE\(^{32}\) measurements for the eight-element single-row (1 × 8) dipole array. MP2RAGE images were acquired at 1 mm isotropic resolution in about 11 min (other parameters: TR\(_{\text{ro/inv}}\) = 6/6000 ms, FA\(_{1/2}\) = 5°/9°, TI\(_{1/2}\) = 800/2000 ms, 6/8 partial Fourier in partition direction, 10 ms TR-FOCI inversion).

3 | RESULTS

3.1 | EM simulations

In the first step, we numerically evaluated coupling between a pair of different shielded dipole antennas loaded by the HS phantom. Figure 4A shows the dependence of the \(S_{12}\) value on the length of the folded portion for four different heights (Figure 3C) of the straight folded-end dipoles. As seen from the figure, for each height of the dipole, a corresponding folded length can be chosen that provides the lowest \(S_{12}\) value. Figure 4B shows similar dependences obtained for two heights of the straight folded-end dipole and two different size HS phantoms. The smaller HS phantom is obtained by scaling the original HS phantom by a factor of 0.9. Dashed lines correspond to the smaller HS phantom. Both solid lines are shown for better comparison and are the same as in A.

Figure 5 (A) Dependences of \(S_{12}\) between two bent folded-end dipole antennas on the length of the folded portion. \(S_{12}\) is measured for four different heights of the dipoles loaded by the HS phantom. (B) Similarly to A, dependences are acquired for two heights of the bent folded-end dipole and two different size HS phantoms. The smaller HS phantom is obtained by scaling the original HS phantom by a factor of 0.9. Dashed lines correspond to the smaller HS phantom. Both solid lines are shown for better comparison and are the same as in A. B also includes data obtained for the unshielded bent folded-end dipoles with the height of 33 mm and different fold values (black solid line). Dipoles were loaded by the full-size HS phantom.
phantom size causes a corresponding decrease of the optimal folded portion length, providing the best decoupling. Figure 5 shows similar data obtained for a pair of bent folded-end dipoles loaded by the HS phantom. For comparison, Figure 5B also includes data obtained for the unshielded bent folded-end dipoles with the height of 33 mm and different fold values. $S_{12}$ values for all of them measure $\sim -10 \text{ dB}$. Qualitatively, the results shown in Figure 5 are similar to those in Figure 4. Quantitatively, for the same height of the bent folded-end dipoles, the best decoupling is obtained with the longer folded portion. For example, while the best decoupling for straight folded-end dipoles with a height of 30 mm is obtained when the length of the folded portion measures between 10 mm and 15 mm, the bent folded-end dipoles with the same height have lowest $S_{12}$ values for lengths between 40 mm and 50 mm. To confirm our idea regarding additional inductive coupling produced by the capacitive bridge between the folded portion and the shield, we also numerically evaluated the imaginary part of the mutual impedance, $\text{Im}(Z_{12})$, for shielded and unshielded bent folded-end dipoles with three different lengths of the folded portion, ie 10 mm, 30 mm, and 60 mm. Supplementary Figure S1A shows results of the simulations. For all three unshielded pairs of dipoles $\text{Im}(Z_{12})$ was negative (capacitive coupling) and measured $\sim -11 \text{ } \Omega$. For shielded dipoles coupling changes from capacitive (negative) to inductive (positive) with an increase of the folded portion (Supplementary Figure S1A). 30-mm folded-end dipoles demonstrated a small capacitive coupling of $-0.72 \text{ } \Omega$. A pair of shielded bent dipoles (Figure 1D and Supplementary Figure S1B) demonstrated capacitive coupling with $\text{Im}(Z_{12})$ measuring $-5.5 \text{ } \Omega$. For a pair of straight dipoles (Figure 1C) measured $-3.6 \text{ } \Omega$. To verify that the mechanism of decoupling of adjacent bent folded-end dipoles is mainly determined by capacitive coupling of the folded portion closest to the shield, we also simulated a pair of folded dipoles shown in Supplementary Figure S1C. These dipoles had the same total length, fold, and height as 30-mm folded-end dipoles (Figure 5A), but one side was completely straightened, which prevents any coupling of this side to the shield. $S_{12}$ between the dipoles measured $-22.4 \text{ } \text{dB}$, which is very similar to the result obtained for a pair of 30-mm bent folded-end dipoles (Figure 5).

After calculating $S_{12}$ values for a pair of dipoles, we also numerically evaluated decoupling, $B_{1+}$, and SAR distributions for various eight-element dipole arrays loaded by the Duke voxel model. Decoupling was evaluated as an average ($S_{12}$) over all eight pairs of adjacent dipole elements. Figure 6A shows examples of four different array designs including the bent dipole array, straight dipole array, straight folded-end dipole array, and bent folded-end dipole array. Figure 6B shows corresponding central sagittal $B_{1+}$ maps. Finally, Figure 6C demonstrates transversal SAR maps cut through slices containing maximum SAR locations. Table 1 summarizes data for exemplary arrays shown in Figure 6A. The best decoupling of $-19.12 \text{ } \text{dB}$ was obtained for the 30-mm bent folded-end dipoles with a height of 33 mm (Figure 5). For comparison, ($S_{12}$) for the same array but without the RF shield measured only $-11.3 \text{ } \text{dB}$. Both unfolded shielded and unshielded dipole arrays demonstrate ($S_{12}$) of $\sim -12 \text{ } \text{dB}$. Table 1 also has data for the 10-mm straight folded-end dipole array, which provided the best decoupling for dipole elements of this type, ie ($S_{12}$) of $-16.6 \text{ } \text{dB}$. However, the 10-mm straight folded-end dipole array demonstrates lower Tx efficiency and worse decoupling than the 30-mm bent folded-end dipole array. For all arrays we also compared maximum local SAR values, Tx efficiency evaluated as $\langle B_{1+} \rangle / \sqrt{P}$, where $P$ is the RF power measured the array input, homogeneity evaluated as $\text{SD}/\langle B_{1+} \rangle$ (SD is the standard deviation), SEE, and average $S_{12}$ between adjacent dipole elements. As seen from the table, the 30-mm bent folded-end dipole array provided the lowest
maximum SAR value and the best SEE value. Placing the additional RF shield at the superior location of the head (Figure 2B) only slightly decreases the 30-mm bent folded-end array performance (Table 1), ie SEE is reduced by less than 4%. At the same time the shield increases the $B_1^+$ field at the superior location of the head. If for some reason this SEE reduction is critical, the superior RF shield can be removed. The unshielded bent array provided $\frac{h B_1^+}{\sqrt{P}}$ higher than the 30-mm bent folded-end array, but still substantially worse decoupling and the SEE value. The unshielded 30-mm bent folded-end array demonstrated the largest $\frac{h B_1^+}{\sqrt{P}}$, which was 1.16 higher than that of the shielded 30-mm bent folded-end array. SEE was similar to that of the shielded array, but decoupling was substantially worse (Table 1).

### 3.2 | Array performance

After construction, we evaluated the new array on a bench. As an example, Figure 7 shows $S_{12}$ matrices measured for the new eight-element TxRx 30-mm bent folded-end dipole array loaded by two human heads with substantially different sizes measuring 620 mm and 560 mm in circumference. Their size ratio measured about 0.9, ie very similar to the value we used for numerical $S_{12}$ evaluation (Figures 4 and 5). Since the larger head produces higher loading of the array, the measured decoupling was also better. However, even for the smaller head, all adjacent elements of the array were decoupled better than $-15$ dB. The average $S_{12}$ value between all adjacent elements measured $-16.6$ dB and $-18.4$ dB for the smaller and larger head, respectively. All nonadjacent dipoles were decoupled better than $-23$ dB.

After numerically evaluating and experimentally testing the array’s safety, we assessed the performance of the new dipole array in-vivo. Figure 8 demonstrates the performance of the bent folded-end dipole array including GE images and $B_1^+$ maps. As seen from the figure, the dipole array provides good longitudinal coverage. Averaged over a 130-mm transversal slab, $B_1^+$ measured 9.22 μT/kW.

Finally, Figure 9 shows the MP2RAGE in vivo images obtained using the eight-element $(1 \times 8)$ dipole array. The array provides whole-brain coverage without showing any signal intensity variations due to incomplete spin inversion.

| Array                                      | SAR* (W/kg) | $(B_1^+)/\sqrt{P}$ (μT/kW) | $(B_1^+)/\sqrt{P}$ (ratio) | $(B_1^+)/hB_1^+$ (SEE) | $(B_1^+)/hB_1^+$ (ratio) | $S_{12}$ 5 (dB) |
|--------------------------------------------|-------------|-----------------------------|-----------------------------|------------------------|--------------------------|-----------------|
| Bent dipole                                | 1.0         | 10.88                       | 1.0                         | 10.88                  | 1.0                      | -12.40          |
| Bent dipole (no RF shield)                 | 0.955       | 12.21                       | 1.12                        | 12.50                  | 1.15                     | 12.45           |
| Straight dipole                            | 0.918       | 8.15                        | 0.75                        | 8.50                   | 0.78                     | -12.07          |
| Straight dipole (no RF shield)             | 0.935       | 11.34                       | 1.04                        | 11.75                  | 1.08                     | -11.81          |
| 10-mm bent fold. Dipole                    | 0.574       | 11.96                       | 1.10                        | 15.79                  | 1.45                     | -13.22          |
| 30-mm bent fold. Dipole                    | 0.465       | 11.06                       | 1.02                        | 16.22                  | 1.49                     | -19.12          |
| 30-mm bent fold (no RF shield)             | 0.63        | 12.83                       | 1.18                        | 16.16                  | 1.49                     | -11.30          |
| 10-mm straight fold. Dipole                | 0.72        | 8.99                        | 0.83                        | 10.59                  | 0.97                     | -16.63          |
| 30-mm bent fold. Dipole (with the superior shield) | 0.483   | 10.85                       | 1.0                         | 15.62                  | 1.44                     | -18.74          |

*Maximum SAR averaged over 10 g of the tissue; calculated for 1 W of RF power at the array input.

#Averaged over 130-mm transversal slab (Figure 6B); calculated for 1 kW of RF power at the array input. $B_1^+$ was averaged only where tissue is present.

$Averaged over all eight pairs of adjacent dipole elements.

**TABLE 1** Numerical comparison of array Tx performance

**FIGURE 7** $S_{12}$ $8 \times 8$ matrices experimentally measured using the 30-mm shielded bent folded-end dipole array for two subjects with heads of different sizes, ie 620 mm and 560 mm in circumference. The numbering of dipole elements is shown on the right.
DISCUSSION

We developed and tested a new method of decoupling distantly located folded-end dipole elements of the human head TxRx array at 9.4 T. The mutual inductance between adjacent dipole antennas, which is produced by folding the dipoles and placing the RF shield close to the folded portion (Figure 1), compensates the intrinsic mutual capacitance and, thus, decouples the dipole elements. The mutual inductance value can be easily varied by adjusting the height of the dipoles (distance to the RF shield) and length of the folded portion. The presence of the “vertical” current (Figures 1A and 1B), ie, the current flowing through the capacitor $C_{eff}$, is critical for producing the mutual inductive coupling. It is also noteworthy that creating such a current on only one side of the dipole element (Figure 1B) is sufficient to compensate the intrinsic capacitive coupling and to decouple bent folded-end dipoles (Figure 5). However, decoupling of this type of dipole element requires a longer folded portion, ie, the larger effective capacitance, than obtained for a pair of straight folded-end dipole elements (Figures 1A and 4). Thus, as seen from Figures 4 and 5, the optimal decoupling value, ie, the lowest $S_{12}$, can be obtained for different combinations of height and folded length. Also, as seen from Table 1, both folded-end dipole arrays, ie, straight and bent arrays, provide significantly better decoupling than unfolded arrays. For example, the optimized 30-mm bent folded-end dipole array shows an averaged $S_{12}$ value (averaged over all eight pairs of adjacent dipoles) of $\sim 7$ dB better than that obtained for unfolded dipole arrays.

As seen from Table 1, the array Tx performance, ie, the maximum SAR, SEE (or $\langle B_1^+ \rangle / \sqrt{P}$), and homogeneity depend on the element geometry. For example, the 30-mm bent folded-end dipole array provides the best SAR and SEE values. After choosing the optimal element geometry based on a specific application, decoupling can be further optimized by alteration of the height of the dipoles (Figure 5), ie, 33 mm in our case. Since the required change of the height is rather small (Figures 4 and 5), it will not strongly modify the Tx performance. It is noteworthy that the geometry which we have chosen for our demonstration is not optimal for the Tx efficiency evaluated as $\langle B_1^+ \rangle / \sqrt{P}$. If for some applications this parameter is important, one can choose shielded bent folded-end dipoles with a smaller fold (eg, 10-mm fold), which delivers higher $\langle B_1^+ \rangle / \sqrt{P}$ but somewhat lower SEE (Table 1). Again, the decoupling can be then optimized by the corresponding adjustment of the dipole height (Figure 5). Shielded folded-end dipoles can still provide the optimal design of an array, which improves the decoupling (or SEE) in comparison to an unshielded dipole array. Thus, the process of optimization of element decoupling can be done independently of optimization of Tx performance. As seen from Table 1, poor decoupling (eg, unshielded arrays) does not

FIGURE 8 CST EM simulation model of the eight-element 30-mm bent folded-end dipole array loaded by the Duke voxel model, GE image, and $B_1^+$ map. The image and $B_1^+$ maps are shown for the central sagittal slice.

FIGURE 9 Sagittal in vivo 1 mm isotropic whole-brain MP2RAGE images obtained using the eight-element 30-mm bent folded-end dipole array.
necessarily cause a decrease in Tx performance. This fact does not diminish the importance of element decoupling for different Tx array designs. CP mode can tolerate relatively poor decoupling between adjacent channels, especially in the case of an eight-element array where each element has only two adjacent neighbors. This is not the case for a more complex 16-element two-row (2 × 8) array, where each element may have up to five adjacent neighbors. Good decoupling is even more important for the parallel transmission method, which uses very different RF shim sets of phases and amplitudes.

The RF shield is an essential component of our design and is commonly used in UHF head arrays and volume coils. The presence of the RF shield minimizes interaction of the RF coil with a surrounding structure, which may cause additional coupling, losses, and shift of the resonance frequency. Another example of usefulness of the RF shield for UHF head array designs is the recently developed method of \( B_0 \) shimming using local loops placed outside of the RF coil. These shimming devices require usage of a shielded RF coil. Finally, at 9.4 T the presence of the RF shield facilitates coupling of the array with the intrinsic TE mode of the head, which in turns improves \( B_1^+ \) and local SAR distributions. Evaluation of the RF shield’s influence on the Tx efficiency of the array is an important part of the development process. One of the most common effects of the RF shield is generation of eddy currents, which can reduce the RF magnetic field inside the head. For example, the unshielded 30-mm bent folded-end dipole array provides 16% higher \( \langle B_1^+ \rangle/\sqrt{P} \) than its shielded version. Also, the unshielded bent dipole array produces 12% higher \( \langle B_1^+ \rangle/\sqrt{P} \) than the shielded one. For the conventional surface loop design, this effect can be easily evaluated by measuring unloaded and loaded Q-factor values, ie \( Q_U \) and \( Q_L \), and making sure that their ratio, \( Q_U/Q_L \), exceeds a value of 3-4, which assures a sample loss domination regime. Since for radiative dipole antennas a large value of the Q-factor ratio does not necessarily prove the sample loss domination, numerical evaluation of the Tx performance is required.

One should also note that the major difference in Tx performance between the 30-mm bent folded-end array and bent dipole array is mainly in peak SAR values (Table 1). This is the case for both shielded and unshielded array versions. Tx efficiencies \( \langle B_1^+ \rangle/\sqrt{P} \) do not substantially differ, eg the bent 30-mm folded-end array has 2% better efficiency in the presence of the shield and 5% better efficiency without the shield than the bent dipole array. At the same time the SEE of the 30-mm bent folded-end array is 49% better than that of the bent array with the shield, and ~30% better without it. Moreover, all unfolded arrays (Table 1, Figure 6A) produce the maximum local SAR at the periphery of the head, ie in the ears, while the shielded 30-mm bent folded-end dipole array has its maximum SAR value located near the head center. This effect of local SAR reduction at the head periphery can be partially explained by the difference in the current distribution along the length of the loaded part of the dipoles. In the case of shorter bent dipoles (170-mm length), the current distribution changes from its maximum at the center to zero at the ends of dipoles. For longer folded-end dipoles (300 mm length), only the 170-mm central portion of the dipoles is loaded. As a result, variation in the current value along the loaded portion is less. A relative decrease in the maximum current at the dipole center causes a decrease in the peak SAR value. The effect of the influence of the dipole current distribution on the SAR and \( B_1^+ \) maps has been previously reported.\(^4\) In the presence of the shield an additional effect on SAR distribution is produced by coupling to the intrinsic TE mode of the head facilitated by the shielded bent folded-end dipole geometry.\(^21\) A decrease in the peak SAR value is another benefit of the bent folded-end geometry of the dipole Tx elements, which makes the array safer and ensures that larger heads placed into the relatively tight-fit RF coil do not cause a higher maximum SAR at the periphery.

Figures 8 and 9 show in vivo data obtained using the new eight-element dipole array. As seen from the figures, the new eight-element dipole array has good coverage of the whole brain, comparable to that of previously reported 2 × 8 surface loop array.\(^22\) It is notable that, at UHF, a 1 × 8 array consisting of longer surface loops circumscribing the head cannot provide for the full-brain coverage, and the substantially more complicated 16-loop 2 × 8 array combined with 3D RF shimming is required.\(^14\)\(^28\) Thus, the new dipole array allows for a significant simplification of the design without compromising Tx efficiency, ie \( \langle B_1^+ \rangle/\sqrt{P} \) and SEE. Obtaining improved whole-brain coverage at UHF using arrays with only eight elements is of great importance. There are currently almost 100 UHF MRI scanners in the world, and a majority of them are equipped with eight or fewer independent Tx channels. Thus, eight-element arrays providing adequate whole-brain coverage would simplify usage of parallel transmission.\(^33\)\(^34\)

5 | CONCLUSIONS

We present a novel method of decoupling dipole elements of the human head transceiver array at 9.4 T. Decoupling is realized by folding the dipoles and placing the RF shield close to the folded portion of the elements, which produces additional mutual inductive coupling between them. The method allows decoupling of distantly located dipole antennas without the necessity of changing their geometry or placing additional electronic circuits between them. Numerical comparison of the new array with various unshielded dipole arrays shows that the RF shield improves decoupling and SEE. We constructed and evaluated the single-row array with eight shielded 30-mm bent folded-end dipole elements. The array provides good decoupling with an average \( S_{12} \) value between adjacent elements of –16.6 dB even for a small head, which is substantially better than that obtained for the unfolded dipole arrays. Finally, we demonstrated that the full-brain coverage can be obtained with just eight dipole Tx elements, which is not feasible using eight surface loops.
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