Unshielded bent folded-end dipole 9.4 T human head transceiver array decoupled using modified passive dipoles

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Purpose: To develop an unshielded dipole transceiver array for human head imaging at 9.4 Tesla and to improve decoupling of adjacent dipole elements, a novel array design with modified passive dipole antennas was developed, evaluated, and tested.

Methods: The new array consisted of 8 bent folded-end dipole elements placed in a single row and surrounding the head. Adjacent elements of RF transceiver arrays are usually decoupled by introducing circuits electrically connected to elements. These methods are difficult to use for dipole arrays because of the distant location of the adjacent antennas. A recently developed decoupling technique using passive dipoles is simple and does not require any electrical connection. However, common parallel passive dipoles can produce destructive interference with the RF field of the array itself. To minimize this interference, we placed the passive dipoles perpendicularly to the active dipoles and positioned them at the ends of the array. We also evaluated the effect of different passive dipoles on the array transmit performance. Finally, we optimized the array transmit performance by varying the length of the dipole folded portion.

Results: By rotating the passive dipoles 90° and moving them toward the ends of the array, we minimized the destructive interference to an acceptable level without compromising decoupling and the transmit efficiency.

Conclusion: While keeping the benefits of the passive dipole decoupling method, the new modified dipoles produce substantially less destructive interference with the RF field of the array than the common design. The constructed transceiver array demonstrated good decoupling and whole-brain coverage.

Keywords
decoupling of the transmit dipoles, folded-end dipole antenna, RF head array, transceiver array, ultra-high field MRI, whole-brain coverage

Nikolai I. Avdievich and Georgiy Solomakha contributed equally to this work.
Dipole antennas have recently been introduced and utilized as elements of transmit and receive RF arrays for imaging of the human body and head at ultrahigh field (≥7 Tesla [T]). Radiative dipoles are usually used for producing an RF field in the far-field regime, that is, at distances much larger than the wave length, \( \lambda \). In contrast, MRI RF coils commonly work in the near-field regime, that is, at distances \( \ll \lambda \). Usage of dipole antennas for ultrahigh field MRI became possible due to substantial shortening of the wave length in human tissue (~7 times), where \( \lambda \) is close to 100 mm at 9.4 T (\(^{1}\text{H} \) frequency of 400 MHz), which is substantially smaller than the size of a human body or even a head. These conditions no longer define a near-field regime, especially in the case of the human body. Although full-length dipole antennas (length of \( \lambda/2 \), ie, 375 mm in free space at 400 MHz) can be used for body imaging, they must be substantially shortened for use as elements of human head RF arrays. Such short dipoles placed around the head are often coupled to the level of ~−10 dB\(^{5,12} \), which may be insufficient for optimal transmit performance, especially in the case of multi-row arrays where each element is surrounded by more than 2 (eg, up to 5 in 2-row array) adjacent elements.

Each pair of adjacent elements of MRI RF arrays is commonly decoupled by introducing circuits electrically connected to both of them.\(^{13-20} \) These decoupling methods are difficult to use for dipole arrays due to distant location of the adjacent antennas. Recently, we suggested a novel method of decoupling adjacent dipole elements of transmit and receiver human head arrays.\(^{12} \) Decoupling was provided without any additional electronic circuits by simply folding the dipoles and using a cylindrical RF shield located close to the folded portions.\(^{12} \) The RF shield is often used to minimize interaction of the MRI RF coil with the surrounding magnet bore structure, which may cause additional coupling, losses, and a shift of the resonance frequency. At the same time, the RF shield can reduce the transmit efficiency and SNR. The RF shield can also facilitate eddy currents when a pulse sequence with a fast switching gradient is used. Thus, an unshielded RF array coil also provides an alternative design, which can improve SNR and transmit performance.

Alternatively, adjacent unshielded transmit dipoles can be decoupled using passive dipole antennas.\(^{9,21,22} \) In this method, passive dipoles, which are through-space electrically coupled to both adjacent transmit elements, are placed parallel and equidistantly between them without any physical connections. This technique is similar to previously reported methods of decoupling adjacent surface loops in the transmit array by additional resonance circuits\(^ {9,23-25} \) placed between the loops. This decoupling practice should be used with care because large passive elements may interact destructively with the RF field produced by the transmit array itself.\(^ {19} \)

Because previously reported\(^ {9,21,22} \) passive decoupling dipoles were about the same size as the dipoles of the transmit array, they also can potentially disturb the RF field. Thus, in our opinion the method of decoupling the dipole elements of human head transmit arrays using passive dipoles still requires further evaluation and optimization.

For human head imaging, the physical sizes of the dipole antennas need to be reduced to accommodate the size of the subject. Multiple studies suggested different designs of dipole antennas with smaller physical sizes while keeping full electrical length, that is, close to \( \lambda/2 \). Examples of such designs include segmenting transmit dipoles interconnected with lumped-element inductors\(^ {3} \) and changing the shape of conductors (eg, meander-end dipole,\(^ {5} \) “snake,”\(^ {26} \) “Egyptian axe”\(^ {27} \)). Recently we developed a novel dipole transceiver array design using bent folded-end dipole antennas.\(^ {11} \) Use of the bent folded-end dipoles in combination with an RF shield demonstrated improved longitudinal (along the magnet axis) coverage and transmit performance as compared to unfolded dipoles of the same physical size. Transmit performance was evaluated as \( B_{1}^{+}/\sqrt{pSAR_{10g}} \), where \( pSAR_{10g} \) is the peak value of the local specific absorption rate (SAR) averaged over 10 g of the human tissue. The transmit performance and coverage was mainly improved by the interaction of the shielded bent folded-end dipole array with the intrinsic TE mode of the head resonating close to 400 MHz.

In this work, we developed a novel decoupling method of the adjacent dipole elements of a human head transceiver unshielded array using modified perpendicular passive dipole antennas. We also showed that such passive antennas produce substantially less distortion of the transmit RF field than previously suggested parallel designs.\(^ {9,21,22} \) Finally, we constructed and evaluated an unshielded transceiver human head 9.4 T array consisting of 8 bent folded-end dipole elements decoupled using the new technique.

## 2 | METHODS

### 2.1 | Folded-end dipole antennas

Recently, we developed novel transceiver\(^ {11} \) and receive\(^ {10} \) array designs using folded-end dipole antennas. The general idea behind the folded-end dipole design as well as a comparison to the common “short” (shorter than the full-length of \( \lambda/2 \)) dipole is demonstrated in Figure 1A and 1B. Commonly, the resonance frequency of a short dipole antenna is decreased by connecting in a series the lumped-element inductors as presented in Figure 1A, which shows the alteration of the current and voltage as compared to
the full-length dipole. The inductors become a part of the resonance circuit and carry most of the central current. Thus, by shortening the physical length of the dipole, we “cut out” the central part of the dipole current and voltage distributions. This implies that the central portion of the distribution, which is shown between two dash lines in Figure 1A, is now “hidden” within the relatively small inductors and does not reach the sample. As a result, the “loaded” portion of the distribution of a short dipole has large voltage values at both ends and a very nonuniform current distribution with the sharp maximum at the center of the dipole. By the “loaded” portion of the current distribution, we imply a part of the distribution that produces an RF field reaching the sample and generating eddy currents inside it. Alternatively, by folding the dipole and moving its ends away from the sample (Figure 1B), we load only the central portion of the dipole, which has lower voltage values and a more uniform current distribution. The expected effects of the modified current and voltage distributions include an extension of the RF magnetic field in the longitudinal direction and a decrease of the pSAR. Both effects are verified below.

2.2 | Modification of the passive decoupling dipole antennas

Coupling between two parallel dipole antennas produces two modes, that is, an antiparallel mode (with currents flowing in opposite directions) and a parallel mode (with currents flowing in the same direction). Figure 1C shows the frequency offsets of both modes. For simplicity, the figure depicts two well-separated modes, whereas in reality a weak coupling between two dipoles may only broaden the resonance line. This fact, however, does not affect the explanation and physics behind the effect. Because of capacitive coupling (through the electric field) between the dipoles, the parallel mode resonates at a higher frequency (Figure 1C). The lower antiparallel mode produces no RF field in the center between the dipoles. Therefore, a common straight passive decoupling dipole placed parallel and equidistantly to transmit dipole elements (Figure 1D, top) interacts only with the parallel mode. Therefore, a passive dipole resonating at a frequency higher than that of both modes shifts the frequency of the parallel mode down without any effect on the antiparallel mode. By
changing the resonance frequency of the passive dipole, the frequency of the parallel mode can be adjusted to the same value as the frequency of the antiparallel mode. Degeneracy of the two modes corresponds to decoupling of the transmit dipole elements. As mentioned above and further investigated below, relatively long (i.e., similar to the length of transmit dipoles) passive dipoles positioned parallel to adjacent transmit dipoles interacts with the RF field of the array. To minimize this interaction, we suggest modifying the passive dipole by turning it by 90°, as shown in Figure 1D (bottom). To provide strong capacitive (electric) coupling between all three dipoles, the modified passive dipole is moved to the very end of the transmit dipoles, where the electric field has its maximum, and is bent as shown in Figure 1D. Positioned in such a way, the passive dipole interacts only with the antiparallel mode. To shift up the resonance frequency of the antiparallel mode, the dipole must resonate at a frequency lower than that of both modes (Figure 1D, bottom). By changing the resonance frequency of the modified perpendicular passive dipole, the resonance frequency of the antiparallel mode can be adjusted without any effect on the parallel mode. Again, degeneracy of the two modes corresponds to decoupling of the transmit dipole elements.

Coupling between the passive dipole and two active dipoles can be also analyzed using the Kirchhoff equations. The matrix for the Kirchhoff equations describing the setup consisting of a pair of transmit active dipoles and a passive decoupling dipole is given by

\[
\begin{pmatrix}
V_1 \\
V_2 \\
0
\end{pmatrix} =
\begin{pmatrix}
Z_{11} - jX C & \pm jX & 0 \\
-jX C & Z_{22} & jX \\
\pm jX & jX & Z_{33}
\end{pmatrix}
\begin{pmatrix}
I_1 \\
I_2 \\
I_3
\end{pmatrix},
\tag{1}
\]

where \(-jX C\) is reactive capacitive coupling between transmit dipoles, and \(jX\) is reactive coupling between each transmit dipole and the passive dipole. \(Z_{ii}, V_i,\) and \(I_i\) are impedances, voltages, and currents corresponding to both active dipoles and the passive dipole. It is important that in the case of the common parallel passive dipole, couplings between the passive dipole and two active dipoles have the same signs. In contrast, couplings between the perpendicular passive dipole and two active transmit dipoles have opposite signs. Solving for \(V_1\) and \(V_2\) we obtain

\[
\begin{pmatrix}
V_1 \\
V_2
\end{pmatrix} =
\begin{pmatrix}
Z_{11} + \frac{X^2}{Z_{33}} & -jX C \pm \frac{X^2}{Z_{33}} \\
-jX C \pm \frac{X^2}{Z_{33}} & Z_{22} + \frac{X^2}{Z_{33}}
\end{pmatrix}
\begin{pmatrix}
I_1 \\
I_2
\end{pmatrix},
\tag{2}
\]

In both Equations (1) and (2), the positive sign in the off diagonal elements corresponds to the model with the parallel passive dipole and the negative sign to the model with the perpendicular passive dipole. Decoupling of active transmit
dipoles corresponds to minimization of off diagonal elements, which occurs when \(X_C = \text{Im} \left( \frac{X}{Z_{33}} \right)\) for the common parallel passive dipoles and \(X_C = -\text{Im} \left( \frac{X}{Z_{33}} \right)\) for the new perpendicular passive dipoles. Here \(\text{Im}\) implies the imaginary part of a complex value. Assuming that \(Z_{33} = R + jA(\omega)\), where \(R\) and \(A(\omega)\) are the resistive and reactive components of the impedance of passive dipoles, we obtain that \(\text{Im} \left( \frac{X}{Z_{33}} \right) = -j \frac{A(\omega)X^2}{R + A(\omega)^2} \approx -j \frac{\omega X}{\omega A(\omega)}\). Here we assumed that \(A(\omega)^2 > R^2\) because passive dipoles resonate off the Larmor frequency (Figure 1). Finally, decoupling conditions correspond to

\[
X_C = \mp \frac{X^2}{A(\omega)} \quad \text{or} \quad A(\omega) = \mp \frac{X^2}{X_C},
\tag{3}
\]

for the parallel and perpendicular passive dipoles. Thus, one can decouple transmit dipoles simply by adjusting the resonance frequency of the passive dipoles until the conditions of Equation (3) are satisfied. From Equation (1), we can also calculate the current induced in the parallel and perpendicular passive dipoles

\[
I_3 \approx -\frac{X}{A(\omega)} (I_1 \pm I_2) = \pm \frac{X_C}{X} (I_1 \pm I_2).
\tag{4}
\]

Because both passive dipoles are shorter than \(\lambda/2\), \(A(\omega)\) is negative below and positive above the dipole resonance frequency. Thus, the parallel passive dipoles must resonate above 400 MHz to ensure that \(A(\omega)\) at 400 MHz is negative. In contrast, the perpendicular passive dipoles must resonate below 400 MHz to produce positive \(A(\omega)\) at 400 MHz.

2.3 | Electromagnetic simulations

Before constructing the array, we optimized and evaluated the new array design using numerical electromagnetic (EM) simulations. EM simulations were performed using CST Studio Suite 2019 (Dassault Systèmes, Vélizy-Villacoublay, France) and the time-domain solver based on the finite-integration technique. In simulations, we used four different models, including a head and shoulder (HS) phantom, which was constructed to match tissue properties (\(\varepsilon = 58.6, \sigma = 0.64\) S/m) at 400 MHz; a cylindrical phantom (160 mm: diameter, 400 mm: length) with the same electromagnetic properties as the HS phantom; and virtual family multi-tissue voxel models, “Duke” and “Ella,” cropped at the chest level. For all models, we used an isotropic resolution of 2 mm.

First, to demonstrate the effect of the change in the current distribution resulting from folding the dipoles, we simulated the \(B_1^+\) field for three single dipoles (Figure 2A), including the full-length, short, and folded-end dipoles all loaded by
the cylindrical phantom. Both the short and folded-end dipoles measured 170 mm in $z$ direction (along the axis of the cylinder). The total length of folded-end dipole was 300 mm. Length of the full-length dipole measured 350 mm. All dipoles were modeled using 1.5-mm copper annealed wire. In addition, we simulated the loaded portions of the current and voltage distributions for both the short and folded-end dipoles. It is difficult to obtain numerically a smooth distribution of the current and voltage along the conductor due to relatively coarse meshing. Therefore, we instead simulated amplitudes of corresponding projections of the RF magnetic ($H_x$) and electric ($E_y$) fields at a very short distance to the wire (5 mm). Values of these RF field components are proportional to the current and voltage.

In the next step, we numerically evaluated whether the addition of passive dipoles$^{9,21,22}$ produces any distortion of the RF magnetic field of the transmit array due to destructive interference. To simplify the model, we used the cylindrical phantom and a pair of the folded-end dipoles positioned at a 20-mm distance from the phantom. Figure 3A depicts 4 EM simulation models for pairs of the transmit folded-end dipoles with and without passive decoupling dipoles. We simulated three types of passive dipoles (Figure 3A), that is, the common straight parallel passive dipole located at the same distance from the phantom as the transmit dipoles (decoupling 1), common straight parallel passive dipole moved further away from the phantom to the level of the folded portion (decoupling 2), and modified perpendicular passive dipole (decoupling 3). Decoupling 2, that is, the straight passive dipole moved away from the sample, was suggested previously$^{30}$ as a method to optimize decoupling and minimize destructive interference. The straight parallel passive dipoles (decoupling methods 1 and 2) had the same length as the active dipoles, that is, 170 mm. This length was chosen to maximize coupling between passive and active dipoles. Further increase of the length does not enhance coupling between passive and active dipoles but may increase interference with the RF field of the array as well as coupling between the passive dipoles themselves. The modified perpendicular passive dipoles measured 200 mm in length. To create capacitive coupling between the active and passive dipoles, the perpendicular passive dipole was folded (Figure 1D). Also to maximize this coupling, lengths of the folded portions of active and passive dipoles were the same and measured 30 mm. All passive and active dipoles were modeled using 1.5-mm copper annealed wire. All $B_1^+$ maps were obtained with a phase shift of 45º between two channels.
To confirm the results obtained for the simple model consisting of the cylindrical phantom and a pair of dipoles, we simulated several single-row 8-element (1 × 8) bent folded-end dipole arrays decoupled using all three methods shown in Figure 3A. Figure 4 (first column) displays four EM simulation models of 1 × 8 bent folded-end dipole arrays with and without decoupling, all loaded by the HS phantom. It is noteworthy that positions of perpendicular passive dipoles (decoupling 3) alternate in the $z$ direction (along the axis of the array). Placing all of them at the same end caused strong coupling, which make this decoupling method nearly impossible to use. All the transmit folded-end dipole elements measured 170 mm in the $z$ direction (along the phantom axis) with a total length of 300 mm. In addition, we tested all three decoupling methods for an array of unfolded dipoles (Figure 4, last column) of the same length, that is, 170 mm. Parallel straight and perpendicular bent passive dipoles measured 170 mm and 200 mm in length, respectively. All dipoles were modeled using 1.5-mm copper annealed wire. All passive dipoles were tuned in the postprocessing step by connecting inductors to the ports placed at the dipole centers (Figure 4). Their resonance frequency was adjusted to minimize coupling ($S_{12}$) between corresponding adjacent transmit dipole elements.

In the next step, we optimized the performance of the 1 × 8 bent folded-end dipole array loaded by the Duke and Ella voxel model. Dipole elements were placed on the surface of a fiberglass array holder with the wall thickness of 3 mm as shown in Figure 5A. The internal diameter of the holder was the same as previously reported$^{11}$ and measured 200 mm in width (left–right) and 230 mm in height (anteroposterior). To better fit a human head, the holder was tapered at its superior location (Figure 5A), where it measured 155 mm in width and 185 mm in height. All simulated arrays did not have an RF shield. A large copper cylinder (640 mm in diameter and 1600 mm in length), which mimicked the RF shield of the gradient coil, was included in all simulations. All matching and tuning circuits were taken into account during postprocessing. In addition, we evaluated the change in the array transmit performance when the total size of the array in the...
transversal plane was increased by 20 mm and 40 mm; that is, each dipole element was moved away from the head by 10 mm and 20 mm, respectively. First, to simplify the model and speed up the simulation process, the optimization was performed for arrays without decoupling. Finally, the best-performing array design was also simulated, including the modified decoupling passive dipoles (Figure 5A). To increase $B^+_1$ field at the superior location of the head and thus improve the longitudinal coverage, we also added a flat elliptical RF shield (175 mm × 140 mm) placed 30 mm away from the head perpendicularly to the z axis (the axis of the cylinder), as shown in Figure 5A.

**FIGURE 4** EM simulation models of 1 × 8 bent folded-end dipole arrays without decoupling (A) as well decoupled using three methods, that is, the common parallel passive dipoles (B) (decoupling 1), common parallel passive dipoles moved away from the sample (C) (decoupling 2), and modified perpendicular passive dipoles (D) (decoupling 3). EM simulation models of 1 × 8 bent (unfolded) dipole arrays without decoupling (E) as well as using three types of decoupling, that is, 1 (F), 2 (G), 3 (H). All arrays are loaded by the HS phantom. All transmit dipoles are shown in black, and the passive dipoles are shown in red. All ports are positioned at the centers of dipoles and shown in red. HS, head and shoulder.
The entire CST model of the 8-element dipole array loaded by a human head voxel model included 60 to 70 million mesh cells. On our computer (2 × Intel Xeon Gold 6142 CPU@2.6 GHz (Santa Clara, CA, USA) with 3 GPU NVIDIA Tesla V100 PCIE-32 GB (Santa Clara, CA, USA) and 192 GB RAM), a single simulation required about 4 to 5 hours.

To characterize the array performance, we evaluated the average $B_1^+$ value, and homogeneity calculated as $\text{SD}(B_1^+)$, $B_1^+$ was averaged over a 130-mm transversal slab, which included the majority of the human brain. In addition, we evaluated the transmit efficiency, both as $\langle B_1^+ \rangle / \sqrt{P}$, where $P$ is forward RF power measured at the array input, and as $\langle B_1^+ \rangle / \sqrt{\text{SAR}_{10g}}$, that is, the safety excitation efficiency (SEE). $B_1^+$ field profiles and local SAR $10g$ (averaged over 10 g of tissue) maps were calculated for 1 W of forward stimulated RF power at the array input and then compared with experimentally measured data. Optimization of the array geometry was performed mostly by varying the length of the folded portion of the dipoles. The optimal length of folded portion was chosen based on the best SEE value.

### 2.4 Array construction

After EM modeling, we constructed the new optimized 1 × 8 transceiver unshielded dipole array (Figure 5B). The geometry of the array holder and all elements were the same as in EM modeling (Figure 5A). The array consisted of 8 30-mm bent folded-end dipole antenna elements (the length of the folded portion was 30 mm). The transceiver dipoles measured 35 mm in height and 170 mm in length in $z$ direction. Total length of the dipole wires measured 300 mm. Modified perpendicular passive dipoles measured ~ 200 mm in the total length. All dipoles were constructed using 1.5-mm tinned copper wire. Figure 5C shows a schematic of a single dipole element, including the tuning inductors, matching capacitor (Johanson Corp., Boonton, NJ), cable trap, and home-built transmit/receive (T/R) switch circuit. T/R switches were connected to each dipole element and located inside the array holder to minimize cable losses. Low-noise preamplifiers (LNA) (WanTcom, Chanhassen, MN) were incorporated into the T/R switch circuits. Based...
on EM simulation data, we used variable capacitors of 20 pF. This was sufficient to provide matching on phantoms and various human heads. 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 (300 mm) was close to the full length (λ/2) of the dipoles. To bring the resonance frequency of the shorter (~ 200 mm) passive dipoles below 400 MHz, an inductor (~ 200 nH) was connected in a series with the dipole (Figure 1D). During construction, these inductors were adjusted to minimize coupling between adjacent active dipoles. All inductors were handmade using 1.0-mm copper wire. A shielded cable trap was introduced at the input of each receiver dipole element to cancel the common mode excited on the outer surface of the cable braid. Because we did not plan using fast pulse sequences, the superior shield (end-cap) was not slotted. However, to minimize eddy current, we used relatively thin copper foil of 25 μm constructed of a kapton laminate (AKAFLEX, Krempel GmbH, Vaihingen/Enz, Germany). During transmission, the array was driven in the quadrature circular polarized (CP) mode, which in our case corresponded to a 45º phase shift between adjacent dipoles. For this purpose, we fabricated an 8-way Wilkinson splitter with corresponding phase shifters constructed of cables and incorporated into the splitter box.

2.5 | Experimental evaluation of the array performance

Before in vivo measurements were obtained, the dipole array was evaluated both on a bench and in the scanner and numerically simulated according to the safety procedure developed in our lab.34 Bench evaluation of the array included measurements of the entire $S_{12}$ matrix (8 × 8) using a network analyzer E5071C (Agilent Technology, Santa Clara, CA, USA). In all bench measurements, we used the HS phantom described above. After initial experiments with phantoms, the array coil was adjusted on a human head. No retuning was required when the coil was used on different subjects.

Human subjects participated in the study after giving signed informed consent according to procedures approved by the local institutional ethics committee. All phantom and in vivo data were acquired on a Siemens Magnetom (Siemens Healthineers, Erlangen, Germany) 9.4 T whole-body human MRI scanner. The scanner is equipped with 8 transmit channels of 1 kW RF power each. $B_1^+$ maps were obtained using the 3D actual flip angle imaging sequence35 (FOV: 244 × 244 × 100 mm$^3$, voxel size: 1.8 × 1.8 × 5 mm$^3$, TR$_1$/TR$_2$: 20/100 ms, TE: 4 ms, flip angle: 60º). All $B_1^+$ maps are normalized to the RF power level at the coil input. The images were obtained with a 3D gradient echo sequence (TR/TE: 18/2 ms, FA: 8º, voxel size 2 mm isotropic, FOV: 240 × 240 × 192 mm). To evaluate the performance of the new dipole array in a typical ultrahigh field application, we also performed MP2RAGE measurements for the dipole array. MP2RAGE images were acquired with 1-mm isotropic resolution in about 11 min (other parameters: TR$_{rot$inv$} = 6/6000$ ms, FA$_{1/2} = 5º/9º$, T1$_{1/2}$: 800/2000 ms, 6/8 partial Fourier in partition direction, 10 ms time-resampled frequency-offset-corrected inversion.37

3 | RESULTS

3.1 | Folded-end dipole antenna

To demonstrate the effect of the change in the current distribution due to folding the dipoles, we simulated the $B_1^+$ field for three single dipoles loaded by the cylindrical phantom (Figure 2). Figure 2B shows coronal $B_1^+$ maps for three different dipole antennas: “full-length”, short, and folded-end. The position of the coronal slices is shown in Figure 2C. Figure 2D depicts longitudinal (parallel to the cylinder axis) plots of the $B_1^+$ field taken along the center of coronal slices. The $B_1^+$ value is normalized to the square root of the pSAR values. As seen in Figure 2D, folding the dipole produces an extension of the RF field in the longitudinal direction as compared to the short dipole. To confirm the hypothesis presented above in Figures 1A,B, we also simulated the current and voltage distribution along the loaded portions of the short and folded-end dipoles (Figure 2E). As seen in Figure 2E, the folded-end dipole provides a more uniform current and lower voltage along the dipole length. The folded-end dipole voltage distribution still demonstrates a small “jump” near the center due to the presence of small inductors (Figure 1B). This also implies that the total length of the dipole is slightly shorter than λ/2.

3.2 | Destructive interference of passive decoupling dipole antennas with the array transmit field

In the next step, we numerically evaluated whether the addition of different passive dipoles9,21,22 produces any distortion of the array RF field using the cylindrical phantom and a pair of folded-end active dipoles (Figure 3). All three types of decoupling demonstrated a sufficiently low cross-talk level between the active dipoles, that is, below −16 dB. Three passive dipoles, however, had different influences on the RF field distribution. Figure 3B shows $B_1^+$ maps calculated for all four models, including the model without decoupling. Finally, Figure 3C demonstrates ratios of corresponding $B_1^+$ maps to those obtained without decoupling. As seen in Figures 3B,C, the modified perpendicular passive dipole (decoupling 3) produces the smallest alteration of the transmit $B_1^+$ field.
To confirm results obtained for the cylindrical phantom, we compared the performance of several 1 × 8 bent folded-end dipole arrays modeled using all three decoupling methods. Figure 6A shows a comparison of simulated transversal $B_1^+$ maps obtained using decoupled 1 × 8 folded-end dipole arrays shown in Figure 4 (first column). Figure 6B shows ratios of $B_1^+$ maps shown in Figure 6A to those obtained without decoupling. Similar to the case of the cylindrical phantom (Figure 3), the common parallel passive dipoles placed close to the phantom (decoupling 1) produce the strongest destructive interference with the transmit RF field among all three decoupling methods. Moving the parallel passive dipoles away (decoupling 2) decreases the distortion. It is important that the destructive interference not only causes voids at the periphery but also decreases the RF field near the center of the phantom. Decoupling types 1 and 2 reduce the $B_1^+$ values near the phantom center by 21% and 16%, respectively.

Turning the passive dipoles by 90° (decoupling 3) substantially minimizes the alteration of the RF field both at the periphery and center and causes the reduction of the central $B_1^+$ by 3% only. In addition, Figure 6C shows central sagittal slices of the $B_1^+$ ratio maps for all three types of decoupling. Finally, Supporting Information Figure S1 shows five transversal slices of the $B_1^+$ ratio maps for all three types of decoupling. Slices are positioned at 30 mm and 60 mm away from the central slice in both directions spanning overall 120 mm (Figure 6C). The central slices (Figure 6B) are also included for comparison. As seen in Figure 6 and Supporting Information Figure S1, decoupling 3 demonstrates the smallest alteration of the RF field over the entire coil excitation volume.

In addition, we verified whether the introduction of passive decoupling dipoles also influences the transmit performance of the 1 × 8 unfolded bent dipole array (Figure 4, last column). Supporting Information Figure S2 shows a comparison of simulated $B_1^+$ maps and corresponding ratios obtained using the unfolded array with and without passive decoupling dipoles. Supporting Information Figure S3 shows five transversal slices of the $B_1^+$ ratio maps for all three types of decoupling. Similar to the folded-end dipole array, decoupling 3 produces the smallest alteration of the $B_1^+$ field. In addition to peripheral voids, the central $B_1^+$ value is reduced by 23%, 44%, and 2% for decoupling methods 1, 2, and 3, respectively. It is noteworthy that whereas quantitative effects of decoupling types 1 and 3 are similar for both folded-end and unfolded dipole arrays, decoupling 2 caused a substantially higher reduction of the central $B_1^+$ field for the unfolded array (Supporting Information Figures S2 and S3). To better

**FIGURE 6** (A) Central transversal $B_1^+$ maps obtained using 1 × 8 bent folded-end arrays shown in Figures 4B-D. (B) Ratios of corresponding $B_1^+$ maps shown in Figure 6A to that obtained without decoupling. (C) Central sagittal maps of the ratios of $B_1^+$ fields obtained using decoupled 1 × 8 bent folded-end arrays to that obtained by the array without decoupling (Figure 4A). Figure 6C shows positions of transversal slices chosen for evaluation of $B_1^+$ map ratios (Supporting Information Figures S1 and S3). All slice are separated by 30 mm.
understand the effect, we calculated the imaginary part of the off diagonal element of the impedance matrix, that is, $jX$, between an active transmit dipole and passive dipole both located at the posterior position (Figure 4) for four cases, that is, the folded-end dipoles and unfolded dipoles decoupled by methods 1 and 2. For method 1, $X$ measured $-22$ Ohm and $-24.8$ Ohm for the unfolded and folded-end dipoles, respectively. For method 2, $X$ measured $-11$ Ohm and $-18$ Ohm for the unfolded and folded-end dipoles, respectively. Thus, moving the parallel passive dipoles away from the sample (method 2) caused a substantially larger reduction of the absolute value of $X$ in the case of the unfolded dipoles. It is also interesting that in the case of decoupling 1, straight passive dipoles produce a smaller peripheral alteration of the transmit field of the unfolded array (Supporting Information Figures S2 and S3) than that of the field of the folded-end array (Figure 6 and Supporting Information Figure S1).

According to Equation (4), the amplitude of $I_3$ is proportional to the absolute value of $X_C/X$. Capacitive coupling, that is, $X_C$, between adjacent active dipoles located at the posterior position measured $-8$ Ohm and $-11$ Ohm for the unfolded and folded-end dipoles, respectively. Finally, for the amplitude of $I_3$ we obtain that $I_3 \sim 8/22 = 0.36$ and $I_3 \sim 11/24.8 = 0.44$ for unfolded and folded-end dipoles, respectively. Thus, current in the passive dipoles (decoupling 1) is $\sim 20\%$ higher in the case of folded-end array. Numerical evaluation of the current values showed a similar increase of the current in the passive dipoles for the case of folded-end dipole array.

### 3.3 Numerical and experimental evaluation of the dipole array design

In the next step, we optimized the performance of the $1 \times 8$ unshielded bent folded-end dipole array loaded by the Duke and Ella voxel models. Optimization was performed for arrays without decoupling. As it was shown previously, intrinsic coupling of $\sim -10$ to $-12$ dB between adjacent dipole elements of a head array does not cause a substantial decrease of the CP mode transmit efficiency or change in the local SAR distribution.**11,12** Figure 7 shows a comparison of different $1 \times 8$ bent and bent folded-end arrays all loaded by the Duke voxel model. Table 1 summarizes all numerically simulated data. Figure 7A displays sagittal $B_1^+$ maps obtained using different $1 \times 8$ dipole arrays. As seen in Figure 7A, increasing the fold size, which makes the current distribution along the loaded part of the dipole more uniform (Figures 1 and 2E), improves the longitudinal coverage by extending the RF field in both directions. Such a change in the current distribution also causes a decrease of pSAR. Maximum local SAR obtained for the bent dipole array is more than $50\%$ higher than that obtained for the $30$-mm bent folded-end dipole array (Table 1). An increase in the fold size from $30$ mm to $50$ mm only slightly ($\sim 1\%$) decreases the pSAR value (Table 1). Figures 7B,C demonstrate transversal SAR$_{10g}$ maps obtained using the $1 \times 8$ bent dipole array and $30$-mm bent folded-end dipole array for two slices (Figure 7A) intersecting the voxel model near the center of the dipoles.

**FIGURE 7** (A) EM simulated central sagittal $B_1^+$ maps obtained using different dipole arrays and the Duke voxel model. Single elements of corresponding dipole arrays are shown in Figure 7A in white. Transversal SAR$_{10g}$ maps obtained for two slices using the $1 \times 8$ bent dipole array (B) and $30$-mm bent folded-end dipole array (C). Positions of both slices are shown in Figure 7A.
(slice 1) and through the ears (slice 2). Although distributions in both central slices are very similar, maps from the slices going through the ears differ considerably. The bent dipole array produces substantially higher local SAR on the ears, that is, 0.947 W/kg versus 0.53 W/kg. Increasing the size of the bent array by 20 mm and 40 mm (by moving each dipole by 10 mm and 20 mm away from the head) decreases pSAR10g (Table 1) while maintaining its location, that is, at the ears. This fact implies that the high local SAR value near the surface of the sample is caused by a nonuniform current distribution, which has a sharp maximum near the dipole center (Figures 1 and 2E), rather than by the high conservative electrical field at the ends of unfolded dipoles. Increasing the size of the bent dipole array by 40 mm causes a drop of $\frac{B_{1+}}{\sqrt{\rho P}}$ by 5% and an increase of SEE by 9% (Table 1). A 40-mm increase in the size of the 30-mm bent folded-end dipole array causes a ~7% decrease of $\frac{B_{1+}}{\sqrt{\rho P}}$ and ~2% decrease of SEE. As seen in Table 1, the 50-mm bent folded-end dipole array provides the highest $\frac{B_{1+}}{\sqrt{\rho P}}$ and SEE values and the best homogeneity (SD/$\frac{B_{1+}}{\sqrt{\rho P}}$).

However, for the sake of comparison of the unshielded array and its shielded version,11 we decided to construct the 30-mm bent folded-end dipole array, which has only a ~1% lower SEE and ~4% lower homogeneity. A 30-mm bent folded-end dipole array with an RF shield was constructed and evaluated previously.11 After choosing the length of the fold, we also evaluated a change in the performance due to addition of the superior RF shield and modified perpendicular passive dipoles (decoupling 3). As seen in Table 1, the addition of the superior RF shield caused an increase of the $\frac{B_{1+}}{\sqrt{\rho P}}$ by 3.6% and SEE by 5.5%. Addition of the passive dipoles caused a decrease of $\frac{B_{1+}}{\sqrt{\rho P}}$ by 2.7% and SEE by 1.5%.

The $\frac{B_{1+}}{\sqrt{\rho P}}$ and SEE values obtained for the Ella voxel model (Figure 8A) were 5.5% and 1.5% lower than those obtained for the Duke voxel model, respectively. Figures 8B,C show simulated sagittal $B_{1+}$ maps and transversal SAR10g maps (cut through the location of the maximum SAR10g) obtained using the unshielded 30-mm bent folded-end dipole array and both the Duke and Ella voxel models. In addition, Figure 8D shows the ratio of the $B_{1+}$ field obtained using the

| Voxel Model | Array | SAR10g, W/kg | $\frac{B_{1+}}{\sqrt{\rho P}}$, $\mu$T/$\sqrt{kW}$ | SD/$\frac{B_{1+}}{\sqrt{\rho P}}$, μT | SEE | SEE Ratio | $P_{T}^{S}$, W |
|-------------|-------|-------------|------------------------|-----------------|-----|---------|-------------|
| Duke EM simulation (no decoupling) | Bent | 0.947 | 12.33 | 0.34 | 1.0 | 12.67 | 1.0 | 0.82 |
| | Bent (20 mm larger**) | 0.836 | 12.11 | 0.32 | 0.98 | 12.50 | 1.05 | 0.83 |
| | Bent (40 mm larger***) | 0.729 | 11.76 | 0.30 | 0.95 | 14.41 | 1.09 | 0.77 |
| | 10-mm bent folded-end | 0.618 | 12.94 | 0.29 | 1.05 | 16.46 | 1.30 | 0.86 |
| | 30-mm bent folded-end | 0.627 | 13.20 | 0.28 | 1.07 | 16.67 | 1.32 | 0.86 |
| | (20 mm larger**) | 0.618 | 12.99 | 0.25 | 1.05 | 16.52 | 1.30 | 0.87 |
| | (40 mm larger***) | 0.567 | 12.37 | 0.24 | 1.0 | 16.43 | 1.30 | 0.81 |
| | 50-mm bent folded-end | 0.621 | 13.24 | 0.27 | 1.07 | 16.80 | 1.33 | 0.88 |
| | 30-mm bent folded-end | 0.604 | 13.67 | 0.25 | 1.11 | 17.59 | 1.39 | 0.86 |
| (super. RF shield) | |
| Duke EM simulation (decoupling 3) | 30-mm bent folded-end | 0.591 | 13.31 | 0.25 | 1.08 | 17.31 | 1.37 | 0.81 |
| (super. RF shield) | |
| Ella EM simulation (decoupling 3) | 30-mm bent folded-end | 0.546 | 12.61 | 0.24 | 1.02 | 17.05 | 1.35 | 0.7 |
| (super. RF shield) | |
| HS phantom EM simulation (decoupling 3) | 30-mm bent folded-end | – | 13.53 | 0.29 | – | – | – | – |
| (super. RF shield) | |
| HS phantom experiment (decoupling 3) | 30-mm bent folded-end | – | 11.7 | 0.26 | – | – | – | – |
| (super. RF shield) | |

EM, electromagnetic; HS, head and shoulder; SAR, specific absorption rate; SEE, safety excitation efficiency.

* Averaged over 130-mm transversal slab; Calculated for 1 kW of RF stimulated power at the array input.

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

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

** The array size was increased by 20 mm by moving each dipole element 10 mm away.

*** The array size was increased by 40 mm by moving each dipole element 20 mm away.
unshielded 30-mm bent folded-end dipole array loaded by the Duke voxel model with and without addition of the decoupling passive dipoles (decoupling 3).

After numerical optimization and evaluation of the new array design, we constructed the 1 × 8 unshielded array consisting of the 30-mm bent folded-end dipoles. Figure 5D shows the S_{12} matrix experimentally measured for the array loaded by the HS phantom. As seen in the figure, the array is well decoupled. All adjacent elements of the array were decoupled better than −15 dB with an average decoupling value between them equal to −18.1 dB. All nonadjacent dipoles were decoupled better than −22 dB. We also experimentally evaluated the transmit performance of the array using the HS phantom. Averaged over a 130-mm transversal slab, \( B_{1+} \) measures 11.7 μT/√kW (Table 1), which is 11% higher than that obtained using the shielded version of the array.\(^1\)

After experimentally testing the array safety in accordance with the institute’s internal rules,\(^3\) we assessed the transmit performances in vivo. Figure 9 shows in vivo data including sagittal and transversal images and \( B_{1+} \) maps obtained for a male subject with a relatively large head (620 mm in circumference). As seen in the figure, the dipole array provided good coverage over the entire brain. Averaged over a 130-mm transversal slab, which includes the entire brain, \( B_{1+} \) measures 9.3 μT/√kW. This is very similar to the value obtained using the shielded version of the array for the same subject, that is, 9.2 μT/√kW.\(^1\)

Finally, Supporting Information Figure S4 shows a few exemplary sagittal, coronal, and transversal isotropic MP2RAGE images acquired using the unshielded 30-mm bent folded-end 1 × 8 dipole array decoupled by method 3 and the same subject as in Figure 9. Driven in the CP mode, the array demonstrates excellent coverage without any additional RF shimming of the transmit field.

**FIGURE 8** (A) EM simulations models of the 30-mm bent folded-end dipole array including the superior RF shield and perpendicular passive decoupling dipoles loaded by the Duke and Ella voxel models. (B) Central sagittal \( B_{1+} \) maps obtained using models shown in Figure 8A. (C) Transversal SAR_{10g} maps cut through positions of peak SAR_{10g} values obtained using models shown in Figure 8A. Positions of both transversal slices are shown in Figure 8B by white dashed lines. (D) Sagittal, coronal, and transversal distributions of the ratio of the \( B_{1+} \) maps obtained using the 30-mm bent folded-end dipole array (Figure 8A) and the Duke voxel model with and without decoupling (decoupling 3). Position of the transversal slice is shown on the sagittal map.
DISCUSSION

We developed and tested a modified version of the previously developed\textsuperscript{9,21,22} passive decoupling dipole elements. Although their relative simplicity and the absence of any electrical connection make these elements appealing, the common passive parallel dipoles are shown to produce a strong destructive interference with the transmit field of the folded-end dipole array (Figures 6 and Supporting Information S1). It is important that destructive interference is not only limited to peripheral voids but also results in a substantial (~20%) decrease of the $B^+_1$ field near the center. Moving the common parallel passive dipoles away from the subject\textsuperscript{30} decreases the destructive interference. However, this still produces peripheral voids and decreases the central RF field. By rotating the passive dipoles by 90° and moving them toward the ends of the active dipoles, we minimized the destructive interference to an acceptable level (Figures 6, 8D and Supporting Information S1). The final version of the unshielded bent folded-end dipole array demonstrates good decoupling (Figure 5D) and full-brain coverage in vivo (Figures 9 and Supporting Information S4).

We also would like to emphasize the fact that the effect of destructive interference between active and parallel passive dipoles is not specific to the case of the folded-end dipole array design. As seen in Supporting Information Figures S2 and S3, 1 $\times$ 8 arrays of unfolded dipoles decoupled by methods 1 and 2 show peripheral voids and a strong decrease of the central $B^+_1$ field, especially in the case of decoupling 2. As indicated above, moving the passive dipoles further away from the sample (decoupling 2) causes the twofold decrease of the absolute value of $X$ and a corresponding increase of the current amplitude in the passive dipoles (Equation (4)). At the same time, the RF field produced by these moved passive dipoles most likely does not decrease at the same rate. As seen in Table 1, an increase of the unfolded array size by 40 mm decreases the average $B^+_1$ field by only ~5%. As a result, the stronger current induced in the passive dipoles produces stronger destructive interference.

After optimization of the fold length, we chose the 30-mm bent folded-end design, which produces only a ~1% lower SEE than the optimal 50-mm bent folded-end design. This choice, however, allows us to compare the results with a previously reported shielded version of the array.\textsuperscript{11} Although the transmit efficiency measured for the unshielded array using the HS phantom was 11% higher than that of the shielded array, in vivo measurements for the same male subject demonstrated very similar values for both unshielded and shielded arrays,\textsuperscript{11} that is, 9.3 $\mu$T/$\sqrt{\text{kW}}$ and 9.2 $\mu$T/$\sqrt{\text{kW}}$ averaged over the 130-mm transversal slab. The lower influence of the RF shield on the transmit efficiency in vivo can be explained by a sensitivity of the dipole array to loading. A decrease in the size of the sample (or increase of the distance to the sample) causes a corresponding decrease in the dipole array transmit efficiency. For example, our simulations of the
unshielded 30-mm bent folded-end array loaded by the Ella voxel model demonstrated a 5.5% decrease in the transmit efficiency (Table 1) as compared to the Duke voxel model, which has the substantially larger head. Presence of the RF shield further decreases loading and causes an even larger decrease of the transmit efficiency obtained for smaller heads. For example, a female subject with a smaller head (560 mm in circumference) demonstrated an 11% decrease of the experimentally measured transmit efficiency as compared to that obtained for the male subject with a larger head (620 mm in circumference). Thus, the male subject, who loaded the array very strongly, provided relative independence of array transmit efficiency on shielding. A smaller size head (or a phantom), which provides less loading, has a stronger dependence on shielding. Therefore, the absence of the RF shield, which produces higher loading of the array elements, decreases the sensitivity of the dipole array transmit efficiency to a change in the head size.

Most of the simulations shown in Table 1 were performed without modified passive dipoles, which were introduced only for the final version of the array. Better decoupling did not cause a substantial improvement of the array CP mode transmit performance (Table 1). This fact, however, does not diminish the importance of using decoupling methods while constructing a transceiver array. The CP mode performance was shown to be relatively independent of the level of decoupling in the case of a single-row 8-element array, in which each element has only two adjacent neighbors. This is not the case for a more complex 16-element 2-row arrays, in which each element may have up to five adjacent neighbors.

An important property of the novel folded-end dipole design, that is, a more uniform current distribution as compared to the unfolded short dipole design (Figures 1A,B, 2E), yields to an improvement in the longitudinal coverage in vivo (Figure 7A). Homogeneity (SD(\(B_1^+\))/\(\sqrt{\text{pSAR}}\)) improved from 0.34 to 0.25 for the bent array design and the final 30-mm bent folded-end array design, respectively (Table 1). A more uniform current distribution produced by the folded-end dipoles also minimizes the pSAR\textsubscript{10g}. In comparison to the bent folded-end design, the bent dipole array produces high local SAR at the ears, which limits the scanned power. Alternatively, the size of the bent dipole array can be increased to minimize the peak local SAR value. This increase, however, has to be sufficiently large, which will cause a corresponding decrease in the transmit efficiency. As seen from Table 1, increasing the size of the array by 40 mm causes a ~30% decrease in the pSAR value but still produces high SAR at the ears. A similar effect of high peak local SAR due to nonuniform current distribution (high current at the center of dipoles) was also demonstrated at 10.5 T for human head unfolded dipole arrays. In their work, the authors bent dipoles, introducing a “bump” at the center to move the central part of the conductor with the largest current distribution away from the subject. This, however, should inevitably decrease loading and, thus, the transmit efficiency. In comparison to “bump” method, making the current distribution along the dipole length more uniform increases loading and provides higher transmit efficiency and SEE as compared to dipole structures with nonuniform current distribution, for example, bent dipoles (Table 1). In addition, this produces a substantial improvement in the longitudinal coverage (Figure 7A). The effect of the longitudinal coverage improvement due to the alteration of the current distribution was also demonstrated earlier for the human head array constructed using meander-end dipole antennas. In this design, the electrical length of the dipole was increased by placing 2 meanders at both ends of the antenna. All the dipoles (including meanders) were placed on the cylindrical surface at the same distance to the sample. Because of this, the meander dipoles may cause a substantial shift of the resonance frequency for larger heads due to higher voltage (electric field) at the dipole ends. In our design, this effect is reduced as a result of moving both dipole ends away from the sample.

Finally, similar to the conclusion of the previous work, which describes use of passive resonant loops for decoupling of surface loop arrays, the geometry of passive dipoles can be selected using the following criteria. Most importantly, the physical size and geometry of passive dipoles has to be chosen to produce minimal interference with the RF field of the active dipoles. In addition, one needs to maximize coupling (ie, \(X\)) between the active and passive dipoles. This allows moving the resonance frequency of passive dipoles further from the Larmor frequency (Equation 3) and minimizing the current induced in the passive dipoles (Equation 4), which also decreases the interference between RF fields produced by passive and active dipoles. However, it is difficult to provide a general recipe for the best passive dipole design, which must be selected individually for each specific transmit array design.

5 | CONCLUSION

We developed a modified passive dipole design for decoupling transceiver human head dipole arrays. Although keeping all the benefits of the common decoupling method using parallel passive dipoles, that is, a simplicity of the design and the absence of electrical connection, the new perpendicular decoupling antennas produce substantially less destructive interference with the RF field of the array than the common design. We used the new decoupling method to construct a single-row 8-element bent folded-end dipole transceiver array for human head imaging at 9.4 T. The array demonstrated good decoupling, whole-brain coverage, and good transmit performance. The resulted SEE (\(\langle B_1^+ \rangle / \sqrt{\text{pSAR}}\)) was improved by ~30% as compared to the unfolded dipole array.
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SUPPORTING INFORMATION

Additional Supporting Information may be found online in the Supporting Information section.

FIGURE S1 Ratios of transversal $B_1^+$ maps obtained using decoupled 1x8 30-mm folded-end dipole arrays shown in Figures 4B (A), 4C (B), 4D C) to that obtained by the array without decoupling (Figure 4A). Positions of transversal slices shown in Figure 6C. Three ratio maps shown in centers of Supporting Information Figures S1A, S1B, and S1C are the same as shown in Figure 6B

FIGURE S2 (A) Transversal $B_1^+$ maps obtained using 1x8 bent unfolded dipole arrays shown in Figure 4F-H. (B) Ratios of transversal $B_1^+$ maps shown in Figure S1A to that obtained without decoupling (Figure 4E). (C) Ratios of sagittal $B_1^+$ maps obtained using decoupled 1x8 unfolded arrays to that obtained without decoupling (Figure 4E)

FIGURE S3 Ratios of transversal $B_1^+$ maps obtained using decoupled 1x8 bent (unfolded) dipole arrays shown in Figures 4F (A), 4G (B), 4H C) to that obtained by the array without decoupling (Figure 4E). Positions of transversal slices shown in Figure 6C. Three ratio maps shown in centers of Supporting Information Figure S3A-C are the same as shown in Supporting Information Figure S2B

FIGURE S4 Exemplary sagittal (A), coronal (B), and transversal (C) in-vivo 1-mm isotropic whole-brain MP2RAGE images obtained using the 1 x 8 30-mm bent folded-end dipole array for the same subject as in Figure 9

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