MRI at ultra-high field (UHF, ≥7 T) provides a natural strategy for improving the quality of X-nucleus magnetic resonance spectroscopy and imaging due to the intrinsic benefit of increased signal-to-noise ratio. Considering that RF coils require both local transmission and reception at UHF, the designs of double-tuned coils, which often consist of several layers of transmit and receive resonant elements, become quite complex. A few years ago, a new type of RF coil, ie a dipole antenna, was developed and used for human body and head imaging at UHF. Due to the mechanical and electrical simplicity of dipole antennas, combining an X-nucleus surface loop array with 1H dipoles can substantially simplify the design of a double-tuned UHF human head array coil. Recently, we developed a novel bent folded-end dipole transceiver array for human head imaging at 9.4 T. The new eight-element dipole array demonstrated full brain coverage, and transmit efficiency comparable to that of the substantially more complex 16-element surface loop array. In this work, we developed, constructed and evaluated a double-tuned 13C/1H human head 9.4 T array consisting of eight 13C transceiver surface loops and eight 1H transceiver bent folded-end dipole antennas all placed in a single layer. We showed that interaction between loops and dipoles can be minimized by placing four 1H traps into each 13C loop. The presented double-tuned RF array coil substantially simplifies the design as compared with the common double-tuned surface loop arrays. At the same time, the coil demonstrated an improved 1H longitudinal coverage and good transmit efficiency.

**KEYWORDS**

13C imaging, double-tuned coil, folded-end dipole, RF head array, transceiver array, ultra-high field MRI, whole-brain coverage

**Abbreviations:** AFI, actual flip angle imaging; CP, circularly polarized; CSI, chemical shift imaging; CW, continuous wave; DT, double-tuned; EM, electromagnetic; FA, flip angle; FoV, field of view; GRE, gradient echo imaging; HS, head and shoulder; MRS, magnetic resonance spectroscopy; MRSI, magnetic resonance spectroscopy imaging; Rx, receive; SAR, specific absorption rate; SEE, safety excitation efficiency; SNR, signal-to-noise ratio; SoS, sum of squares; ST, single-tuned; T/R, transmit/receive; Tx, transmit; TxRx, transceiver; UHF, ultra-high field; WSVD, whitened singular value decomposition.

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Magnetic resonance spectroscopy (MRS) and imaging (MRSI) using nuclei other than protons (1H) (X-nuclei), i.e., 13C, 31P, 23Na etc., provide valuable tools for biomedical research. However, due to a generally much lower gyromagnetic ratio and commonly smaller natural abundance (e.g., ~1% for 13C), these methods suffer from low signal-to-noise ratio (SNR). Thus, MRI at ultra-high field (UHF, ≥ 7 T), provides a natural strategy for improving the quality of X-nucleus MRS and MRSI due to the intrinsic benefit of increased SNR.

RF coils designed for X-nucleus MRS and MRSI often need to be double-tuned (DT). By double-tuning an RF coil we imply a capability of the entire RF coil structure to resonate and provide transmission and reception at two different frequencies, i.e., 1H and X-nucleus frequencies. An additional ability to transmit and receive at the higher 1H frequency is necessary for providing static magnetic field (B0) shimming, spatial localization, and proton decoupling and/or application of polarization techniques in the case of 13C nuclei. Proton decoupling as well as nuclear Overhauser enhancement can also be beneficial for other X-nuclei such as 31P. Considering that RF coils require both local transmission and reception at UHF, the designs of DT coils, which often consist of several layers of transmit (Tx) and receive (Rx) resonant elements, become quite complex. To simplify the DT coil development, usually the major design idea is to preserve high SNR and Tx efficiency (B1+ / P, where P is the RF input power) at the X-nucleus frequency, while the performance of the 1H part of the coil is compromised intentionally. However, the quality of 1H imaging is often equally important for many applications. In addition, the ability to use the same coil for a comprehensive examination, including both X-nucleus and 1H scanning, without moving the subject to replace the DT coil with a single-tuned (ST) 1H coil, is an important step toward clinical acceptance of X-nucleus MRS and MRSI.

There are two major options in designing a DT RF coil (or an array), i.e., double-tuning the same physical coil structure7–9 (or each element of the array1,14) to two different frequencies simultaneously or arranging two separate ST coils (or coil arrays) resonating at different frequencies close to each other.1,3–5,10 The first method allows optimization of the coil mainly at one frequency (commonly X-nuclei) while the Tx performance and SNR at the other frequency (commonly 1H) is degraded.1 The second method, i.e., combining two separate ST coils (arrays), allows for independent optimization of the DT coil at both frequencies. At the same time, the entire design becomes more complex.

Recently, we presented a DT transceiver (TxRx) 31P/1H 20-loop (10 loops at each frequency) array coil for human head imaging at 9.4 T.11 By making the entire coil design relatively simple and limiting the number of 31P elements to 10, we were able to place both the ST 31P array and the ST 1H array in the same layer and at the same distance to the head, which provided high loading and, thus, a good Tx efficiency for both arrays. As demonstrated previously, moving one of the arrays (e.g., 1H) further away from the sample into a second more remote layer degrades its Tx performance and SNR.3 It is also important that a relatively low number of elements (10 at each frequency) was sufficient to preserve high central SNR. It is known that increasing the number of smaller surface loops in a human head Rx array improves mostly the peripheral SNR and parallel imaging performance, while the central SNR does not substantially change12–14 or is even degraded due to insufficient loading.11 For non-proton applications, intrinsically low SNR largely limits parallel imaging acceleration in any case, and highly inhomogeneous reception profiles of high-density Rx arrays are not favorable because of a lack of normalization or quantification standards correcting for it. Hence, a smaller number of larger TxRx elements with deeper penetration and good coverage represent a better compromise for non-proton and DT coils.

In addition to high Tx efficiency and SNR, a well-designed state-of-the-art human head RF coil has to provide for longitudinal (along the magnet axis) coverage of the whole brain. This, however, is difficult to realize at UHF for 1H arrays using a single row of loops due to shortening of the RF wavelength (i.e., close to 100 mm at 400 MHz), and was demonstrated mainly by using substantially more complex multi-row (i.e., two rows) loop arrays1,15–19 capable of 3D RF shimming. In addition, multi-row arrays improve parallel (head-to-feet direction) and multi-band imaging. However, combining such a multi-row 1H array with an X-nucleus array into a single layer would make the design too complex, and to the best of our knowledge has never been presented. For example, to improve the 1H longitudinal coverage and still keep the DT coil relatively simple and robust, in the above described design of the 31P/1H 20-loop array coil11 we added two “vertical” cross-loops at the superior location of the head, changing the total number of TxRx 1H loops from eight to ten. This design improved the coverage as compared with a single-row ST eight-loop (1 × 8) 1H array,20 but was still worse than that of a double-row ST 16-loop (2 × 8) 1H array.19 Thus, further improvement of the Tx longitudinal coverage at the 1H frequency without substantially increasing the number of array elements is highly desirable.

A few years ago, a new type of RF coil, i.e., a dipole antenna, was developed and used for human body21–24 and head25–29 imaging at UHF. Due to the mechanical and electrical simplicity of dipole antennas, combining an X-nucleus surface loop array with 1H dipoles into a DT array can substantially simplify the design of a DT UHF human head coil. Previously, combining 1H dipoles with X-nucleus loops was demonstrated for body imaging.30 A human head hybrid 1H-dipole/23Na-loop array coil was also constructed in the past for imaging of the human head at 9.4 T.4 In this three-layer array design,4 four 1H TxRx dipoles were quite inefficient because of the distant location from the head and the presence of two more inner layers of elements, i.e., a four-loop 23Na-nucleus Tx array and a tight-fit 27-loop 23Na-nucleus Rx array. This work also showed substantial influence of the presence of the high-density Rx-only 23Na-nucleus array on the distribution of the RF magnetic field produced by the 1H TxRx dipoles.5 Thus, combining X-nucleus loops and 1H dipoles into a DT array requires further optimization to minimize alteration of the of the 1H dipole array performance. Recently, we developed a novel bent folded-end dipole TxRx array for human head imaging at 9.4 T.31 The new eight-element dipole array demonstrated full brain coverage, and Tx efficiency comparable to that of the substantially more complex 2 × 8 surface loop array.19
In this work, we developed, constructed and evaluated a DT $^{13}$C/$^1$H human head 9.4 T array consisting of eight $^{13}$C TxRx surface loops and eight $^1$H TxRx bent folded-end dipole antennas all placed in a single layer. We showed that interaction between loops and dipoles can be minimized by placing four $^1$H traps (parallel resonance circuit tuned to $^1$H frequency) into each $^{13}$C loop. The presented DT RF array coil substantially simplifies the design as compared with the $^{31}$P/$^1$H 20-loop array. At the same time, the coil demonstrated an improved $^1$H longitudinal coverage and good Tx efficiency.

2 | METHODS

2.1 | Array design

Following the design of the previously described $^{31}$P/$^1$H 20-loop array, we kept the same size of the array holder, ie 200 mm in width (left-right) and 230 mm in height (anterior-posterior). The holder was tapered at the superior location to improve fitting to a human head and measured 155 mm in width and 185 mm in height at the very top. In addition, the geometric arrangement and length of the $^{13}$C surface loops (ie 175 mm) was similar to the previously published $^{31}$P array, while the $^1$H dipole array was very similar to our previously published ST $^1$H bent folded-end dipole TxRx array. All surface loops and dipoles were constructed using 1.5 mm copper wires. The choice of 1.5 mm wire was mostly determined by its mechanical properties. 1.5 mm wire is sufficiently thick to be mechanically stable, and still can be relatively easily formed manually. Figures 1A-C show the electromagnetic (EM) simulation models and a photograph of the $^{13}$C/$^1$H array coil presented herein. To increase loading and minimize voids in the RF field near the gap between the loops, $^{13}$C surface loops were approximately 15% overlapped, in contrast with the gaps between the loops in our previous $^{31}$P/$^1$H 20-loop array design. $^1$H bent folded-end dipoles were placed in the center of each $^{13}$C loop as shown in Figure 1. Dipoles were slightly longer than surface loops (190 mm in length) and were extended outside of the head at the superior

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**Figure 1**  
A. EM simulation model of the new DT $^{13}$C/$^1$H array coil loaded by the HS phantom. B. Photograph of the constructed $^{13}$C/$^1$H array with the RF shield removed for better visualization. C. EM simulation model of the $^{13}$C/$^1$H array loaded by the Duke voxel model. Dipoles and loops are marked by red and yellow lines, respectively.
To minimize coupling between non-adjacent $^{13}$C loops and to efficiently decouple adjacent $^1$H dipoles, the array was shielded with a cylindrical RF shield placed at a 40 mm distance from the surface of the loops. The distance to the shield was chosen based on previously published data for UHF human head size arrays. As we showed previously, the presence of the RF shield significantly decreases coupling between non-neighboring surface loop elements of the array. This effect was demonstrated at both lower (124 MHz) and higher (300 MHz, 400 MHz) frequencies. Since decoupling of non-adjacent distantly located loops is hindered by the necessity of using long cables or wires, which have to be properly routed to avoid introduction of additional coupling, this helps to improve overall decoupling of the Tx loop array and optimize its performance. In addition, the RF shield placed close to the folded portion of the folded-end dipoles produces a mutual inductance between adjacent dipoles and compensates their intrinsic capacitive coupling. The RF shield also helps to avoid interaction of the coil elements with surrounding electronic and metallic structures in the magnet bore including the gradient coils.

2.2 | EM simulations

Before constructing the array, we evaluated the new design with respect to performance and safety using numerical 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 of the DT $^{1}$H/$^{13}$C array, we used three voxel models: a head and shoulder (HS) phantom (Figure 1A), which was constructed to match tissue properties ($\nu = 58.6$, $\sigma = 0.64$ S/m at 400 MHz), and Virtual Family multi-tissue models, “Duke” (Figure 1C) and “Ella”, cropped at chest level. For all voxel models, we used an isotropic resolution of 2 mm. In addition, we used an elliptical (140 mm width, 180 mm height, 200 mm length) phantom rounded at the top and filled with ethylene glycol. Simulations performed using the HS phantom and elliptical phantom were used for experimental comparison against the scanner-based evaluation of the $^{1}$H and $^{13}$C arrays, respectively.

To evaluate the interaction between $^{13}$C and $^1$H arrays and its effect on RF EM field distributions at both frequencies, the CST model of the DT array coil (Figure 1) included all 16 elements, ie eight $^{13}$C loops and eight $^{1}$H dipoles. For comparison, we also simulated corresponding ST eight-element loop and dipole arrays at both frequencies. In addition to minimizing coupling between the array elements, the presence of the RF shield can alter the Tx efficiency of the RF array coil. This effect is especially critical at the lower $^{13}$C frequency due to poorer loading. Therefore, to evaluate the effect of the RF shield on the $^{13}$C array performance, we simulated an unshielded version of the ST $^{13}$C array coil. In all simulations we also used a large copper cylinder (640 mm in diameter and 1600 mm in length), which mimicked the RF shield of the gradient coil.

All matching and tuning circuits were considered in the CST Design Studio (DS) module. The 16-element DT array loaded by a human head voxel model included 35 to 40 million mesh cells at the $^{13}$C frequency (100.5 MHz) and 60 to 70 million mesh cells at the $^1$H frequency (400 MHz). On our computer (2 × Intel Xeon Gold 6142 CPU @ 2.6 GHz with three GPU NVIDIA Tesla V100 PCIe-32 GB and 192 GB RAM), one simulation required about 30 h at 400 MHz and 17 h at 100.5 MHz.

As mentioned above, the presence of X-nucleus loops resonating at a substantially lower frequency can still strongly affect the performance of $^1$H dipole arrays. The common way to minimize the interaction of the X-nucleus and $^1$H elements of a DT array at the $^1$H frequency (400 MHz) is to introduce 400 MHz band stop filters (resonant $^1$H traps) into the X-nucleus coil elements. In the case of stronger interaction, several traps have to be distributed along the X-loop length. Therefore, we systematically evaluated how many $^1$H traps need to be introduced into each $^{13}$C loop to minimize alteration of the $^1$H dipole $B_1^+$ distribution and peak local specific absorption rate (SAR) value. We compared the performance of the final version of the DT $^{13}$C/$^1$H array (four $^1$H traps in each $^{13}$C loop) with that with the same design but with only two $^1$H traps. Placement of the traps is discussed below in the Section 2.3. In contrast, the performance of the X-nucleus portion of the DT coil often is not compromised by the presence of the $^1$H coil, and X-nucleus traps are not required to be placed into the $^1$H structure. We still verified this below by numerically comparing the performance of corresponding $^{13}$C ST and DT array coils.

$B_1^+$ 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. Additionally, we evaluated the Tx efficiency, both as $(B_1^+)/\sqrt{P}$, where $P$ is forward RF power measured at the array input (ie. feed-points of the antennas/loops), and as $(B_1^+)/\sqrt{\text{pSAR}_{10g}}$ (pSAR is the peak local SAR value), ie the safety excitation efficiency (SEE). The $B_1^+$ values were averaged over a 130 mm central transversal slab, which included the majority of the human brain.

2.3 | Array construction

After EM modeling, we constructed the new $^{13}$C/$^1$H DT array (Figure 1B). The geometry of the array holder and all elements were the same as described in Sections 2.1 and 2.2 (Figures 1A and 1C). Dipole and loop elements were placed on the surface of a fiberglass array holder with the...
wall thickness of 3 mm. As in the simulation model (Figure 1B), the array was shielded. The RF shield was constructed from a double-sided Kapton (25 μm thickness) copper clad laminate (AKAFLEX, Krempel, Vaihingen/Enz, Germany) as previously described.39

Figure 2A shows a schematic of a single ¹H dipole element including tuning capacitors (C\textsubscript{tune}) (a), the matching capacitor (b), a DT cable trap (c), and a home-built transmit/receive (T/R) switch circuit (d). Since the dipole length was slightly larger than half of the wavelength, to tune the dipoles we used capacitors (~30 pF) instead of inductors. Low-noise preamplifiers (WanTcom, Chanhassen, Minnesota, USA) were incorporated into the respective T/R switch circuits for both ¹H and ¹³C elements of the coil. Based on EM simulation data, we used variable matching capacitors of 20 pF. This was sufficient to provide matching on the HS phantom and various human heads. Following the previous work,21,36 the length of the fold and height of the ¹H dipoles (Figure 2A) measured 30 mm and 33 mm, respectively.

Figure 2B shows a schematic of a single ¹³C surface loop including the tuning variable capacitor (a) (Johanson, Boonton, New Jersey, USA), matching capacitor (b), DT cable trap (c), and home-built T/R switch circuit (d) integrated with a preamplifier as well as four ¹H traps. ¹H resonance traps were placed on every side of the loop and distributed relatively uniformly along its length to minimize coupling between ¹³C loops and ¹H dipoles. Adjacent ¹³C loops were decoupled by overlapping the loops.20,41 This also allowed us to bring the nearest non-adjacent loops (ie 1 and 3, 2 and 4 etc) closer to each other and utilize transformer decoupling to minimize crosstalk between them as suggested previously.42,43 Optimal overlapping (~15%) was evaluated numerically and further adjusted (within 5 mm) during construction to minimize interaction between neighboring ¹³C loops.

**FIGURE 2**  A, Schematic of a single TxRx ¹H bent folded-end dipole element including tuning capacitors (a), matching capacitor (b), the DT cable trap (c), and the T/R switch combined with the ¹H (400 MHz) preamplifier (d). B, Schematic of a single TxRx ¹³C surface loop including tuning (a), matching (b), the DT cable trap (c), and the T/R switch combined with the ¹³C (100 MHz) preamplifier (d). Four ¹H traps are introduced on each side of the ¹³C loop.
A requirement of MRS $^{13}$C experiments with proton decoupling is application of a very high (in a kW range) RF power at the $^1$H frequency to the DT coil during $^{13}$C reception.\textsuperscript{44-46} To avoid saturation of the $^{13}$C preamplifiers, a very high (–80 to –100 dB) isolation between the preamplifier inputs and the Tx input of the $^1$H array is required. To provide such a high isolation, in addition to minimizing coupling between individual $^1$H dipoles and $^{13}$C loops using $^1$H traps (Figure 2B), we placed the entire $^{13}$C eight-channel interface, which included the $^{13}$C eight-way splitter and eight $^{13}$C T/R switch boards (integrated with preamplifiers), in a separate box outside of the array holder. The $^{13}$C array was connected to this interface box using an eight-channel multi-modular coaxial connector (ODU, Mueldorf, Germany). In addition, at the input of each $^{13}$C T/R switch we introduced a second order 400 MHz band stop filter providing about –50 dB isolation. All $^{13}$C T/R switches, preamplifiers, and filters were shielded. Finally, transmitting 400 MHz high power amplifier during $^{13}$C reception may inject noise at the lower $^{13}$C frequency. To avoid such noise injection, we introduced a second order 100 MHz band stop filter at the 400 MHz array Tx input. The filter provided better than –40 dB isolation at 100 MHz and less than 0.1 dB insertion loss at 400 MHz.

During transmission, the array was driven in the quadrature circularly polarized (CP) mode at both frequencies, which in our case corresponded to a 45° phase shift between adjacent elements. For this purpose, we fabricated two eight-way splitters with corresponding phase shifters constructed of coaxial cables. Eight $^1$H T/R switches and the $^1$H eight-way splitter were located inside of the array holder to minimize cable losses.

### 2.4 Experimental evaluation of the array performance

After construction, we evaluated the new DT dipole/loop array in the scanner using phantom and in vivo experiments. Before in vivo measurements, a comprehensive evaluation according to the safety procedure previously developed in our laboratory\textsuperscript{47} was performed for the DT array presented herein, combining numerical simulations with evaluations on a bench and in the scanner. Bench evaluation of the array performance included measurements of the entire S-matrix ($8 \times 8$) at both frequencies using a network analyzer (E5071C, Agilent Technology, Santa Clara, California, USA). Finally, we evaluated the isolation between the $^1$H Tx port (the input of the $^1$H eight-way splitter) and inputs of the $^{13}$C preamplifiers at 400 MHz. In all bench measurements we used the HS phantom described above. We also compared the Tx performance of the array at the $^1$H frequency with that of the DT $^{31}$P/$^1$H 20-loop array\textsuperscript{11} and $^1$H ST double-row 16-loop array\textsuperscript{19} described previously.

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 (Erlangen, Germany) 9.4 T whole-body human MRI scanner. $B_{1^+}$ maps were acquired using the 3D actual flip angle imaging (AFI) sequence\textsuperscript{48} (field of view (FoV) 244 $\times$ 244 $\times$ 100 mm$^3$, voxel size 1.8 $\times$ 1.8 $\times$ 5 mm$^3$, $T_R$/100 ms, $T_E$ 4 ms, flip angle (FA) 60°). Experimental $^1$H SNR maps were evaluated using 3D gradient echo imaging (GRE) with a low FA (FoV 230 $\times$ 23 $\times$ 216 mm$^3$, voxel size 1.8 $\times$ 1.8 $\times$ 1.8 mm$^3$, FA 6°, $T_E$/TE 8/2 ms), and a corresponding noise scan (without RF), which was used to determine a noise correlation matrix.\textsuperscript{49} SNR was then calculated as an optimal weighted root sum-of-squares (SoS) combination taking into account a noise correlation.\textsuperscript{41} $^1$H SNR maps were also corrected for the spatial FA variations using AFI $B_{1^+}$ maps.

All $^{13}$C phantom measurements were performed with an elliptical phantom filled with ethylene glycol. A Tx field $^{13}$C $B_{1^+}$ measurement was acquired with the double-angle method\textsuperscript{50}: 3D chemical shift imaging (CSI), FoV 180 $\times$ 200 $\times$ 240 mm$^3$, matrix size 18 $\times$ 20 $\times$ 24, $T_E$ 600 ms, one average, rectangular hard pulse $T_P = 0.5$ ms, FA 20°/40°, vector size 1024, elliptical k-space shuttering scheme and 2 kHz acquisition bandwidth. SNR images were acquired with a 3D CSI sequence using the following parameters: 180 $\times$ 200 $\times$ 200 mm$^3$, matrix size 48 $\times$ 48 $\times$ 30, $T_R$ 113 ms, one average, rectangular hard pulse $T_P = 0.5$ ms, FA 30°, vector size 512, elliptical k-space shuttering scheme and 5 kHz acquisition bandwidth. Proton decoupling was tested on the phantom applying continuous wave (CW) decoupling\textsuperscript{51} with different voltages and a non-localized pulse acquire sequence: $T_R$ 30 s, two averages, FA 90°, vector size 128, acquisition bandwidth 3 kHz, 100% decoupling duration of 42 ms. Another decoupling method implemented for our 9.4 T scanner, ie WALTZ-4, showed worse results. The phantom measurement only required low band decoupling, which makes CW a suitable option. Additionally, we evaluated the noise injection at $^{13}$C frequency due to proton decoupling. This was measured using the sequence parameters of the non-localized acquisition and a pulse voltage of 0 V for the $^{13}$C excitation pulse. The noise was then calculated from the standard deviation of the time-domain signal without averaging (total number of time points - 256).

The reconstruction of the MRSI dataset was performed in MATLAB (Version R2018a) with a self-implemented reconstruction algorithm. Image reconstruction included FFT, spatial Hanning filtering and a WSVD (whitened singular value decomposition) coil combination.\textsuperscript{52} Signal amplitudes were calculated from a time-domain fitting using a home-built version of the AMARES algorithm (MATLAB R2018a).\textsuperscript{53} For optimization, the fmincon solver was applied.

The $^{13}$C SNR was calculated from the signal amplitude of the central peak of ethylene glycol (combined in the SoS manner), which was estimated from the time-domain fit, and divided by the noise. For the noise calculation, the signal from eight voxels outside of the phantom were used. Thereby, the noise was calculated as the standard deviation of the real component of the time-domain signal. As described above, a WSVD coil combination was used, which accounts for the noise correlation of the eight $^{13}$C loops. Since the $B_{1^+}$ map was relatively homogeneous, no $B_{1^+}$ correction was applied.
RESULTS

In the first step, we numerically evaluated how many $^1$H traps need to be introduced into each $^{13}$C loop to minimize the alteration of the dipole $B_{1+}$ distribution and increase of the peak local SAR value. We compared the performance of the DT $^{13}$C/$^1$H array with that of the ST $^1$H eight-element dipole array (without $^{13}$C loops) both loaded by the Duke voxel model. The DT array was simulated in two versions, i.e., $^{13}$C surface loops having two or four $^1$H traps. Figure 3 presents results of this comparison. Table 1 provides more details of the simulations. As seen in Table 1, the addition of $^{13}$C loops with four traps and two traps decreases the average $(B_{1+})/\sqrt{P}$ of the $^1$H dipole array by about 10% and 14%, respectively. SEE evaluated for both models (i.e., dipoles combined with two-trap loops and four-trap loops) measure about the same value, which is about 4% lower than that of the ST $^1$H dipole array (without $^{13}$C loops). In addition, the dipole $^1$H $B_{1+}$ field map in the presence of two-trap loops shows a higher $B_{1+}$ field near the nose (Figure 3A) due to a current induced in the corresponding $^{13}$C loops. This also increases local SAR in this location (Figure 3B).

Evaluation of the maximum local SAR produced by a DT RF coil at both frequencies is an important step of the safety evaluation procedure. Figures 3B and 3C show examples of SAR$_{10g}$ maps simulated for the presented DT array loaded by the Duke voxel model at both frequencies. Positions of maximum local SAR significantly differ at the two frequencies. While the $^{13}$C loop array generates the maximum SAR at the superior head location, the $^1$H bent folded-end dipole array produces the highest local SAR near the center of the head. Maximum SAR$_{10g}$ also slightly depends on the head size and measures 0.637 W/kg and 0.539 W/kg at 400 MHz for the Duke and Ella voxel models, respectively. At 100.5 MHz, we calculated the peak local SAR of 0.334 W/kg and 0.373 W/kg for the Duke and Ella voxel models.

We also evaluated the effect of combining $^{13}$C loops with $^1$H dipoles on the RF magnetic field and peak local SAR distributions at the $^{13}$C frequency. In addition, we simulated the performance of the $^{13}$C surface loop array without the RF shield. Simulation results are presented in Table 1 and Figures 4A-D. As seen in Figures 4A and 4C and Table 1, the addition of $^1$H dipoles has a minimal effect on the shielded $^{13}$C array Tx performance. $(B_{1+})/\sqrt{P}$ is reduced by less than 1% and SEE by only 2% (Table 1). The addition of four $^1$H traps caused further reduction of $(B_{1+})/\sqrt{P}$ and SEE by less than 4% and 1%, respectively (Table 1). At the same time, the removal of the RF shield (Figures 4B and 4D) causes a significant reduction of the $^{13}$C Tx efficiency $(B_{1+})/\sqrt{P}$ (by 16%) while SEE does not substantially change. Also, as seen in Figure 4A, the ST $^{13}$C loop array aligned...
with the top of the head produces a lower $\mathbf{B}_1^+$ value in the superior head area. Further improvement of the $\mathbf{B}_1^+$ distribution in this area can be simply achieved by extending the $^{13}\text{C}$ loops above the head or slightly (~20 mm) moving the head out of the coil as demonstrated in Reference 32. After constructing the DT array, we evaluated the coil performance on the bench. Figure 5 shows $8 \times 8$ S-matrices obtained for the presented DT array loaded by the HS phantom at both frequencies. The S-matrix measured at 400 MHz (Figure 5A) shows relatively strong coupling only between adjacent dipole elements, with a highest value of $-13.9$ dB and average value of $-15.6$ dB. Coupling between other elements was much lower and measured below $-21$ dB. The S-matrix obtained for the $^{13}\text{C}$ loop array (Figure 5B) demonstrates higher coupling between adjacent elements, with a highest value of $-13.3$ dB and average value of $-14.5$ dB. The worst coupling was measured for pairs of loops separated by two elements, i.e., 1 and 4, 2 and 5, etc (Figure 5C), with a highest value of $-11.7$ dB and average value of $-13.1$ dB. Coupling between other elements measured below $-19$ dB. In addition, we evaluated an interaction between $^1\text{H}$ dipoles and $^{13}\text{C}$ loops at both frequencies. At 400 MHz, the coupling measured $-27$ dB or lower. At 100.5 MHz, cross-talk was much less and measured below $-40$ dB between all the $^1\text{H}$ and $^{13}\text{C}$ elements. Finally, we measured the isolation between the $^1\text{H}$ Tx port and inputs of $^{13}\text{C}$ preamplifiers. Addition of second-order filters at the input of the $^{13}\text{C}$ combined T/R switches and pre-amplifier boards and shielding the $^{13}\text{C}$ interface allows us to obtain a total isolation better than $-80$ dB, as required for $^{13}\text{C}$ MRS experiments with proton decoupling.

In addition to simulations, the effect of the introduction of multiple $^1\text{H}$ traps into $^{13}\text{C}$ loops can be evaluated simply by measuring a change in the unloaded Q-factor ($Q_0$) value of the $^{13}\text{C}$ loop. We made these measurements for a single $^{13}\text{C}$ loop placed near the HS phantom, which loads the loop similarly to the human head. $Q_0$ measured 330, 300, and 265 and ratios of $Q_0/Q_L$ measured 4.1, 3.75, and 3.3 for the no-trap, two-trap, and four-trap cases, respectively. Based on analysis provided in Reference 54, insertions of two and four traps decrease SNR by 1.6% and 4.2%, respectively, as compared with the SNR of the loop without traps. However, large $^{13}\text{C}$ loops require at least two $^1\text{H}$ traps. Thus, the SNR of the four-trap loop array is decreased only by 2.6% in comparison with the two-trap loop design.

After constructing the DT array coil and numerically evaluating its safety, we tested the coil in the scanner using phantom and in vivo experiments. Figure 6A shows an experimentally measured $^{13}\text{C} \mathbf{B}_1^+$ map obtained using the presented DT array loaded by the ethylene glycol

### Table 1: Tx performance of the $^{13}\text{C}/^1\text{H}$ array

| Frequency (MHz) | Voxel model | Array | $\text{SAR}^a$ (W/kg) | $\langle \mathbf{B}_1^+ \rangle / \mathbf{P} (\mu\text{T}/\sqrt{\text{W}})$ | $\langle \mathbf{B}_1^+ \rangle / \mathbf{P}$ (ratio) | $\langle \mathbf{B}_1^+ \rangle / \mathbf{pSAR}$ (SEE) | $P_{\text{tissue}}^c$ (W) |
|----------------|-------------|-------|------------------------|---------------------------------|---------------------------------|---------------------------------|-----------------|
| 400           | Duke, EM sim. | dipoles | 0.637 | 12.49 | 1.0 | 15.65 |   |
|               |             | dipoles + loops (4 traps) | 0.571 | 11.29 | 0.9 | 14.94 |   |
|               |             | dipoles + loops (2 traps) | 0.511 | 10.76 | 0.86 | 15.04 |   |
|               | HS phantom, EM sim. | dipoles + loops (4 traps) | 9.4 |   |   |   |   |
|               | HS phantom, experim. | dipoles + loops (4 traps) | 8.7 |   |   |   |   |
|               | HS phantom, experim. | dipoles + loops (2 traps) | 7.9 |   |   |   |   |
|               | In vivo, experim. | dipoles + loops (4 traps) | 8.3 |   |   |   |   |
| 100.5         | Duke, EM sim. | loops | 0.324 | 30.44 | 1.0 | 53.47 | 0.544 |
|               |             | loops (no RF shield) | 0.222 | 25.46 | 0.84 | 54.04 | 0.389 |
|               |             | loops + dipoles | 0.334 | 30.20 | 0.99 | 52.25 | 0.537 |
|               |             | loops + dipoles (2 traps) | 0.323 | 29.63 | 0.98 | 52.14 | 0.522 |
|               |             | loops + dipoles (4 traps) | 0.316 | 29.11 | 0.96 | 51.78 | 0.508 |
|               | Elliptical phantom, EM sim. | loops + dipoles | 37.2 |   |   |   |   |
|               | Elliptical phantom, experim. | loops + dipoles | 32 |   |   |   |   |

$^a$Maximum SAR averaged over 10 g of the tissue; calculated for 1 W of RF stimulated power at the array input, i.e., eight feed-points of the antennas/loops.

$^b$ $\mathbf{B}_1^+$ is averaged over 130 mm central transversal slab; calculated for 1 W of RF stimulated power at the array input, i.e., eight feed-points of the antennas/loops.

$^c$Calculated for 1 W of RF stimulated power at the array input, i.e., eight feed-points of the antennas/loops.
phantom. Averaged over the 130 mm transversal slab, $\langle B_1^+ \rangle / \sqrt{P}$ measured 32 $\mu$T/√kW. This value was obtained considering losses in the cable and splitter and normalized to the RF power value at the coil input, i.e., eight feed-points of the antennas/loops. The simulated $\langle B_1^+ \rangle / \sqrt{P}$ value was higher and measured 37.2 $\mu$T/√kW (Table 1). Figure 6B presents the $^{13}$C SNR map.

Results of experimental tests at the $^1$H frequency of the presented DT array coil loaded by the HS phantom are shown in Table 1. As seen in the table, a placement of four $^1$H traps into each $^{13}$C loop provided approximately 10% higher experimentally measured $\langle B_1^+ \rangle / \sqrt{P}$ as compared with the case of $^{13}$C loops with only two traps. At the same time, the simulated four-trap array version demonstrated 8% higher $^1$H $\langle B_1^+ \rangle / \sqrt{P}$ field strength in comparison with what was measured experimentally.
Figure 7 presents examples of transversal, coronal, and sagittal in vivo human brain GRE images (Figure 7A) and corresponding $^1\text{H}$ $B_1^+$ maps (Figure 7B) obtained using the $^1\text{H}$ part of the presented DT $^{13}\text{C}/^1\text{H}$ array. Averaged over the $130 \text{ mm}$ transversal slab, $\langle B_1^+ \rangle / \sqrt{P}$ measured $8.3 \mu\text{T}/\sqrt{\text{kW}}$ as normalized to the RF power at the array coil input, i.e., eight feed-points of the antennas/loops. In addition, Figure 8 shows a comparison of central sagittal images (Figure 8A) and corresponding $B_1^+$ (Figure 8B) and SNR maps (Figure 8C) obtained using three head array coils, i.e., the $^1\text{H}$ part of the new DT 16-element $^{13}\text{C}$/ $^1\text{H}$ array, the $^1\text{H}$ part of the DT 20-loop $^{31}\text{P}/^1\text{H}$ array, and the 16-loop double-row ST $^1\text{H}$ array. Averaged over the $130 \text{ mm}$ transversal slab, $\langle B_1^+ \rangle / \sqrt{P}$ measured $9.7 \mu\text{T}/\sqrt{\text{kW}}$ and $9.9 \mu\text{T}/\sqrt{\text{kW}}$ for the DT $^{31}\text{P}/^1\text{H}$ 20-loop array and $^1\text{H}$ ST double-row 16-loop array, respectively. SNR was about $15\%$ and about $50\%$ higher using the DT $^{31}\text{P}/^1\text{H}$ array and 16-loop double-row ST $^1\text{H}$ array, respectively.

Figure 9A and 9B demonstrates $^{13}\text{C}$ MRS with direct $^{13}\text{C}$ detection and proton decoupling using the ethylene glycol phantom. Different voltages for the CW proton decoupling pulse are applied and corresponding non-localized spectral and time domain signals are presented. At a voltage of $250 \text{ V}$, the desired decoupling behavior is achieved. At lower voltages, only a partial decoupling is detected over the entire volume of the phantom. For a decoupling voltage of $250 \text{ V}$, the peak local SAR reaches $31\%$ of the maximal allowed value compared with no significant absorption without decoupling and the applied sequence parameters. This calculation of peak local SAR was based on a conservative approach that considers an additional safety factor of $2$ for the $k$-factors at both frequencies. $^5$ The increase of the noise level (standard deviation) is within the measurement error ($\sim 10\%$).

4 | DISCUSSION

As mentioned above, maintaining reasonably high Tx and Rx performance of the DT coil at the $^1\text{H}$ frequency is very important for many applications. However, it is often compromised for the sake of design simplification. Previously, we developed the UHF DT $^{31}\text{P}/^1\text{H}$ head 20-loop array coil, which could preserve the $^1\text{H}$ Tx performance and central SNR. The coil demonstrated an average Tx ($B_1^+$) field value at the $^1\text{H}$ frequency, similar to that of ST single-row ($1 \times 8$) and double-row ($2 \times 8$) $^1\text{H}$ arrays. The addition of two TxRx “vertical” cross-loops at the superior location of the head improved the longitudinal brain coverage as compared with an ST $^1\text{H}$ eight-loop array. $^2^0$ However, the coverage was still worse than that obtained by the $2 \times 8$ ST $^1\text{H}$ array. $^1^9$ The presented novel design of the DT array coil combining $^{13}\text{C}$ surface loops and $^1\text{H}$ bent folded-end coil...
dipoles further improves the brain coverage at the $^1$H frequency without increasing the number of elements. As seen in Figures 7 and 8, the new DT array coil provides a Tx longitudinal coverage comparable with that of the substantially more complex $2 \times 2$ array. The $B_1^+$ maps obtained using the developed array (Figure 7B) have some artifacts at the top of the head, which are not present in the maps of $31$P/$^1$H and $^1$H ST $2 \times 8$ arrays (Figure 8B). The better coverage of the new array leads to aliasing artifacts in the $B_1^+$, as signal from the neck area is folding back into upper parts of the head. At the same time, in vivo $^1$H $\langle B_1^+ \rangle$ field strength (averaged over the 130 mm transversal slab) produced by the new DT array was about 17% lower than that of the $^1$H ST surface loop arrays. Thus, the decrease in the Tx efficiency ($\langle B_1^+ \rangle/\sqrt{P}$) is a compromise for the improved longitudinal coverage. Also, the DT array presented herein exhibits an approximately 11% lower $^1$H Tx efficiency than the $^1$H ST bent folded-end dipole array constructed previously. Such difference is similar to that expected from numerical simulations, eg the presence of $^{13}$C loops with four $^1$H traps reduces the Tx efficiency of the $^1$H dipole array by 10% for the Duke voxel model (Table 1). The simulated SEE values obtained for the presented DT coil (Table 1) and ST $2 \times 8$ array are similar.

It also important that in comparison to $^{31}$P/$^1$H array, which in addition to eight surface loops surrounding the head has two TxRx cross-loops at the superior head location, the developed dipole/loop array (without having two superior cross-loops) provides comparable SNR at the superior head area and somewhat better SNR down the brain stem (Figure 8). For comparison, an eight-loop array design, which does not include the superior cross-loops, has significantly lower SNR in the superior head area. The SNR of the $^1$H-dipole array can be further improved by adding a pair of Rx-only cross-loops or cross-dipoles at the top of the head.

As we demonstrated in our study, the correct choice of the number of $^1$H traps and their placement in X-nucleus surface coils is critical for maintaining the $^1$H dipole array Tx performance. It is not easy to provide a general recipe for choosing the trap number and their placement, which depends on the specific geometry of the array, eg the size and geometry of the X-nucleus loops. However, we would like to emphasize the following features. First, dipoles interact mainly with two $^{13}$C surface loops located beneath adjacent dipoles and much less with the loop beneath the dipole itself. In our case, this interaction is further increased because we used wider overlapped $^{13}$C loops. Making surface loops narrower and increasing gaps between them will most likely decrease the interaction between dipoles and loops, but also compromise the performance of the X-nucleus array. However, the smaller number of $^1$H traps distributed along the X-nucleus loop may suffice. Second, in our design, the length of the wire that was used to construct the $^{13}$C loop measured about 600 mm. By interrupting the wire with two $^1$H traps (Traps 2 and 4, Figure 2A), we produced two 300 mm pieces of wire. This is very close to half of the wavelength at 400 MHz and the total length of the folded-end dipoles.

**FIGURE 7** Examples of central transversal, coronal, and sagittal in vivo human brain $^1$H GRE images (A) and corresponding $B_1^+$ maps (B) obtained using the presented DT $^{13}$C/$^1$H array. The averaging 130 mm transversal slab is shown in B.
Thus, these wires can strongly interact with the dipoles and disturb the RF field distribution (Figure 3A). Again, for other designs with loops of different size, the results may differ. While $^1$H traps were critical to maintaining the performance of the dipole array, the performance of the $^{13}$C loop array was not significantly altered by the presence of $^1$H dipoles (Table 1) and hence no $^{13}$C traps were placed in the $^1$H elements.

Evaluation of the peak local SAR is an important part of the safety evaluation procedure. As seen in Figure 3C, the $^{13}$C surface loop array demonstrates a SAR distribution, which is characteristic for the CP mode, ie with local SAR increased at the superior location and very low SAR near the center of the head. At the same time, the $^1$H dipole array shows the highest local SAR near the center of the head. This fact is explained by coupling of the 9.4 T bent folded-end dipole array to the intrinsic TE mode of the head, which produces a tangential component of electric field near the head center. As demonstrated previously, bending and folding the dipoles in the presence of the RF shield facilitates coupling of the array to the TE mode at 400 MHz. Optimization of the length of the folded portion of the dipole allows minimization of the peak SAR value and SEE. 

A difference in Tx efficiency was observed for both frequencies between the EM simulation results (higher) and the experimental results (lower, Table 1), which can be explained by difficulties in taking into account all losses produced in coil components and conductors in EM simulation models. Commonly, EM simulations give overestimates of the $B_1^+$ value. This issue is even more pronounced for dipole antennas, where relatively coarse meshing is performed over thin (1.5 mm in diameter) dipole wires. Importantly, this difference between simulated and experimental data does not compromise the coil safety due to overestimations of peak local SAR values as well.

An RF shield is commonly used in the UHF RF array coil designs to minimize radiation losses, decrease coupling between non-adjacent elements, and reduce interaction with the magnet bore environment including the gradient coil. At the same time, the presence of the RF shield may decrease the Tx efficiency and SNR due to out-of-phase current induced in the shield. This is more important at the lower X-nucleus frequency due to lower tissue losses, and, therefore, higher Q-factors. Therefore, we investigated the effect of the RF shield presence on the $^{13}$C loop array Tx performance. Interestingly, without the shield the $^{13}$C loop array demonstrated lower (16%) Tx efficiency than the shielded array.
This is explained by a significant increase of coupling between non-adjacent loops due to RF shield removal, which in turn causes an increase of the reflection power and, as a result, a decrease in the coil accepted power and power deposited into tissue (Table 1). As seen in Table 1, the tissue power deposition was reduced by 38% for the unshielded array, which matches the reduction of the $^{13}$C-array $B_1^+$ value. Thus, in our case, the RF shield plays an important role for both $^1$H and $^{13}$C array designs due to an improvement of the element decoupling.

Choosing an appropriate decoupling method is a critical component of any TxRx-array development. In our design, decoupling of dipole elements was provided by the presence of the RF shield. As demonstrated recently, decoupling between adjacent folded-end dipoles is produced due to a capacitive coupling of the folded portion to the RF shield. Double-tuning and the presence of the $^{13}$C loops increased coupling between adjacent dipoles, which was still quite reasonable, i.e. average $S_{12}$ of $-15.6$ dB between adjacent dipoles. This is about 2-3 dB worse than that obtained for the $^1$H ST bent folded-end dipole array.

At a lower frequency (100.5 MHz), array decoupling is more challenging. Due to substantially lower loading, there is a need to decouple adjacent as well as closest non-adjacent loops (i.e. 1 and 3, 2 and 4, etc). At the end, we still measured the highest coupling (~$-13$ dB on average) between loops separated by two elements (i.e. 1 and 4, 2 and 5, etc), which were not decoupled at all. Incorporating more decoupling circuits to minimize coupling between these $^{13}$C loops would make the design quite complex, and to the best of our knowledge has never been presented before. This residual coupling caused a small inhomogeneity seen in the transversal in vivo $^{13}$C $B_1^+$ map (Figure 6).

As discussed above, the new DT array substantially simplifies the coil design and provides for good longitudinal $^1$H coverage without increasing the number of elements. At the same time, the Tx efficiency is still lower than that of the $^1$H ST loop array. We believe that optimization of the dipole geometry and the trap placement may help to further optimize the performance of the dipole array. In addition, with only eight $^1$H Rx-elements, the parallel imaging performance (mainly head-to-feet direction) and ability to perform multi-band reception are still compromised as compared with state-of-the-art 32-channel Rx head coils available on UHF systems.

**5 | CONCLUSIONS**

In this study, we developed, evaluated, and constructed a DT $^{13}$C/$^1$H human head 9.4 T array design consisting of eight TxRx $^{13}$C surface loops and eight TxRx $^1$H bent folded-end dipole antennas, all placed in a single layer. The new DT array coil demonstrated improved $^1$H longitudinal coverage and good Tx efficiency. We also showed that coupling between $^{13}$C loops and $^1$H dipoles can be minimized by placing four $^1$H traps in each $^{13}$C loop. The presented DT RF array also substantially simplifies the head coil design as compared with previously developed DT array designs, e.g. a $^{31}$P/$^1$H 20-loop array.
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DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available from the corresponding author upon reasonable request.

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