Double-tuned $^{31}$P/$^1$H human head array with high performance at both frequencies for spectroscopic imaging at 9.4T

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Purpose: To develop a robust design of a human head double-tuned $^{31}$P/$^1$H array, which provides good performance at both $^{31}$P and $^1$H frequencies for MR spectroscopic imaging at 9.4T.

Methods: Increasing the number of surface loops in a human head array improves the peripheral signal-to-noise ratio (SNR), while the central SNR doesn’t substantially change. High peripheral SNR can contaminate MR spectroscopic imaging data at both $^1$H and $^{31}$P frequency. To minimize this effect, we limited the number of elements in the $^{31}$P array to 10, i.e., 8 transceiver surface loops circumscribing the head and 2 receive “vertical” loops placed at the superior location. The $^1$H-portion of the array also consists of 10 elements, i.e., 8 transceiver surface loops circumscribing the head and 2 transceiver “vertical” loops at the superior location of the head. Both the $^{31}$P array and $^1$H array are placed in a single layer at the same distance to the head, which provides high loading and, thus, a good performance for both arrays.

Results: Transmit efficiency of the $^1$H-portion of the double-tuned array was very similar to that of the single-tuned arrays of similar size. Also, addition of the cross-loops substantially improved the brain coverage.

Conclusion: We developed a novel $^{31}$P/$^1$H double-tuned array for MR spectroscopic imaging of a human brain at 9.4T. Placing both $^{31}$P and $^1$H loops in a single layer provides for high transmit efficiency at both frequencies without compromising SNR near the brain center at the $^{31}$P-frequency. Addition of the cross-loops at the superior location improves the brain coverage.

Keywords: central SNR, double-tuned arrays, MRSI, RF human head coil, ultra-high field MRI
1 INTRODUCTION

X-Nuclei MRI and MR spectroscopic imaging (MRSI) of the human brain provides valuable information about metabolic changes in many pathologies and is sensitive to detect abnormalities at an early disease stage. However, imaging other than hydrogen nuclei, i.e., X-nuclei, such as $^{31}$P, $^{13}$C, or $^{23}$Na, is often difficult due to a lower gyromagnetic ratio and, thus, a lower signal-to-noise ratio (SNR). In addition, the natural abundance of some X-nuclei (e.g., $^{13}$C, $^2$H) is in the low percentage range, which further decreases the detected signal. Therefore, an enhancement of SNR with an increase of the magnetic field strength, $B_0$, one of the major advantages of ultra-high field (UHF, ≥7T) MRI, is very critical for X-nuclei imaging.

To provide high-resolution $^1$H anatomical human head imaging and $B_0$ shimming, the radiofrequency (RF) coil must be double-tuned (DT). The latter implies that the same coil is capable of resonating at 2 substantially different frequencies, i.e., $^1$H and X-nuclei. It is rather difficult to optimize the DT RF coil at both frequencies at the same time. Therefore, often the coil performance at the X-nuclei frequency is optimized while the $^1$H-performance is not.1 Good performance of the DT array at $^1$H frequency, however, is still important for many applications. In fact, widespread translation of X-nuclei imaging into neuroscientific, physiological, and clinical studies is currently hindered by the need of changing the RF coil between X-nuclei data acquisition and conventional $^1$H based study protocols to provide higher transmit (Tx) and receive (Rx) performance and better coverage for $^1$H MRI. The ability to use the same RF coil for comprehensive anatomical, functional, and metabolic scan protocols, including $^1$H MRI and X-nuclei MRI, without the necessity to move the subject out of the magnet and replace a DT coil with a single-tuned (ST) $^1$H coil is an important step towards the establishment of X-nuclei imaging.

For a human head ST array, a nested double-layer combination of a Transmit-only array (or a volume coil) with a Receive-only array, i.e., a ToRo setup,4,7 provides the most flexible design, which allows optimizing the Tx and Rx performance separately. In this design, a multi-element tight-fit Rx array is placed inside of a larger local Tx array or body transmit coil. At the same time, the ToRo-coil design requires additional electronics for detuning both Tx and Rx arrays,4,7 which makes the design more complicated. Alternatively, the same elements of an array can be used during both transmission and reception. This design, a so-called transceiver (TxRx) array coil,3,8-10 is more simple to construct than a ToRo-setup but more difficult to optimize simultaneously for both transmission and reception. Also, any TxRx-design suggests a presence of additional Tx/Rx switches, which, however, do not have to be built into the RF coil itself.

There are several previously reported DT UHF head array coil designs.3,11-17 To the best of our knowledge, a 4-layer DT array coil consisting of 2 double-layer ToRo-setups (both X-nuclei and $^1$H) has never yet been reported due to its high complexity. Commonly, DT human head array coils consist of 2,5,12,13,16,17 or 3 layers.11,14,15 The double-layer design contains 2 TxRx volume coils or TxRx arrays. A more complicated 3-layer design usually consists of an X-nuclei ToRo-set up and either a local $^1$H TxRx volume coil11,15 or TxRx array.14 This design mostly aims to optimize SNR at the lower X-nuclei frequency by increasing the number of elements (commonly surface loop) in the tight-fit X-nuclei Rx array. The efficiency of the X-nuclei Tx array (or a volume coil), and especially performance of the $^1$H TxRx coil, which both made larger to fit the X-nuclei high loop count Rx array inside, are not optimized.

It is also noteworthy that often increasing the number of smaller surface loops in a human head Rx array improves mostly the peripheral SNR, while the central SNR does not substantially increase18-20 or is even impaired due to insufficient loading. High peripheral SNR can be harmful for MRSI due to contamination of near-cortical voxels by strong fat ($^1$H)21 or muscle ($^{31}$P)22,23 signals. Low central SNR is a major limiting factor, because for most applications the metabolic information from either the whole brain or a combination of regional cortical and deeper structures is required. In addition, the strong signal intensity gradient from the periphery to the center of the brain makes it difficult to obtain quantitative X-nuclei images. A respective signal intensity correction is hard to achieve for X-nuclei MRI and MRSI due to the lack of a suitable reference signal. Summarizing all the above, a major idea for a new DT array design, which performs well at both frequencies and has a sufficiently low difference between central and peripheral SNR, includes limiting the number of elements in both $^1$H and X-nuclei arrays to what is necessary not to compromise central SNR and Tx-performance. Thus, a double-layer DT coil consisting of only 2 TxRx arrays, which have a relatively low count of elements at each frequency, can offer a simple and robust design for X-nuclei MRI and MRSI studies.

In this work, we developed and constructed a novel 20-element DT $^{31}$P/ $^1$H 9.4T (399.72 MHz - $^1$H, 161.8 MHz - $^{31}$P) human head array, which provides for good coverage of the human brain, high central SNR, and efficient Tx-performance at both frequencies. The array consists of 16 TxRx surface loops (8 loops at each frequency) circumscribing the head. All 16 loops are placed on the surface of the same tight-fit array holder to make sure that both $^{31}$P and $^1$H surface loops are located at the same short distance from the sample. In addition, the array has 2 pairs ($^{31}$P and $^1$H) of “vertical” loops placed at the superior location of the head. We performed a comprehensive array evaluation and comparison to the performance of a previously described 8-loop single-row (1 × 8) ST $^1$H TxRx array of similar geometry24 and 16-loop double-row (2 × 8) ST $^1$H TxRx array.10 We also compared the array $^{31}$P SNR to that of the commercially available 7T 3-layer $^{31}$P/$^1$H array head coil.
2 METHODS

2.1 Phased array design

As mentioned above, increasing the number of smaller surface loops in a human head Rx array can be harmful for X-nuclei MRI and MRSI. In addition, smaller loops can compromise the central SNR due to insufficient loading and, therefore, higher contribution from intrinsic coil losses. To minimize these effects, we limited the number of loops in both the 1H and 31P portions of the DT array to 10 each. This relatively small loop count helps to substantially reduce SNR at the periphery of the head while keeping the central SNR high. Figure 1 shows the electromagnetic (EM) simulation models of the 10-element 1H and 31P arrays loaded by phantoms. Larger sized 31P loops help to maintain higher loading, which is very critical at the lower 31P-frequency. Surface and vertical loops are constructed using 1.5-mm copper annealed wires. The choice of 1.5-mm wire is mostly determined by the mechanical stability and convenience in constructing the loops.

The 1H array (Figure 1A) consists of eight 11-cm long overlapped TxRx surface loops circumscribing the head and 2 TxRx “vertical” cross-loops placed at the superior location. Use of a single-row 8-loop array seems to be sufficient not to compromise central SNR as well as B1 homogeneity. Two 1H vertical loops are slightly different in size and measured 90 mm x 40 mm and 110 x 40 mm. The 31P array (Figure 1B) also consists of 10 elements, i.e., eight 17-cm long TxRx surface loops gapped by 10 mm and circumscribing the head. In addition, the 31P array had 2 Rx-only “vertical” loops at the superior location. Two 31P Rx vertical cross-loops are larger than the corresponding 1H loops and measure 110 mm x 50 mm and 130 x 50 mm. Two pairs of vertical loops (Figure 1) have coplanar geometry with 1H loops placed symmetrically inside corresponding 31P loops.

The choice of the vertical loop geometry is relatively simple. An increase of loops’ size improves the penetration depth and coverage at the superior location of the head. Therefore, our selection procedure consists of an increase of the loop length and width within the limitation of the current mechanical design, which is determined by the size and geometry of the array holder, position of RF electronics inside the array (e.g., Tx/Rx switches), etc. All 16 TxRx surface loops (1H and 31P) are placed on the surface of the same array holder (Figures 1C and 2A), which provides a similar tight fit for both 31P and 1H arrays. In addition, to minimize interaction between adjacent 31P and 1H surface loops, 1H loops are shifted by half a loop size (Figure 1C). The 1H array is decoupled entirely by overlapping the adjacent loops. The 31P array is decoupled by transformer decoupling. The transformer decoupling method has advantages over capacitive decoupling, since it does not require expensive high-voltage nonmagnetic

FIGURE 1 A, CST EM simulation model of the 1H part of the DT 31P/1H array loaded by the HS phantom. B, CST EM simulation model of the 31P part of the DT 31P/1H array loaded by the elliptical phantom. C, Photo of the DT 31P/1H array. Cover is removed for better visualization. 1H and 31P loops are marked by red and yellow dashed lines, respectively.
variable capacitors, and is intrinsically broadband. To decrease radiation losses\textsuperscript{26} and minimize coupling between nonadjacent loops,\textsuperscript{27} the array is shielded with the cylindrical shield placed at a 40-mm distance from the surface loops. The distance to the shield is chosen based on previously published data for UHF human head arrays.\textsuperscript{3,10,24}

### 2.2 EM simulations

EM simulations of the transmit $B_1^+$, where $B_1^+$ is the circularly polarized (CP) component of the RF magnetic field, and the local specific absorption rate (SAR) were performed using CST Studio Suite 2017 (CST, Darmstadt, Germany) and the time-domain solver based on the finite-integration technique. The solver was stopped when the total energy in the system fell below $-50$ dB of the maximum monitored system energy. We also used approximately 40 million mesh cells with the smallest cell size of 0.8 mm.

The array was modeled following the array geometry described above (Figure 1), and consisted of 20 (16 surface and 4 vertical) loops. Ten and 8 fixed capacitors were distributed over the $^1$H loop length for surface and vertical loops, respectively. Both vertical Rx and surface TxRx $^{31}$P loops had 6 distributed capacitors. To account for additional losses, all fixed capacitors were modeled with equivalent resistors placed in series. Resistor values were obtained from component datasheets. Tuning and matching capacitors were substituted by ports and adjusted during co-simulations. $^1$H surface loops were decoupled by overlapping. Overlap between adjacent loops was adjusted individually by changing the loop width. After adjustment, overlapping measured $\sim 16$ mm. For evaluation of the $B_1^+$ and SAR distribution maps at $^{31}$P frequency (162 MHz), 2 vertical Rx loops were actively detuned. As demonstrated previously, a presence of detuned Rx loops may alter the maximum local SAR value.\textsuperscript{28}

Three voxel models were used in simulations, i.e., the head and shoulder (HS) phantom (Figure 1A), which was constructed to match tissue properties ($\varepsilon = 58.6$, $\sigma = 0.64$ S/m) at 400 MHz,\textsuperscript{7,29} and 2 virtual family multi-tissue models, “Duke” and “Ella”,\textsuperscript{30} cropped at the level of the chest. For all 3 voxel models, we used a 2-mm resolution. For evaluation of the transmit performance of the $^{31}$P array, we used an elliptical phantom (length, 17 cm; axes of ellipse, 18 cm x 15 cm; $\varepsilon = 62.4$; $\sigma = 0.54$ S/m at 160 MHz) shown in Figure 1B. $B_1^+$ field profiles and local SAR$_{10g}$ (averaged over 10 g of tissue) maps were calculated for 1 W of total (per 8 channels) power at the coil input and compared with experimentally measured data. Averaging of SAR was performed using the CST Legacy method.

Numerical optimization of the size of vertical loops within the entire design (including all surface loops) is a lengthy
process since the total model is rather large. However, since coupling between vertical and surface loops is sufficiently low, we simulated separately a pair of $^1$H vertical loops with 20-mm larger width (i.e., width of 60 mm) to evaluate an increase in the $B_1$ field. As a result, larger loops generated a 7% higher magnetic field at the depth of 20 mm from the top of the head. There was practically no difference near the center of the head.

### 2.3 Array Construction

After EM modeling, we constructed the 20-loop DT $^{31}$P/$^1$H array (Figure 2A,B). Figure 2C,D show the assembled array connected to the splitter and loaded by the HS phantom. The geometry of the array holder and the loops is the same as in EM modeling (Figure 1). The size of the fiberglass array holder measures 230 mm in height and 200 mm in width. The array holder is also tapered to improve fitting to a human head and measures 155 mm in width (left-right) and 185 mm in height (anterior-posterior) as shown in Figures 1C and 2A. Eight $^1$H surface loops circumscribing the head are decoupled by overlapping adjacent loops without any additional decoupling circuits. As shown previously, this method provides for very good decoupling and substantially simplifies the entire array design. During the construction procedure, we started with the overlap distances used in EM simulations, i.e., ~16 mm, which was then adjusted manually to minimize coupling.

Figure 3A shows the schematic of a single $^1$H surface loop including the tuning and matching capacitors, DT cable trap, and home-built Tx/Rx switch circuit (−0.25 dB of insertion loss, −40 dB of isolation). Each surface loop has 12 fixed chip capacitors (100C series, American Technical Ceramics, Huntington Station, NY) distributed along the loop. Fixed capacitor values range from 5.6 pF to 6.8 pF, except for two 3.3-pF capacitors, which are connected in parallel to tuning and matching variable capacitors (Figure 3A). Tx/Rx switches are connected to each loop and located inside the array holder to minimize losses. A low-noise preamplifier (WanTcom, Chanhassen, MN) is incorporated into each Tx/Rx switch circuit. All surface loops are individually tuned and matched using variable capacitors (Figure 3A). Tx/Rx switches are connected to each loop and located inside the array holder to minimize losses. A low-noise preamplifier (WanTcom, Chanhassen, MN) is incorporated into each Tx/Rx switch circuit.

During transmission, $^{31}$P vertical loops are actively detuned. Active detuning circuitry is constructed using 2 PIN diodes (MACOM, Lowell, MA) connected in series (Figure 3D). Commonly active detuning circuits are constructed using 1 PIN diode. The second PIN diode in our design provides additional blocking of 6 dB or more. PIN diodes are driven using MRI system direct current drivers, which provide 100 mA of current. Direct current voltage is delivered to the PIN diodes through the RF cable using an RF choke (RFC, Figure 3D). As a secondary safety feature, each $^{31}$P vertical loop has a protective fuse. In addition, all $^{31}$P Rx and TxRx loops have 2 $^1$H traps constructed by connecting an inductor in parallel to a 7.5 pF capacitor (Figure 3C,D). Placement of 2 $^1$H traps increases Q$_0$ by ~20% and further improves decoupling of the $^1$H surface loops. Each $^1$H trap is individually tuned to maximize its impedance at $^1$H frequency (399.72 MHz).

During transmission, both $^1$H and $^{31}$P arrays are driven in the quadrature CP mode, i.e., with a 45° phase shift between adjacent loops. For this purpose, we constructed 2 Wilkinson splitters both placed into the same box (Figure 2D), i.e., one 9-way $^1$H splitter and one 8-way $^{31}$P splitter. Two $^1$H TxRx vertical cross-loops (Figure 1A) are driven in quadrature (90° phase shift between the loops) using an additional home-built 90° hybrid splitter. Also, to align the vertical loops’ $B_1$ near the center of the array with that of corresponding surface loops, superior $^1$H cross-loops are driven in phase with surface loops 1 and 7 (Figure 4). To simplify connection of the array to the splitters, all RF cables are combined into 2 bundles (Figure 2C, ODU Tx), each with a modular connector (ODU GMBH, Muehldorf, Germany).

### 2.4 Performance and safety evaluation

Before doing in vivo measurements, the phased array was evaluated on a bench and in the scanner and numerically simulated according to the safety procedure developed in our lab. The human subjects participated in the study after
giving signed informed consent according to procedures approved by the local institutional review board committee.

Bench evaluation of the array includes measurements of the loaded and unloaded Q-factors ($Q_L$ and $Q_U$) and decoupling between each pair of the surface loops. Q-factors were measured from the absolute value of the $S_{11}$ reflection coefficient as twice the ratio of the resonance frequency over the 3-dB bandwidth using a network analyzer (E5071C, Agilent Technology, Santa Clara, CA). Decoupling between TxRx loops (Tx-mode) was evaluated by measuring the $S_{12}$ transmission coefficient with 31P Rx-only vertical loops actively detuned. In all these measurements, we used the HS phantom.

All data were acquired on a Siemens Magnetom (Erlangen, Germany) 9.4T human imaging system. $^1$H $B_1^+$ maps were acquired using the 3D actual flip angle imaging sequence (field of view [FOV], $244 \times 244 \times 100 \text{ mm}^3$; voxel size, $1.8 \times 1.8 \times 5 \text{ mm}^3$; repetition time (TR)/TR$_2$, 20/100 ms; echo time [TE], 4 ms; flip angle, $60^\circ$). Experimental $^1$H SNR maps were evaluated using 3D gradient echo imaging (GRE) with a low flip angle (FOV, $244 \times 244 \times 100 \text{ mm}^3$; voxel size, $1.8 \times 1.8 \times 5 \text{ mm}^3$; flip angle, $6^\circ$; TR/TE, 8/2 ms), and
a corresponding noise scan, which was used to evaluate a noise correlation matrix. SNR was then calculated as an optimal weighted root sum-of-squares combination taking into account a noise correlation matrix. SNR maps were also corrected for the spatial flip angle variations using actual flip angle imaging B+1 maps. Experimental 1H B+1 maps were also compared to maps obtained using the HS phantom as well as in vivo. Experimental 1H B+1 maps were also compared to maps obtained using the 1 × 824 and 2 × 810 ST arrays constructed previously. The 1 × 8 array has very similar size and geometry to the 1H portion of the DT array while the 2 × 8 array is longer and measures 17.5 cm. Both ST arrays lack a pair of TxRx cross-loops at the superior location of the head.

31P B+1 maps were evaluated using MRSI phantom data. For the SNR measurement, we used an elliptical phantom with a rounded top (length, 18 cm; axes of ellipse, 19 cm × 15 cm; ε = 62.4; σ = 0.54 S/m at 160 MHz). SNR maps were acquired with a 3D MRSI pulse sequence (FOV, 220 × 220 × 220 mm3; voxel size, 5.9 × 5.9 × 5.9 mm3; weighted elliptical k-space; TR = 100 ms; 3 averages; flip angle, 25°; rectangular pulse excitation with 0.5 ms pulse duration; 5 kHz acquisition bandwidth; vector size, 512; acquisition time, 50 min). Experimental 31P B+1 maps were compared with maps obtained using commercial 7T 3-layer 31P/1H array head coil (length, 26.4 cm; axes of ellipse, 22.4 cm × 19.5 cm) made by Rapid Biomedical GmbH (Ripan, Germany). The commercial array contains 32 31P Rx loops.

3 | RESULTS

First, we evaluated the array on a bench, which included measurements of Q-factors and the S12 matrix between TxRx loops. The ratio of Q_U/Q_L of the surface loops in the 1H array loaded by the HS phantom measures from ~9 (anterior) to 6 (posterior). Q_U measures ~240. Q_U/Q_L of the 1H vertical TxRx loops was lower and measures ~3.5. Q_U measures ~280. Figure 4 shows S12 matrices measured for both the 1H and 31P arrays loaded by the HS phantom. The 1H array is well decoupled by overlapping the loops with the strongest coupling (less than −18 dB) measured between closest nonadjacent 31P TxRx loops, i.e., 1 and 3, 2 and 4, etc.

![Figure 4](image-url)

FIGURE 4 8 × 8 S12 matrices measured for 1H (A) and 31P (B) TxRx surface loops of the DT array loaded by the HS phantom. For S12 matrix measurements, 31P vertical Rx-only loops were actively detuned (Tx-mode). Numbering of 1H and 31P surface loops are shown on the left side of the figure. White lines show strongest coupling between closest nonadjacent 31P TxRx loops, i.e., 1 and 3, 2 and 4, etc.
All adjacent TxRx loops are decoupled better than −18 dB with an average $S_{12}$ value of −19.2 dB. Closest nonadjacent loops (i.e., 1 and 3, 2 and 4, etc) show the strongest coupling, which ranged from −13 dB to −16 dB with an average $S_{12}$ value of −14.6 dB. Isolation between $^1$H and $^{31}$P neighboring loops measured −30 dB or better at both frequencies.

Together with the bench evaluation, we performed the safety evaluation of the new DT array according to the procedure developed in our institution. As an example, Figure 5 shows the $B_1^+$ and SAR$_{10g}$ distributions obtained for the $^1$H and $^{31}$P arrays both loaded by the Duke voxel model. Maximum SAR$_{10g}$ values were calculated for the CP mode at both frequencies, and various loading conditions mimicking variation in the head size. In these simulations, we first evaluated the maximum local SAR$_{10g}$ for perfectly tuned and matched $^{31}$P and $^1$H arrays for both voxel models. In addition, we calculated the maximum SAR$_{10g}$ for each model using the array tuned and matched on the other model. For example, an array adjusted on the Duke model was loaded by the Ella model and vice versa. At the end, the maximum local SAR$_{10g}$ value at each frequency was chosen from 4 values. Final maximum SAR$_{10g}$ values measured 0.76 W/kg and 0.64 W/kg at $^1$H and $^{31}$P frequencies, respectively. In both cases, worse maximum SAR$_{10g}$ values were obtained for the Ella voxel model when tuning and matching were performed on the Duke model. We also calculated a change in the maximum SAR of the $^1$H array due to presence of the vertical loops. The maximum local SAR increased by 8% and 2% for Duke and Ella models, respectively.

In the next step, we evaluated Tx and Rx performance of the DT array loaded by phantoms described above. Figure 6 shows simulated and experimentally measured $^1$H and $^{31}$P $B_1^+$ maps. Simulated and experimental data match each other well. For both arrays, simulated $B_1^+$ values are ~10% higher than experimentally measured, which commonly happens due to difficulties taking into account losses in all components of the RF coil. This, however, only leads to a small overestimation of the maximum local SAR and does not cause any safety issues. The experimentally measured $^{31}$P $B_1^+$ value averaged over the entire phantom is $19.4 \pm 4.2 \mu T/\sqrt{W}$. The $B_1^+$ value near the center is 30.3 $\mu T/\sqrt{W}$. Figure 7 shows SNR maps obtained using the $^1$H and $^{31}$P arrays loaded by phantoms. Figure 7B,C compare $^{31}$P SNR obtained with and without Rx vertical cross-loops. The Supporting Information Figure S1A, which is available online, shows the $^{31}$P SNR map for the central coronal slice. It is noteworthy that the addition of vertical cross-loops improves the $^1$H $B_1^+$ distribution (Figure 6A,B) and SNR of both arrays (Figure 7) at the superior location of the head. Transversal $^{31}$P SNR maps (Figure 7B) shows left/ right asymmetry. The asymmetry most likely occurred due to residual coupling between closest nonadjacent loops (e.g., 1 and 3, 2 and 4, etc), which is seen in the $S_{12}$ matrix (Figure 4B) and is difficult to eliminate due to the distant location.

After performing coil test measurements using phantoms and evaluating coil safety, we tested the array in vivo. Figure 8 shows in vivo results obtained using the new DT array at $^1$H-frequency (399.72 MHz). In addition, Figure 8 also demonstrates experimental data obtained using the $^1$H 2 × 8 and 1 × 8 ST arrays. Similar to the phantom data, the addition of the cross-loops substantially improves the longitudinal brain coverage as compared to the 1 × 8 ST array. It is also of importance that the Tx efficiency of the DT array is very similar to that of the ST arrays. Averaged
FIGURE 6  A, Experimentally measured and simulated central sagittal, coronal, and transversal $B_1^+$ maps obtained using the $^1$H part of the $^{31}$P/$^1$H array loaded by the HS phantom. B, Experimentally measured and simulated central sagittal, coronal, and transversal $B_1^+$ maps obtained using the $^{31}$P part of the array loaded by the elliptical phantom.

FIGURE 7  A, Central sagittal, coronal, and transversal SNR maps obtained using the $^1$H part of the $^{31}$P/$^1$H array loaded by the HS phantom. B, Central sagittal and transversal SNR maps obtained using the $^{31}$P part of the array loaded by the elliptical phantom with the rounded top. C, Central sagittal SNR map obtained using the $^{31}$P part of the array without vertical cross-loops.
over a 120-mm transversal slab (Figure 8C), the new DT array $B_1^+$ measures 9.98 ± 3.21 µT/√kW. For the ST 1 × 8 and 2 × 8 arrays, $B_1^+$ measures 10.07 ± 4.05 µT/√kW and 10.06 ± 2.57 µT/√kW, respectively. The Tx efficiency of the DT array is practically the same as that of both ST arrays. Also, the relative standard deviation of the DT array $B_1^+$ value over 120-mm transversal slab measures 0.32 versus 0.40 and 0.26 for the 1 × 8 and 2 × 8 ST arrays, respectively. Figure 8E,F compare SNR obtained by DT and ST 1H arrays. The DT array delivers much higher SNR at the superior location than the 1 × 8 ST array. Both arrays have very similar SNR near the center. The 2 × 8 ST array provides ~35% higher SNR than the DT array both at the center and periphery.

Figure 9 demonstrates PCR $^{31}$P MRSI images obtained using the new DT array. Shown are transversal and sagittal slices, which exhibit good coverage over the entire brain. As an example, we show 2 voxels, 1 from the periphery and 1 from the central area of the brain, to demonstrate that we achieve a good signal in the center relative to the periphery with the constructed $^{31}$P/1H array. The central coronal slice is shown in Supporting Information Figure S1B. Finally, Figure 10 shows a comparison of the new 9.4T DT array’s Rx performance to that of the commercial 7T DT array using the same human head shaped $^{31}$P phantom. In this example, we also show SNR at 2 voxels located near the center and periphery. While both arrays have very similar peripheral SNR (less than 3% difference), the 9.4T array has 3.9 times higher SNR near the center. The ratio of the peripheral to the central SNR measured 10.8 and 2.7 for the commercial array and our arrays, respectively.

### DISCUSSION

We developed, constructed, and evaluated the new DT $^{31}$P/1H (161.8 MHz/ 399.72 MHz) array design for brain studies at 9.4T. Commonly, to simplify decoupling and placement of multiple elements of a DT array, 1H loops are moved away from the subject and positioned in a second layer. Alternatively, they are made substantially smaller than X-nuclei loops and are located inside of them. Both methods decrease loading of 1H loops and, thus, compromise the Tx-performance and SNR of the DT array at 1H frequency. It is also noteworthy that, at UHF, simple increasing of the length of a single-row 1H 1 × 8 loop array is not sufficient to provide for whole-brain coverage. Good longitudinal coverage of the whole brain has been demonstrated by using multi-row (≥2) arrays, e.g., 2 × 8 arrays, which provide the capability of 3D RF shimming. However, the combination of a 2 × 8 1H TxRx array with any $^{31}$P array has never been reported due to the complexity of the design.

At the same time, the high Tx-performance and whole-brain coverage of the DT array at 1H frequency are important for many applications. Our new DT array design provides for a solution to these issues by using 2 new features as compared to previously reported designs. First, we placed TxRx surface loops of both $^{31}$P and 1H parts of the array on the surface of the same head and at the same distance to a head. Second, we added 2 pairs of 1H vertical cross-loops at the superior location of the head. These modifications allowed us not to compromise the Tx-efficiency of the 1H portion of the array as compared to the ST arrays of similar size. The
**Figure 9** A. Transversal GRE in vivo image and $^{31}$P spectra obtained for 2 voxel positions shown in the image. B. Central sagittal and transversal in vivo PCr MRSI $^{31}$P maps. Data are acquired using 5 averages over FOV: 240 mm $\times$ 240 mm $\times$ 200 mm. Nominal voxel size: 12 mm $\times$ 12 mm $\times$ 20 mm. Sagittal and transversal MRSI maps are masked to remove strong contribution from temporalis muscle PCr.

**Figure 10** A. Central coronal, sagittal, and transversal in vivo PCr MRSI $^{31}$P maps obtained using the new 9.4T DT array and commercial 7T DT array. Data are acquired using 3 averages over FOV: 220 mm $\times$ 220 mm $\times$ 200 mm. Nominal voxel size: 5 mm $\times$ 5 mm $\times$ 10 mm. $^{31}$P spectra obtained for 2 voxel positions shown on the transversal maps in A using our new DT array (B) and the commercial DT array (C).
new DT array showed very similar $B_1^+$ value averaged over the 120-mm transversal slab (Figure 8C) as compared to both the 1 × 8 and 2 × 8 ST arrays. Such a tight-fit single-layer arrangement is feasible only because of the relatively small number of elements. With a higher element count, $^{1}H$ loops have to be placed in the second layer at a larger distance to the sample, which inevitably decreases the $^{1}H$ Tx-efficiency. 3

Second, an addition of 2 pairs of $^{1}H$ vertical cross-loops provides a significant improvement of the longitudinal coverage compared to the $^{1}H$ ST 1 × 8 array. As seen from Figure 8, vertical cross-loops substantially improve both the $^{1}H$ $B_1^+$ distribution and SNR at the superior location of the head. While the coverage (both Tx and Rx) is still worse compared with the $^{1}H$ ST 2 × 8 surface loop array, it is much better than that of the $^{1}H$ ST 1 × 8 array. As previously shown, combining a pair of cross-loops with a 2 × 8 TxRx surface loop array further improves the longitudinal coverage of the $^{1}H$ array. 41 However, such a design, i.e., the 18 TxRx loop array, would be even harder to combine with a multi-channel $^{31}P$ array within a single layer. Thus, the new design provides for a solution having a reasonable longitudinal coverage using a relatively small loop count.

Regarding performance at the $^{31}P$ frequency, as seen from Figure 7, the addition of $^{31}P$ vertical cross-loops improves SNR at the superior location of the head. Also, data shown in Figure 9 demonstrate that we achieved very good coverage over the whole brain using the new DT array with a reasonable measurement time of 30 min and a nominal voxel size of 1.75 mL and an improved $^1$H performance. The representative central voxel shows good SNR compared to the peripheral location. It is also noteworthy that our low loop count DT array provides a substantially more uniform SNR distribution than that of the commercial high loop count array. As seen from Figure 10, the peripheral SNR of the 9.4T DT array is only 2.7 times higher than the central SNR. For comparison, the 32-channel commercial array shows a significantly greater (i.e., ~4 times) difference, which is harmful for MRSI experiments. It is also important that the 9.4T array demonstrates almost 4 times higher SNR near the center of the phantom while showing very similar SNR at the periphery.

According to recent theoretical 42 and experimental 43 studies, the central human head SNR has a nearly quadratic dependence on the constant magnetic field $B_0$, while the peripheral SNR grows rather linearly. Therefore, such a large difference in the central SNR cannot be explained by the difference in $B_0$ values, i.e., 7T versus 9.4T, which provides only factor of 2 difference. The residual difference occurs due to a larger count of smaller loops in the $^{31}P$ commercial Rx array, i.e., 32 versus 20. Smaller loops are less loaded and, therefore, do not provide for the sample noise domination. 24 As a result, the central SNR is compromised as compared to that of an array with larger loops. On the other hand, because the 7T $^{31}P$ resonance frequency is very close to that of the 3T $^{1}H$ resonance frequency, a reasonable comparison can be made with previously reported 32-loop 3T Rx arrays, which demonstrate about 6 times difference between peripheral and central SNR. 18,44,45 The residual difference between central and peripheral SNR could also occur due to the presence of the $^{1}H$ coil or some differences in array geometries.

## CONCLUSIONS

We developed a novel $^{31}P/^{1}H$ DT array for MRSI of a human brain at 9.4T. Placing both $^{31}P$ and $^{1}H$ loops in a single layer at the same distance to the sample provides for high Tx efficiency at both frequencies without compromising SNR near the brain center at the $^{31}P$ frequency. Addition of the cross-loops at the superior location of a head substantially improves the brain coverage at $^{1}H$ frequency as compared to the ST single-row 1 × 8 array.

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SUPPORTING INFORMATION
Additional Supporting Information may be found online in the Supporting Information section.
FIGURE S1 A. Central sagittal, coronal, and transversal SNR maps obtained using the $^{31}$P part of the array loaded by the elliptical phantom with the rounded top. B, Central sagittal, coronal, and transversal in vivo PCr MRSI $^{31}$P maps. Data are acquired using 5 averages over FOV: 240 mm × 240 mm × 200 mm. Nominal voxel size: 12 mm × 12 mm × 20 mm. Sagittal and transversal MRSI maps are masked to remove strong contribution from temporalis muscle PCr.

How to cite this article: Avdievich NI, Ruhm L, Dorst J, Scheffler K, Korzowski A, Henning A. Double-tuned $^{31}$P/$^{1}$H human head array with high performance at both frequencies for spectroscopic imaging at 9.4T. Magn Reson Med. 2020;00:1–14. https://doi.org/10.1002/mrm.28176