Design and characterization of an eight-element passively fed meander-dipole array with improved specific absorption rate efficiency for 7 T body imaging

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Objective: To evaluate the transmit efficiency and specific absorption rate (SAR) efficiency of a new eight-element passively fed meander-dipole antenna array designed for body MRI at 7 T, and to compare these values with a conventional directly fed meander-dipole array.

Methods: The main radiating element of the passively fed dipole is printed on one side of a dielectric substrate and is capacitively coupled to a shorter feeding element (connected to the coaxial cable) printed on the opposite side of the substrate. The transmit ($B_1^+$) field and SAR were simulated on a phantom and on a human voxel model for both a passively fed and a directly fed single element. Two eight-channel arrays containing, respectively, directly and passively fed meander dipoles were then simulated, and experimental $B_1^+$ maps and $T_2$-weighted spin echo images of the prostate were obtained in vivo for four healthy volunteers.

Results: In simulations, the mean transmit efficiency ($B_1^+$ per square root input power) value in the prostate was $\sim 12.5\%$ lower, and the maximum 10 g average SAR was $44\%$ lower for the array containing passively fed dipoles, resulting in $\sim 15\%$ higher SAR efficiency for the passively fed array. In vivo RF-shimmed turbo spin echo images were acquired from both arrays, and showed image SNRs within 5% of one another.

Conclusion: A passive-feeding network for meander-dipole antennas has been shown to be a simple method to increase the SAR efficiency of a multi-element array used for body imaging at high fields. We hypothesize that the main reason for the increase in SAR efficiency is the storage of the strong conservative electric field in the dielectric between the feeding element and the radiating element of the dipole. The passive-feeding approach can be generalized to other dipole geometries and configurations.

KEYWORDS
7 T MRI coils, dipole antennas, passive feed, specific absorption rate, transmit efficiency
Human MRI at high field strengths is challenging due to significant inhomogeneities in the transmit ($B_1^+$) fields that occur as a result of the relatively short RF wavelength in tissue compared with the dimensions of the human body. This is particularly evident in body applications at field strengths of 7 T and higher, which has led to the use of transmit arrays consisting of multiple (typically 8–32 elements) transmit antennas. This approach uses individually controlled transmit channel phases and/or magnitudes. A number of different resonators have been used for the individual elements of these arrays, including loops and microstrips, but in the past few years various forms of dipole antennas have become the predominant geometry. The intrinsic far-field nature of a dipole is advantageous in terms of transmitting electromagnetic energy into deep-lying organs within the body. Dipole antennas have been shown both theoretically and experimentally to be efficient transmit elements for body imaging at 7 T in particular, and have also recently been used at 10.5 T.

After the initial demonstration of the performance of a conventional straight dipole, a substantial amount of research has concentrated on optimizing the dipole geometry in order to improve a combination of the transmit efficiency ($B_1^+$ field per square root input power) and the specific absorption rate (SAR) efficiency ($B_1^+$ field to the square root maximum specific absorption rate ($SAR_{10g, max}$)). For example, the fractionated dipole antenna presented in is a straight printed dipole with several meandered structures included in each leg of the dipole, which act as inductances to increase the electrical length of the dipole. This design produces the lowest $SAR_{10g, max}$ among current dipoles used for high field imaging. Dipoles can be used as transmit-only or transceiver elements, either alone or in combination with loop antennas as additional receive elements. Fractionated dipoles can be combined in pairs to give independent excitation of the even and odd modes of the dipole pair.

For all of the dipole geometries described in the literature, including the meander dipole, the element is directly fed via a partially or fully balanced lumped matching element network consisting of capacitors and/or inductors. However, it is also possible to feed dipole elements passively, i.e., via inductive and capacitive coupling from a passive structure that is coupled to the dipole. In this study, we evaluate experimentally and in simulations the passive-feeding approach for a 300 mm long meander-dipole antenna used as an array element for prostate imaging at 7 T. The main radiating antenna element was printed on one side of a dielectric substrate, coupled to a shorter passive feeding element printed on the opposite side. Numerical and experimental comparisons were performed with a conventional array containing directly fed meander-dipole antennas.

### 2 | MATERIALS AND METHODS

#### 2.1 | Dipole fabrication

The passively and directly fed meander-dipole geometries are shown in Figure 1. The length and geometry of the short feeding dipole (in the passively fed meander-dipole design) was chosen via a parametric study (data not shown), in which the length of the feeding dipole was varied from 90 to 150 mm, in steps of 10 mm, and both a straight and meandered geometry were studied, in order to obtain the maximum SAR efficiency. The design of the directly fed meander dipole is identical to that described previously. The dipoles were each 300 mm long and were fabricated on an FR-4 printed circuit board (PCB) substrate ($\varepsilon_r = 4.3$, $\tan \delta = 0.025$, substrate thickness = 1.5 mm). For a passively fed dipole (Figure 1A), one side of the PCB (“back” side) contained a short feeding element (length = 110 mm) with a small meander at the end of each leg. The other side of the PCB (“front” side) contained a larger radiating dipole and was oriented towards the phantom/subject. The passively fed meander dipole had a 33 nH
inductor connected between the legs of the short feeding element and two identical 3.9 pF matching capacitors (Figure 1A). The directly fed meander dipole (Figure 1B) had two identical matching capacitors with a value of 22 pF. For the transmit/receive array eight directly fed and eight passively fed meander antennas with the same total dimensions as described above were constructed. A 20 mm thick polymethylmethacrylate (PMMA) spacer was attached to the face of the dipole.

$S_{11}$ was measured for both the directly and passively fed meander dipoles as a function of the distance between the single antenna and the phantom: this distance was varied from 20 to 70 mm in steps of 10 mm. The reference point for both antennas was the feed point, i.e., the short feeding dipole was at the same distance from the phantom as the feed point of the directly fed dipole.

### 2.2 Electromagnetic simulations and antenna measurements

Electromagnetic simulations were performed in CST Microwave Studio 2016 (CST Studio Suite, Computer Simulation Technology, Darmstadt, Germany). Simulations were first performed with a single antenna element on a square phantom (Figure 2A). Subsequently, eight-element arrays containing directly or passively fed dipoles (Figure 2B) were simulated using the voxel model Gustav (CST Studio Suite, Computer Simulation Technology). The voxel size in Gustav was 2 x 2 x 2 mm. The PMMA spacer placed between the antenna and phantom/voxel model was included in the simulations. The dielectric properties of the phantom used were $\varepsilon_r = 50$ and $\sigma = 0.6 \, \text{S/m}$. To evaluate the $B_1^+$ and SAR$_{10\, \text{g, max}}$ efficiency of the single antenna, results were normalized to 1 W of accepted power. In the eight-element body array simulations, $B_1^+$ and SAR$_{10\, \text{g, max}}$ were normalized to 8 W of accepted power. Simulations of the SAR and $B_1^+$ of passively and directly fed dipole antennas were performed with open boundary conditions in all directions. The mesh type used was hexahedral. The total number of cells for the single antenna phantom simulations was 1.5 million cells and the simulation time was around two minutes on a standard desktop computer. The total number of cells for the eight-channel body array with voxel model Gustav was ~ 14.5 million and the simulation time was around 30 minutes.

Simulated RF shimming was performed by adjusting the phases of the individual channels to produce the maximum constructive interference of all channels in the prostate region. The phase of the $B_1^+$ field produced by each channel in the prostate was estimated, and the corresponding negative phase applied to the transmit channel was such that the phases of all channels were zero in the prostate, and that the $B_1^+$ fields added constructively.

### 2.3 MRI measurements

All studies were approved by the medical ethics committee of the Leiden University Medical Center. In vivo experiments were performed using a Philips (Philips Best, The Netherlands) 7 T whole body scanner equipped with an eight-channel parallel transmit architecture (Philips Achieva). Single antenna $B_1^+$ maps were measured on a phantom, and in vivo $B_1^+$ maps were obtained using the dual refocusing echo acquisition mode (DREAM)$^{29}$ sequence with the following parameters: field of view (FoV) = 400 x 320 x 25 mm, voxel size = 5 x 5 x 5 mm, slices = 5, tip angle = 10°, stimulated acquisition mode (STEAM) angle = 50°, TE/TR = 1.97/15 ms, one signal average. Based on the in vivo $B_1^+$ maps, phase-based RF shimming was performed as described above (with a linear optimization algorithm written in MATLAB on a PC linked to the MR scanner) to obtain the maximum $B_1^+$ in the prostate of a healthy volunteer. Equal magnitude excitation was used on all channels.

MRI-based thermal measurements were performed using the proton reference frequency method$^{30}$ with antennas placed on the phantom. The 3DGRE sequence used for heating had the following parameters: TR/TE = 14/10 ms, flip angle (FA) = 10°, scan duration = 1800 s; the final temperature was measured after 30 minutes of scanning.

Four healthy volunteers with different body mass indices (BMIs) were imaged. The BMIs varied from 20.8 to 28.7 kg/m$^2$. The sequence parameters used for in vivo imaging were as follows: a $T_2$-weighted turbo spin echo (TSE) sequence, TE/TR = 110/6000 ms, flip angle = 90°, TSE
factor = 20, FoV = 250 x 380 x 16 mm, voxel size = 0.8 x 0.8 mm, slice thickness = 3 mm, one signal average. A noise correlation matrix was generated by acquiring data with the RF and gradient channels turned off, with the noise correlation coefficient ($\rho_{ij}$) defined as

$$\rho_{ij} = \frac{\text{cov}(n_i, n_j)}{\sigma_i \sigma_j}$$

where $n_i$ is the noise measured in coil element $i$. SNR maps were acquired using the method described previously. \(^{31}\)

3 | RESULTS

Figure 3 shows simulated and measured $B_1^+$ and SAR$_{10g, \text{max}}$ values, as well as the thermometry measurements from single dipole elements placed on the phantom. For the passively fed meander-dipole antenna, the simulated and measured $B_1^+$ field (normalized to 1 W of accepted power) of a single element is lower than that of the directly fed antenna (Figure 3A-D), while the SAR$_{10g, \text{max}}$ is reduced by 37% (Figure 3E and 3F). Figure 3G

FIGURE 3 (A, B) Simulated and (C, D) measured $B_1^+$ fields for the single directly and passively fed meander dipoles on a phantom. (E, F) Simulated max SAR$_{10g}$ for the single directly and passively fed meander dipoles on a phantom. (G) Thermometry measurements of (left) directly and (right) passively fed meander dipoles. (H) Simulated (left) $B_1^+$ per square root of power and (right) $B_1^+$ per square root of max SAR$_{10g}$ of both antennas along a central line on a phantom.
shows the results of the MRI-based thermal measurements performed for the same input power and the same heating time. The passively fed antenna produced ~ 50% less heating (measured at the point of maximum heating, which was directly under the central point of the dipole) than the directly fed antenna. The maximum temperature increase produced by the directly fed meander antenna was 12°C while the maximum temperature increase for the passively fed meander antenna was 6°C.

Figure 3H shows a graph of the $B_1^+$ per square root of unit power (transmit efficiency) and $B_1^+$ per square root of SAR$_{10g, max}$ (SAR efficiency) for both directly and passively fed dipoles plotted along the central line in the phantom. The figures show that the transmit efficiency is higher for the directly fed dipole at the surface of the phantom, whereas the SAR efficiency is higher for the passively fed antenna for all depths. The reference point for the measurements was the feed point, ie the short feeding dipole was at the same distance from the phantom as the feed point of the directly fed dipole (note that in the case where the radiating conductors are instead used as reference points, the $B_1^+$ efficiency increased by 2.7% at the surface and by less than 0.5% at 5 cm depth for the directly fed meander antenna).

In order to investigate the origin of the increased SAR efficiency, Figure 4A shows simulated surface current distribution on an antenna’s conductor facing a phantom of (left) directly and (right) passively fed meander dipole. Figure 4B shows the simulated z-component of the E-field ($E_z$) on the surface of the dipoles. For both cases, the strongest $E_z$ values were at the feed point, around the meanders and at the end of the dipoles. Figure 4C shows the $E_z$ field from a side view of the simulated setup containing a directly/passively fed meander-dipole antenna placed on a 20 mm spacer above the phantom. Figure 4D shows the $E_z$ field at the surface of the phantom (top view). The $E_z$ field component was higher for the directly than for the passively fed dipole throughout the entire phantom.

**FIGURE 4**  (A) Surface current density on a conductor's surface of the (left) directly and (right) passively fed meander dipole. Orientation of the conductor's surface was towards a phantom. (B) Z-component of the E-field at the surface of the (left) directly and (right) passively fed meander dipole. (C) Side view of the E-field z-component of the dipole antenna placed 20 mm away from the phantom. (D) Z-component of the E-field at the surface of the phantom of the (left) directly and (right) passively fed meander dipole.
Figure 5 shows the measured $S_{11}$ of both dipole antennas as a function of the distance from the phantom, varying from 20 to 70 mm in steps of 10 mm. Both antennas were impedance-matched for a 20 mm distance from the phantom. As the distance increased, the $S_{11}$ minima of both antennas shifted towards higher frequencies by the same amount, indicating an equal amount of inductive coupling to the sample (note that the $S_{11}$ plot of the passively fed antenna has two peaks: one from the long passive element and the other from the short feeding dipole).

Figure 6 shows the simulated $B_1^+$ and $B_1^-$ fields and $SAR_{10g, max}$ values for the two different eight-channel arrays (both passively and directly fed antennas) on the body voxel model. The simulated $B_1^+$ field was ~12.5% lower in the prostate region for the passively fed meander-dipole arrays while the $B_1^-$ field was ~6% lower for the passively than the directly fed meander. The simulated $SAR_{10g, max}$ produced by the passively fed dipole array is 44% lower than that from the directly fed dipole array, giving an increase in SAR efficiency of ~15%.

Figure 7 shows in vivo images from one of the four healthy volunteers. The experimentally (in vivo) measured noise correlation matrices of both arrays and for all four volunteers are shown in Figure 8. The coupling coefficients between individual channels of both arrays were below ~15 dB. The measured averaged $B_1^+$ field in the prostate region was ~15% lower, while the averaged SNR in a prostate was ~5% lower for the array containing passively fed dipoles.

Figure 8 shows measured noise correlation matrices of four volunteers with BMIs spanning from 20.8 to 28.7 kg/m$^2$. In cases of both passively and directly fed arrays, the highest coupling coefficient was ~15 dB. The highest inter-element coupling corresponded to the subject with the highest BMI value, which can be explained by the fact that tuning and matching of antennas were performed on a homogeneous phantom, where the antennas were placed at a 20 mm distance from the phantom, and for this particular volunteer the effective distance was much greater than this distance.

4 | DISCUSSION

This study compares the transmit and SAR efficiencies of two identical meander-dipole configurations, the difference being that one is passively fed and the other is directly fed. The results show that the transmit efficiency of a passively fed dipole or dipole array is reduced by ~15% at penetration depths of ~10–15 cm, whereas the $SAR_{10g, max}$ is reduced by more than 40% and the SAR efficiency is increased by ~15%. Because
most high field body imaging applications are limited by the SAR rather than the absolute power available from the multiple RF amplifiers, these results suggest that changing from an actively fed to a passively fed architecture could have significant benefits for body imaging at high fields.

Why does a passive feed reduce the SAR and increase the SAR efficiency? In order to reduce SAR the total electric field penetrating the sample should be minimized, ideally without significantly affecting the magnetic $B_1^+$ field component. The SAR depends on both the conservative and inductive (nonconservative) electric field components. The conservative electric field originates from a gradient in charge accumulation on the antenna’s conductor. From Figure 4B and 4C, we conclude that the highest $E_z$ components (which can be attributed to the conservative electric field) occur at the location of the meanders, at the feed point, and at the end of the antenna’s legs. The $E_z$ component is important because, based on the boundary conditions at the surface of the phantom, the tangential component of the electric field in the air and in the phantom must be the same. The main attribute of a passive-feed network is that it provides partial shielding of the $E_z$ component of an electric field in a driven device. Figure 4C shows that the strongest $E_z$ field of a directly fed antenna originates from the middle (feed) point of the dipole. In the case of the passively fed dipole, the middle point of the feed is “shielded” from the sample by the passive element and the induced $E_z$ field in the phantom is reduced. From the surface current distribution at the antenna’s conductor facing the phantom (Figure 4A), the maximum current distribution for the directly fed meander is in the central section of the antenna. The magnitude of the surface current distribution for the passively fed meander is lower in value and more evenly distributed along the conductor. Therefore, the inductive electric field, related to the surface current distribution on the antenna’s radiating conductor, is lower for the passively fed meander. To confirm the hypothesis that the total electric field in a

FIGURE 7  In vivo TSE prostate images made with eight-channel body arrays of (upper row) directly fed and (lower row) passively fed meander dipoles with zoomed prostate images

FIGURE 8  Noise correlation matrices of four volunteers scanned with (upper row) directly fed and (lower row) passively fed array. The body mass indices of volunteers varied from 20.8 to 28.7 kg/m$^2$
sample is a combination of both conservative and inductive components, we placed a small shielding strip below a conventional actively fed dipole. We observed that the reductions in SAR were not as effective as those with the proposed passive-feed concept.

The inter-element coupling of the passively fed dipole (−15 to −18 dB) was found to be slightly less sensitive to in vivo loading conditions than that of the directly fed dipole (−20 to −28 dB), as reflected by the noise correlation coefficients shown in Figure 7, although the absolute values were higher. Again, these results can be explained in terms of the partial shielding effect of the passive element, which reduces slightly the coupling to the sample.

The actual in vivo SAR value depends on the specific phases applied to the individual channels. In this study, we presented simulated SAR values of directly and passively fed dipole arrays where the phases are adjusted to produce the maximum constructive interference in a prostate region. One question to be addressed in future research is whether or not the directly fed dipole antennas would be able to reach a similar SAR efficiency with a different phase setting if some of the superior B1+ efficiency were to be traded against avoiding local maxima in SAR.

5 CONCLUSIONS

This study has shown that the SAR efficiency of a dipole element used for high field MRI can be increased by using a passive- rather than an active-feeding network. Although the results shown here are specifically for a meander dipole, the passive-feed network approach can also be applied to different dipole geometries. For example, we have performed simulations on the SAR efficiency of a passively fed straight dipole and found similar results to those presented here in terms of the improvement in SAR efficiency. We anticipate that similar behaviour would be found for other types of dipole such as folded, snake, or the recently introduced dual mode. We also anticipate that a passive-feeding network would increase the SAR efficiency of electrically short dipoles, which are known to have high transmit efficiency but relatively low SAR efficiency, and are suitable, for example, for cardiac or pancreatic imaging.

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