Published in final edited form as:

Magn Reson Med. 2018 August ; 80(2): 822–832. doi:10.1002/mrm.27048.

A Rigid, Stand-off Hybrid Dipole and Birdcage Coil Array for 7T Body Imaging

Jan Paška^{1,2}, Martijn A. Cloos^{1,2}, and Graham C. Wiggins^{*,1,2}

¹Bernard and Irene Schwartz Center for Biomedical Imaging, Department of Radiology, New York University School of Medicine, New York, NY, USA

²Center for Advanced Imaging Innovation and Research (CAI²R), Department of Radiology, New York University School of Medicine, New York, NY, USA

Abstract

Purpose—To design a robust and patient friendly RF coil array (8-channel transmit and 16-channel receive) for cross-sectional body imaging at 7 T, and to improve our understanding of the combination of dipole and loop like elements for ultra high field strengths.

Methods—The hybrid coil array was optimized in EM-simulations. Considered array candidates were the dipole, loop and birdcage array. The winning design was constructed and the SNR was compared to a close fitting array at 3 T. Transmit and receive properties for different body sizes were assessed, and multi-parametric maps were acquired with the PnP-MRF method.

Results—The winning design consists of a dipole array for transceive combined with a birdcage array for receive only. The central SNR improved by a factor of 3 as compared to a 3T system with a local receive array. A transmit efficiency between 2.4 and $3.9 \mu\text{T}/\sqrt{kW}$, a SAR efficiency of 0.25 to $0.53 \mu\text{T}/\sqrt{W/kg}$, and a high SNR was achieved in the center for the targeted patient population.

Conclusion—The constructed coil array is easy to handle, safe, and patient friendly, allowing further development of abdominal imaging at 7 T. Quantitative MRI in the abdomen is possible using PnP-MRF using the designed coil array.

Keywords

Dipole antenna; 7T hybrid body coil; abdominal imaging; Birdcage array; parallel transmit; PnP-MRF

Introduction

Body imaging at 7 T promises a higher SNR (1), (2), (3), due to a larger magnetization and higher induced voltage in the RF coils. In addition, the localized and complimentary nature of the receive RF fields B_1^- help parallel imaging techniques such as SMASH (4), SENSE (5), or GRAPPA (6). The higher SNR in combination with the better parallel imaging

Corresponding Author: Jan Paska, PhD, Center for Biomedical Imaging, Department of Radiology, New York University School of Medicine, 660 1st Avenue, Room 401, New York, NY 10016, USA, jan.paska@nyumc.org.

*Deceased

performance at 7 T can be used to reduce acquisition time, which is especially helpful for body imaging that suffers from breathing motion, making high-resolution single breath hold images feasible. On the transmit side, the increased orthogonality of the transmit RF fields B_1^+ can be exploited in a similar way with parallel transmit technologies (7), (8) to achieve a targeted excitation or a more uniform flip-angle distribution.

Transmit RF body coil design at high field, however, remains a challenge. Unlike clinical systems operating at 1.5 and 3 Tesla, 7 Tesla systems are generally not equipped with a general purpose bore sized transmit coil. Wavelength effects, leading to non-uniform images (9), in combination with the skin-effect, leading to a low transmit efficiency (10), (11), pose a formidable challenge for body imaging at 7 Tesla.

For the last decade, the field has focused on close fitting elements for body imaging at 7 T. Radiative dipole antenna elements have proven to be the most common design (12), and more recently a combination of dipole and loop elements forming so-called hybrid elements (13, 14). These designs exhibit a high transmit and receive efficiency. Although this is sufficient to perform parallel transmission experiments targeting isolated regions, such as the prostate (12, 14) or heart (10), it remains extremely difficult to create an adequately uniform excitation needed for general purpose cross-sectional imaging applications, and only few groups have shown a large coverage in the body (15–19). In addition, meticulous coil placement may be necessary to avoid gaps in the excitation field between elements, which can impede the work-flow.

In this work, we present a 7 Tesla coil array for cross-sectional body imaging in mind. In anticipation of the need for a work horse coil to help further develop abdominal imaging, extra emphasis was placed on practicality and safety. To illuminate each location within the field-of-view we decided for a rigid coil array that is in between a close-fitting and in-bore body coil, which fits the 50th percentile of the male population. With this kind of coil array no tedious geometrical adjustment of coil elements during the setup is necessary for any patient size. In addition, we wanted to advance our understanding of the potential for a combination of dipole and loop like elements to improve the SNR at ultra-high field strengths. Based on previous theoretical (1) and experimental (12), (13) work, we explored different configurations of dipole antennas, loop coil elements, and birdcage arrays to find the optimal combination of eight transceiver channels complemented by an additional eight element receive only array. The methods and results section of the paper is therefore divided into two parts. The first part of each section focuses on the simulation work performed to identify the best coil configuration, while the second part focuses on the construction and experimental evaluation of the winning design.

Furthermore, we demonstrate quantitative body imaging with Plug-and-Play MRF (PnP-MRF) (20) using the hybrid body coil. This technique, based on principles of magnetic resonance fingerprinting (21), enables the rapid quantification of multiple tissue properties while simultaneously characterizing the B_1^+ field distribution. Therefore the B_1^+ bias can be removed automatically from the measurement, and the requirements for the transmit coil

array are more relaxed in terms of field uniformity as long as each area of the subject is illuminated sufficiently by at least one of the coil elements.

Methods

To facilitate the development of new strategies for abdominal imaging at 7 Tesla a reliable and safe workhorse body coil is needed that fits a substantial fraction of the population. We decided to limit ourselves to the 50th percentile of the male population (22), leading to an elliptical inner shape of the coil housing with semi-axes of 14.5 cm and 19 cm. To ensure load stability, the coil elements are placed 3 cm from the inner diameter. We decided to use an RF-shield for patient safety and reproducibility of the imaging experiments. The shield was placed as far away as possible from the coil elements (2 cm) to limit the decrease in efficiency and uniformity of the magnetic field distribution.

Given these boundary conditions, the coil array candidates (dipole-, loop-, and birdcage-array) will be optimized using the procedure described in the first part of the method section *Optimization of the Coil Array Design*. The construction, safety assessment, SAR efficiency, and experimental evaluation of the winning design will be outlined in the remainder of the *Methods* section, in the subsections *Construction*, *Safety Assessment and SAR Efficiency*, and *Imaging Experiments and Evaluation*, which precludes the revelation of the winning design in the *Result* section.

Optimization of the Coil Array Design

A full wave EM simulation software based on the finite element method (HFSS, ANSYS Inc., Canonsburg, PA, USA) was used to identify the optimal coil design. All simulations were performed at 297.2 MHz, corresponding to the proton larmor frequency at 7 Tesla. The numerical domain considered in the simulations was confined to a cylinder with a length of 1.2 m and a diameter of 69 cm, corresponding to the RF shield present in our gradient system. Consequently, the surface of the cylinder was modeled as a perfectly electrically conducting boundary condition and a radiating boundary condition was applied to the top and bottom faces. The coil conductors were 7 mm wide and modeled as a conducting boundary condition, with the conductivity of copper ($\sigma = 5.7 \cdot 10^7$ S/m). All coil designs were loaded by an elliptical phantom (semi-axes 14.5 cm and 19 cm, length 60 cm) with electrical properties similar to the average body composition at 7 T (relative permittivity $\epsilon_r = 32$, conductivity $\sigma = 0.4$ S/m). All lumped elements were replaced with 50 Ω ports, allowing the final values to be set in a post-processing step using co-simulation (23), (24). All coils were matched to 50 Ω . Losses for all lumped elements were included by assuming a quality factor of 300 for the inductors (25), the loss of the capacitors was estimated from a model provided by a manufacturer of multi-layer ceramic chip capacitors to be approximately $Q(C[pF]) \approx 6300 \cdot C[pF]^{-0.9}$ (AH series, Dielectric Laboratories Inc., Cazenovia, NY, USA). The simulations were performed on a dedicated workstation equipped with 512 GB of RAM and 2 CPUs with 10 cores each (Intel Xeon E5-2650 v3, 2.30 GHz), allowing 13 million tetrahedrons per simulation. We chose a direct solver with an adaptive mesh refinement solution process to refine the meshing of the numerical domain. The total simulation time of

the adaptive solution process did not exceed 15 mesh refinement passes and 50 hours per simulation.

The optimization of the coil array was divided into two parts. First the transmit performance was optimized, by finding an optimal 8 channel transceive array. In the second part the receive performance was optimized by adding an 8 channel receive only array. The conductors of the transmit and receive array were placed 2 and 2.2 cm from the RF shield, respectively. Three different basic array designs are considered in the optimization: dipoles, loops, and a birdcage array. The considered arrays for optimization are shown in Supporting Figure S1 a-c.

Transmit Optimization—To find the best possible dipole array the length of the dipole and the position of the shortening inductors on the antenna was varied in simulation. For this purpose, each dipole element was modeled by ten copper strips spaced 5 mm apart along the axis of the dipole (cumulative length of 50 cm). Nine ports were used to link the strips together. The central port was used for feeding. The effective antenna length could be adjusted by setting the appropriate ports to open or short in co-simulation. This way, the outer copper strips were disconnected, while the inner copper strips were enabled to radiate. Similarly, the position of the shortening inductance was set by replacing the short with an inductance, whose value was chosen such that the input impedance at the feed port is real. This way all dipole variants could be tested based on one simulation. The adjustment of geometry in co-simulation is possible for the dipole antenna, because it does not radiate along its axis (26). Therefore, adding conductors along its axis that are small compared to the wavelength do not affect the overall field distribution. Although, strictly speaking, this is only true in free space, the error induced by the violation of this condition is acceptably small for our setup.

To validate the proposed fast simulation method, we compared results obtained for dipoles with an element length of 30cm to a conventional simulation. As shown in Supporting Figure S2 and Supporting Table S1, a good agreement was observed between methods.

The loop coil array consisted of eight loop coil elements with equal dimensions. In this case, the optimization parameter was the width of the individual elements, which was varied from 5 cm to 15 cm in 2.5 cm steps. The length of the elements was kept fixed at an approximately optimal length of 20 cm, as determined based on Ultimate Intrinsic SNR (UISNR) analysis (1). The feeding ports were placed at the service end of the array. Capacitors were placed on the loop elements with a maximal distance of 7.5 cm between each other, resulting in 8 to 12 capacitors on each loop coil element. The capacitor values on the loop elements were chosen such that the input impedance at the feed port is real.

The birdcage array was implemented as a high-pass structure with 8 rods, with the feeding ports in the center of each rod. Here, the optimization variable is the length of the birdcage, which was varied from 15 cm to 25 cm in 2.5 cm steps. Three capacitors of equal value were placed between each of the rods in the end-rings. The value of the end-ring capacitors for each length was chosen in the co-simulation such that the transmit efficiency in the center of the phantom is maximized.

For the transmit performance we chose to optimize the power and SAR efficiency, defined as the B_1^+ in the center of the phantom per total input power and maximum 10g SAR, respectively. Although the final array will have parallel transmit capability, a maximum efficiency shim for the center voxel of the phantom will be used to evaluate the coil performance. In addition, we also looked at the extremes of the field uniformity defined by the ratio of the maximal to the minimal B_1^+ in the central axial slice, and the maximal coupling of the coil array.

Receive Optimization—After the optimal 8-channel transceiver array is found, the previously described array candidates were re-considered as additional receive only elements. The dipole and birdcage array elements were shifted to sit in between the transceive elements, the loop array elements remained on the same position (see Supporting Fig. S1 d-f). As before, the loop array elements were 20 cm long and the width was varied from 5 to 15 cm. Contrary to the transmit optimization, the length of the birdcage receive array was held fixed at 20 cm. Instead, the end-ring capacitance was varied to achieve either a minimal coupling or a maximal optimal SNR. For comparison we also considered the 8-channel transceive array itself without an additional receive only array as a candidate.

To check for a possible degradation of the transmit field by the presence of the receive coil array, we compared the transmit performance of the optimal hybrid array with the optimal transceive array.

For the receive performance we looked at the SNR in the center voxel for optimal reconstruction (27) and the maximal coupling in the receive coil array to estimate the effect of SNR degradation due to noise coupling from the input of a preamplifier to other channels.

Construction

The optimal coil array consists of eight dipole elements for transceive and an eight channel birdcage array for receive only, see Fig. 1a, whose dimensions are described in the *Results* section. The coil array with housing has an overall length of 34 cm. The top part of the coil array is removable for easier patient access and handling, see Fig. 1b. To ensure a reliable electrical connection between the top and bottom halves of the coil array, we used ODU connectors (ODU-USA, Camarillo, CA, USA) for the coaxial cables carrying the transmit and receive signals, and socket connectors for the RF shield and the end-ring conductors of the birdcage array. The coil array is interfaced with ODU plugs to the MR system.

The structure is held together by 2 elliptical 3D printed holders, that are connected by sixteen 30-cm long printed circuit boards (PCB) that act as an RF shield. The RF shield was slotted in 2 cm strips and shorted at larmor frequency with 370 pF capacitors to suppress gradient induced eddy currents. Plastic sheets were screwed on the inside and outside of the holders to serve as a housing. They protect the electronic circuitry, and ensure a distance of 3 cm between the coil elements and the patient for safety and load stability. The coil elements are mounted 2 cm from the RF-shield PCBs with distance holders, see Fig. 1c. We used hand wound inductors to shorten the dipole elements. Symmetry of the two arms of the dipole elements was ensured by tuning each inductor to the same value using a reflection

measurement at RF. We used 5.6 pF capacitors on the end-rings of the birdcage (11-series, Voltronics Corp., Denville, NJ, USA). Each coil element was connected at its feed-point to two wires which are routed behind the RF-shield. The wires of the dipole elements were connected to a transceiver board, and those of the birdcage to a receive board, mounted on the outer side of the RF shield, see Fig. 1d.

The receive and transceiver boards were based on previously published designs described by Reykowski et al. (28), and by Shajan et al. (29), respectively. On both, the receive and transceiver boards we used lattice baluns to transform the balanced voltage of the two wires to the unbalanced voltage on the PCBs. Furthermore, both boards contain preamplifiers (Stark Contrast, Erlangen, Germany), and a passive crossed pair of diodes to protect the preamplifiers in transmit mode. An active PIN diode on the receive board serves to detune the birdcage array in transmit mode in order not to disturb the transmit field. The transceiver boards contain a matching circuit consisting of two high power trimmer capacitors C_1 and C_2 in parallel on both sides of the balun (NMAJ19HV, Voltronics Corp., Denville, NJ, USA), see Fig. 1e. They allow for two degrees of freedom to match the dipole elements, while not disturbing the function of the balun considerably, within their range.

Special care was taken to adjust the lattice baluns, to minimize cable currents for a safe and stable operation independent of cable routing (30). For this purpose, a measurement fixture was built to characterize each balun with a three port measurement, see Fig. 1f. The transceiver or receive boards were excited at the unbalanced port, and the two output waves behind the lattice balun at the balanced port were measured against ground. For a proper functioning of the balun the two output waves on the balanced side need to have equal amplitude and be out of phase. This relationship was achieved by tuning the inductances of each lattice balun individually, to cancel parasitic effects of the PCB and variances in capacitor values.

The array elements were matched on an approximately elliptically shaped phantom mimicking the loading of a large patient (35×25 cm, length 45 cm, $\epsilon_r = 70$, $\sigma = 0.60$ S/m). The transmit elements consisting of the dipole sub-array were matched to 50 Ω in transmit mode. The elements of the 16-channel receive array consisting of the dipole and birdcage sub-arrays were matched to the optimal noise impedance of the preamplifiers of 75 Ω in receive mode. Preamplifier decoupling was adjusted using appropriate phase shifters. The matching condition was kept constant for all imaging experiments.

Safety Assessment and SAR Efficiency

A safe power limit for the imaging experiments was derived from EM-simulations using three different body models representing the target population (31). The used body models were a female body model (32), a male body model (33), and the male body model scaled by a factor of 1.1. The winning hybrid body coil was simulated as described in the *Optimization of the Coil Array Design* subsection. The excitation ports of the dipole array were modeled as 50 Ω ports and matched in post-processing using co-simulation. The capacitors and inductors were modeled as lumped elements, and the detuned ports of the birdcage array were left open. The effective SAR distribution for the MRF sequence was computed in MATLAB by averaging the quadrature and the gradient mode SAR distributions. For the

computation of the maximum input power into the array we chose the SAR limit for local transmit coils for the trunk in normal operation mode (34) of 10 W/kg, and an additional safety factor of 5 to account for uncertainties.

Imaging Experiments and Evaluation

The transmit and receive performance of the array with respect to three different body sizes, with body mass indexes of 18 (S), 23 (M), and 28 (L) was performed.

To measure the reflection and coupling for the different loading conditions in transmit mode, we measured the scattering matrices at 297.2 MHz on the bench with an 8-port PXIe network analyzer (Keysight Technologies Inc., Santa Rosa, CA, USA). The deviations of the scattering matrix of the coil array when placed inside the MR-scanner are expected to be negligible, due to the continuous RF-shield covering the coil array.

Imaging experiments were performed at 7 Tesla using a MAGNETOM MR scanner (Siemens Healthcare, Erlangen, Germany) equipped with 8 independent transmit amplifiers. An equal amplitude quadrature RF shim was found by aligning the RF phase in the center of a body phantom. This shim was kept constant for all 7T experiments. All 7T experiments were performed in a central slice of the abdomen for all body sizes.

Cross-sectional abdominal imaging was performed using a low-power PnP-MRF sequence designed for single breath hold imaging (35), which enables the mapping of proton density (PD), longitudinal relaxation time (T_1), and transmit efficiency (B_1^+). The sequence consists of 4000 excitations with driving voltages ranging between 0 V and 40 V per channel. The RF shim is varied between quadrature and gradient mode (90° phase increment per channel) every repetition time (TR). The image was acquired with a acquisition time (T_{aq}) of 32 s, matrix size of 128×128 , Field Of View (FOV) of 420×420 mm, and a slice thickness of 5 mm. The maps were reconstructed as shown in (35).

To assess the SNR at 7 T single breath hold gradient echo (GRE) images were acquired with the same matrix size, FOV, slice thickness, and slice as the PnP-MRF experiments. A T_{aq} of 26 s, TR of 200 ms, echo time (TE) of 5 ms, flip-angle (FA) of 30°, and a bandwidth (BW) of 800 Hz/pixel were used for the acquisition. Noise data was acquired to compute the noise correlation matrix. Optimal SNR reconstruction was used to compute a raw SNR map (36). The B_1^+ and T_1 maps were used to correct the effects of excitation non-uniformities and T_1 saturation effects in the raw SNR map, from heron forth we will refer to these as “SNR maps”.

For comparison, SNR maps were also obtained with commercially available coils at 3 Tesla using a Biograph mMR scanner (Siemens Healthcare, Erlangen, Germany) equipped with a behind-the-covers transceive body coil. Two SNR maps in the medium sized volunteer were computed based on the method described above for 7 T. For both maps we acquired B_1^+ , and T_1 -maps with the PnP-MRF method, GRE and noise data scans with the same imaging parameters as for the 7T imaging experiments. For the first SNR map the body coil was used in transceive mode. For the second SNR map we used the body coil for transmit only and a

local receive coil array (Flex-Array, Siemens Healthcare, Erlangen, Germany) consisting of a 6-channel array on top of the subject combined with 12 channels from the spine array integrated into the patient table.

Results

EM-Simulation: Optimizations and Safety

The optimal transmit RF coil array based on our optimizations is the dipole array consisting of 30 cm long elements with shortening inductors 5 cm from the feed point. This array has a high power efficiency of $5.5\mu\text{T}/\sqrt{\text{kW}}$ compared to the other evaluated coil array designs, a good SAR efficiency of $0.37\mu\text{T}/\sqrt{\text{W/kg}}$, a low ratio of the maximal to minimal B_1^+ of 17, and a low maximal coupling of -20 dB, this is shown in Figure 2. None of the considered loop or birdcage arrays beat optimal dipole array in terms of the considered optimization parameters, see Tables 2 and 1. The transmit field efficiencies in the central transversal plane of the optimal Dipole, Birdcage and Loop array driven by a maximum efficiency shim are shown in Supporting Figure S3.

The optimal receive only coil array that is added to the transceive dipole array is the birdcage array, that is optimized for lowest coupling (Table 3). It has a high relative sum-of-square SNR of 0.96, a high relative optimal SNR of 0.97, and a low maximal coupling of -12 dB. This configuration offers a good detuning in transmit mode. The transmit efficiency of the dipole array is degraded by only 1 % by the presence of the detuned birdcage array, as found experimentally.

There were four other promising candidate receive configurations. The 16-channel dipole array has a slightly better SNR performance, but a slightly higher coupling, and is easier to construct, but we found that the receive dipole array is more difficult to detune in transmit mode. The loop arrays with 10 cm and 12.5 cm wide loop elements were also promising candidates with similar receive performance, but the higher coupling makes the array more difficult to tune in the receive case. The birdcage array optimized for maximal optimal SNR was another promising candidate. It has a 3 % higher optimal SNR as compared to the chosen design, but we expect an SNR degradation in the experiment, and a difficult tuning of the array in the receive case due to the high coupling.

Safety Assessment and SAR Efficiency

A power limit of 10 W was set based on the simulated SAR values, see Table 4. The simulated SAR efficiency for the maximum power efficiency shim in the center voxel is $0.53\mu\text{T}/\sqrt{\text{W/kg}}$ for the female body model, and 0.36 and $0.25\mu\text{T}/\sqrt{\text{W/kg}}$ for the male and the scaled male body model respectively.

Imaging Experiments

The measured B_1^+ and T_1 normalized SNR (for brevity, we will refer to this quantity as SNR) of a body coil at 3T, a close coupling receive coil array at 3T, and the hybrid body coil at 7T are shown in Figure 3. The central SNR of the hybrid body coil array averaged over a square

with a side length of 9 voxels was ten- and three-fold higher and the SNR in the periphery was 28- and 4-times higher compared to the SNR of a body coil at 3T or a local receive coil array at 3T, respectively.

The transmit properties of the hybrid coil array for three different body sizes is shown in Figure 4. The transmit efficiency per 1 kW total input power in the center of the imaging subject averaged over the same ROI as before, decreases monotonically from 3.9 to $2.4\mu T/\sqrt{kW}$ as the body mass index increases, see first column of Figure 4. The scattering matrices of the hybrid array in transmit mode are shown in the second column of Figure 4. The highest reflection coefficients were -15 (S), -15 (M), and -12 dB (L) and the highest coupling was -12 (S), -12 (M), and -14 dB (L) for the different patient sizes.

The receive properties of the hybrid coil array for three different body sizes are shown in Figure 5. The central SNR decreases monotonically from 1100 to 630 from the small to the large subject, see first column of Figure 5, and the highest noise correlation coefficient decreases monotonically from 56 to 50% from the small to the large subject (second column of Fig. 5).

Robust mapping of proton density, longitudinal relaxation (T_1) and the transmit fields was possible in the chosen patient population using the PnP-MRF method, see Figure 6. Two transmit modes, the quadrature, and gradient mode were sufficient in all patient sizes to avoid nulls in the transmit field by filling each others voids.

Discussion

An extensive optimization with 3D full wave EM simulations of a hybrid coil array for body imaging is presented, that can be a guide for future construction and optimizations of coil arrays. A new and fast method for optimizing the length and inductor position of dipole arrays was shown, exploiting the fact that the dipole antenna radiates only along its axis. This approach, based on co-simulation, not only optimizes the lumped element distribution, but also the length of the dipole antennas using only a single EM-simulation run.

It was found that among all array candidates an eight channel dipole array is the best choice for transmission, it achieves the highest power efficiency and a good SAR efficiency. This result is confirmed by new theoretical findings obtained with the UISNR framework (37), which predict that, at high-field strengths, the z-directed currents form the dominant contribution to the maximal possible SNR obtainable from the center voxel in a cylindrical load. Among the considered options, we found that the receive performance was best bolstered by adding an eight channel receive only birdcage array.

The dipole coil array has a maximum of power efficiency for an antenna length of 30 cm. This can be explained by two counteracting effects. Shorter antennas focus the transmit field more which increases the power efficiency in the central voxel. However, the inductors inserted to shorten the dipole add additional losses (the shortening inductance, and the lattice balun), which diminish the net gain provided by a more focused field.

The loop coil array exhibits a maximum power efficiency when the elements are 10 cm wide. This can also be explained by two counteracting effects. Smaller loop elements have a lower transmit efficiency in deeper regions. On the other hand, larger loop elements couple more strongly when placed in an array configuration, which decreases the transmit efficiency due to scattered power.

The birdcage array shows a maximum of power efficiency for a length of 20 cm. This is not surprising, as the birdcage array can be considered as a special case of the loop array, whose optimal length is also 20 cm as predicted by UISNR simulations.

The constructed array behaves as expected, and outperforms clinical body imaging at 3T in terms of SNR by a factor 3. The coupling between elements decreases with BMI, as larger subjects are closer to the coil elements and the incoming EM-wave is attenuated by the tissue rather than scattered into the other channels. Similarly, the power efficiency and SNR in the center voxel decrease, because the EM-waves must penetrate deeper into the lossy tissue to reach the central voxel of large subjects.

The hybrid coil array shows potential as a workhorse coil array for body imaging at 7 T. The array is rigid and can be split, for easy handling and patient access. A high efficiency, homogeneity, and SNR for a broad range of body sizes was achieved, without the need to rematch the coil elements. Currently, we decided to limit the subject size to the 50th percentile of the male population. The limit of this coil concept is the scanner's bore. Larger subjects could possibly be accommodated by building the coil into the space between the gradient and the covers of the bore (38). It should be noted, however, that reliable cross-sectional imaging in obese subjects can be challenging even in clinical systems operating at 3 Tesla (39), (40). Special care was taken to avoid cable currents, by a precise adjustment of the lattice baluns. These baluns together with the RF shield make the operation of the array stable and safe (30).

Furthermore, it was shown that two modes of the coil array, the quadrature and gradient mode found in a phantom, were sufficient for a robust multi parametric mapping using PnP-MRF across a range of different subject sizes. Although this suggests that a calibration free work-flow for quantitative abdominal MR may be possible, further work is necessary to improve the spatial resolution, eliminate breath holds, and identify more complementary RF shim configurations.

The idea to use two complementary modes to mitigate signal voids during body imaging experiments at 7 T was first proposed by Orzada et al (19). This technique provides an attractive workflow to obtain contrast weighted images with greatly improved signal distribution, but some residual contrast variations may remain due to the non-uniform flip angles interwoven into the acquisition. PnP-MRF can overcome these B_1^+ induced contrast variations through a model-based reconstruction process. In principle the heterogeneous RF fields produced by each individual transmit channel could also be used directly (41). With the power peak currently available, however, we found it difficult to obtain sufficiently high flip-angles in the deeper regions of the body when pulsing on individual channels.

Compared to other proposed designs, the hybrid body coil exhibits a lower transmit efficiency. This was a conscious design choice. By allowing a greater distance between the coil elements and imaging object and incorporating an RF shield, we traded some transmit efficiency in favor of increased user friendliness, and safety. The hybrid body coil array achieves transmit efficiencies of 2.4 to $3.9\mu\text{T}/\sqrt{\text{kW}}$. These efficiencies were experimentally measured in the center voxel of cross-sectional images at the height of the liver of three different subjects with a BMI ranging from 18 to 28. Ertürk et al achieved 5.4 to $6.2\mu\text{T}/\sqrt{\text{kW}}$ experimentally in the prostate of subjects with a comparable BMI range using a close fitting combined loop-dipole array (14). While Raaijmakers et al, using a close fitting dipole array, measured 3 to $4.5\mu\text{T}/\sqrt{\text{kW}}$ in the prostate of subjects with a similar BMI range (18). However, the SAR-efficiency of $0.36\mu\text{T}/\sqrt{\text{W/kg}}$ is comparable to that reported by Ertürk et al in (14). They reached 0.46 and $0.37\mu\text{T}/\sqrt{\text{W/kg}}$ for a 16-elements loop-dipole array, and an 8-elements dipole array, for a maximum power efficiency shim in a comparable anatomical human body model.

Conclusions

The hybrid coil array described in this work shows great potential as a workhorse coil array for body imaging at 7 T that is easy to use, and allows further development of abdominal imaging at 7 T. The optimal array configuration was found with extensive optimizations that can be a guide for future coil array design at high field strengths. Quantitative abdominal imaging of the entire transversal slice at 7 T without subject specific calibration or pulse-design is possible using the PnP-MRF method in combination with the developed coil array.

Future work will include a more rigorous safety assessment based on virtual observation points for a less conservative power limit. The imaging performance of PnP-MRF techniques for body imaging at 7T can now be studied, allowing further optimization of the sequence design and the RF shim modes configuration.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

Our dear co-author, mentor, and friend Graham C. Wiggins tragically passed away before the publication of this work and leaves a gap in our hearts.

We would like to thank Bei Zhang, Gilliam Haemer, and Jerzy Walczyk for their help constructing the array and editing.

This work was supported in part by NIH R01 AR070297 and NIH R01 EB011551 and was performed under the rubric of the Center for Advanced Imaging Innovation and Research (CAI2R, www.cai2r.net), a NIBIB Biomedical Technology Resource Center (NIH P41 EB017183).

References

1. Lattanzi R, Sodickson DK. Ideal current patterns yielding optimal signal-to-noise ratio and specific absorption rate in magnetic resonance imaging: computational methods and physical insights. *Magn Reson Med*. 2012; 68(1):286–304. [PubMed: 22127735]

2. Wiesinger F, Van de Moortele PF, Adriany G, De Zanche N, Ugurbil K, Pruessmann KP. Parallel imaging performance as a function of field strength—an experimental investigation using electrodynamic scaling. *Magn Reson Med*. 2004; 52(5):953–964. [PubMed: 15508167]
3. Vaughan JT, Garwood M, Collins CM, Liu W, DelaBarre L, Adriany G, Andersen P, Merkle H, Goebel R, Smith MB, Ugurbil K. 7T vs. 4T: RF power, homogeneity, and signal-to-noise comparison in head images. *Magn Reson Med*. 2001; 46(1):24–30. [PubMed: 11443707]
4. Sodickson DK, Manning WJ. Simultaneous acquisition of spatial harmonics (SMASH): fast imaging with radiofrequency coil arrays. *Magn Reson Med*. 1997; 38(4):591–603. [PubMed: 9324327]
5. Pruessmann KP, Weiger M, Scheidegger MB, Boesiger P. SENSE: sensitivity encoding for fast MRI. *Magn Reson Med*. 1999; 42(5):952–62. [PubMed: 10542355]
6. Griswold MA, Jakob PM, Heidemann RM, Nittka M, Jellus VI, Wang J, Kiefer B, Haase A. Generalized autocalibrating partially parallel acquisitions (GRAPPA). *Magn Reson Med*. 2002; 47(6):1202–1210. [PubMed: 12111967]
7. Katscher U, Börner P, Leussler C, van den Brink JS. Transmit SENSE. *Magn Reson Med*. 2003; 49(1):144–150. [PubMed: 12509830]
8. Zhu Y. Parallel excitation with an array of transmit coils. *Magn Reson Med*. 2004; 51(4):775–84. [PubMed: 15065251]
9. Collins CM, Liu W, Schreiber W, Yang QX, Smith MB. Central brightening due to constructive interference with, without, and despite dielectric resonance. *J Magn Reson Imaging*. 2005; 21(2): 192–6. [PubMed: 15666397]
10. Vaughan JT, Snyder CJ, DelaBarre LJ, Bolan PJ, Tian J, Bolinger L, Adriany G, Andersen P, Strupp J, Ugurbil K. Whole-body imaging at 7T: preliminary results. *Magn Reson Med*. 2009; 61(1):244–8. [PubMed: 19097214]
11. Orzada, S., Bitz, AK., Kraff, O., Oehmigen, M., Gratz, M., Johst, S., Völker, MN., Rietsch, SHG., Flöser, M., Fiedler, T., Shooshtary, S., Solbach, K., Quick, HH., Ladd, ME. Proc 24th Annu Meet. Vol. 167. ISMRM; Singapore, Singapore: 2016. A 32-channel integrated body coil for 7 Tesla whole-body imaging.
12. Raaijmakers AJE, Ipek O, Klomp DWJ, Possanzini C, Harvey PR, Lagendijk JJW, van den Berg CAT. Design of a radiative surface coil array element at 7 T: the single-side adapted dipole antenna. *Magn Reson Med*. 2011; 66(5):1488–1497. [PubMed: 21630342]
13. Wiggins, GC., Zhang, B., Cloos, M., Lattanzi, R., Chen, G., Lakshmanan, K., Haemer, G., Sodickson, DK. Proc 21st Annu Meet. Vol. 2737. ISMRM; Salt Lake City, USA: 2013. Mixing loops and electric dipole antennas for increased sensitivity at 7 Tesla.
14. Ertürk MA, Raaijmakers AJE, Adriany G, Ugurbil K, Metzger GJ. A 16-channel combined loop-dipole transceiver array for 7 Tesla body MRI. *Magn Reson Med*. 2017; 77(2):884–894. [PubMed: 26887533]
15. Wu X, Schmitter S, Auerbach EJ, Ugurbil K, Van de Moortele PF. Mitigating transmit B₁ inhomogeneity in the liver at 7T using multi-spoke parallel transmit RF pulse design. *Quant Imaging Med Surg*. 2014; 4(1):4–10. [PubMed: 24649429]
16. Philips BWJ, Fortuin AS, Orzada S, Scheenen TWJ, Maas MC. High resolution MR imaging of pelvic lymph nodes at 7 Tesla. *Magn Reson Med*. 2016; 1028:1020–1028. epub ahead of print.
17. Orzada S, Maderwald S, Poser BA, Johst S, Kannengiesser S, Ladd ME, Bitz AK. Time-interleaved acquisition of modes: An analysis of SAR and image contrast implications. *Magn Reson Med*. 2012; 67(4):1033–1041. [PubMed: 21858867]
18. Raaijmakers AJE, Italiaander M, Voogt IJ, Luijten PR, Hoogduin JM, Klomp DWJ, van den Berg CAT. The fractionated dipole antenna: A new antenna for body imaging at 7 Tesla. *Magn Reson Med*. 2015; 00:1–9.
19. Orzada S, Maderwald S, Poser BA, Bitz AK, Quick HH, Ladd ME. RF excitation using Time Interleaved Acquisition of Modes (TIAMO) to address B₁ inhomogeneity in high-field MRI. *Magn Reson Med*. 2010; 64(2):327–333. [PubMed: 20574991]
20. Cloos MA, Knoll F, Zhao T, Block KT, Bruno M, Wiggins GC, Sodickson DK. Multi-parametric imaging with heterogeneous radiofrequency fields. *Nat Commun*. 2016; 7:1–10.

21. Ma D, Gulani V, Seiberlich N, Liu K, Sunshine JL, Duerk JL, Griswold MA. Magnetic resonance fingerprinting. *Nature*. 2013; 495(7440):187–192. [PubMed: 23486058]
22. Tilley, AR. Henry Dreyfuss Associates. The Measure of Man and Woman: Human Factors in Design. New York, USA: John Wiley & Sons; 2002.
23. Kozlov M, Turner R. Fast MRI coil analysis based on 3-D electromagnetic and RF circuit co-simulation. *J Magn Reson*. 2009; 200(1):147–152. [PubMed: 19570700]
24. Paska, J., Froehlich, J., Brunner, DO., Pruessmann, KP., Vahldieck, R. Proc 17th Annu Meet. Vol. 3038. ISMRM; Honolulu, USA: 2009. Field Superposition Method for RF coil design.
25. Chen G, Collins CM, Sodickson DK, Wiggins GC. A method to assess the loss of a dipole antenna for ultra-high-field MRI. *Magn Reson Med*. 2017; 00
26. Ramo, S., Whinnery, JR., Van Duzer, T. Fields and Waves in Communication Electronics. 3rd. New York, USA: John Wiley & Sons; 1994.
27. Kellman P, McVeigh ER. Image reconstruction in SNR units: a general method for SNR measurement. *Magn Reson Med*. 2005; 54(6):1439–47. [PubMed: 16261576]
28. Reykowski A, Wright SM, Porter JR. Design of Matching Networks for Low Noise Preamplifiers. *Magn Reson Med*. 1995; 33(6):848–852. [PubMed: 7651124]
29. Shajan G, Hoffmann J, Budde J, Adriany G, Ugurbil K, Pohmann R. Design and evaluation of an RF front-end for 9.4 T human. MRI *Magn Reson Med*. 2011; 66(2):596–604. [PubMed: 21381102]
30. Brunner, DO., Zanche, N., Froehlich, J., Baumann, D., Pruessmann, KP. Proc 15th Annu Meet. Vol. 15. Vol. 448. ISMRM; Berlin, Germany: 2007. A symmetrically fed microstrip coil array for 7T.
31. Hoffmann J, Henning A, Giapitzakis IA, Scheffler K, Shajan G, Pohmann R, Avdievich NI. Safety testing and operational procedures for self-developed radiofrequency coils. *NMR Biomed*. 2016; 29(9):1131–1144. [PubMed: 25851551]
32. Modiri A, Kiasaleh K, Chandrasah S. Characterizing a proposed sixteen-element array antenna designed for microwave imaging of breast cancer. *BioWireless 2013 - Proc 2013 IEEE Top Conf Biomed Wirel Technol Networks, Sens Syst - 2013 IEEE Radio Wirel Week, RWW 2013*. 2013:61–63.
33. Ketterl TP, Arrobo GE, Sahin A, Tillman TJ, Arslan H, Gitlin RD. Vivo Wireless Communication Channels. *IEEE 13th Annu Wirel Microw Technol Conf*. 2012; 13:1–3.
34. International Electrotechnical Commission (IEC). International standard, medical equipment -part 2: particular requirements for the safety of magnetic resonance equipment for medical diagnosis, 2nd revision, 60601-2-33. Technical report, International Electrotechnical Commission (IEC); Geneva: 2002.
35. Cloos, MA., Paska, J., Zhou, T., Yu, Z., Wiggins, GC., Sodickson, DK. Proc 25th Annu Meet. Vol. 1125. ISMRM; Honolulu, USA: 2017. Abdominal Imaging with Heterogeneous Radiofrequency Fields at 7 Tesla.
36. Roemer PB, Edelstein WA, Hayes CE, Souza SP, Mueller OM. The NMR phased array. *Magn Reson Med*. 1990; 16(2):192–225. [PubMed: 2266841]
37. Sodickson, DK., Wiggins, GC., Chen, G., Lakshmanan, K., Lattanzi, R. Proc 24th Annu Meet. Vol. 389. ISMRM; Singapore, Singapore: 2016. More than meets the eye: The mixed character of electric dipole coils, and implications for high-field performance.
38. Orzada S, Bitz AK, Johst S, Gratz M, Völker MN, Kraff O, Abuelhaija A, Fiedler TM, Solbach K, Quick HH, Ladd ME. Analysis of an Integrated 8-Channel Tx/Rx Body Array for Use as a Body Coil in 7-Tesla MRI. *Front Phys*. 2017; 5(June):1–10.
39. Bernstein MA, Huston J, Ward HA. Imaging artifacts at 3.0T. *J Magn Reson Imaging*. 2006; 24(4): 735–746. [PubMed: 16958057]
40. Merkle EM, Dale BM. Abdominal MRI at 3.0 T: The Basics Revisited. *Am J Roentgenol*. 2006; 186(6):1524–1532. [PubMed: 16714640]
41. Cloos, MA., Wiggins, C., Wiggins, G., Sodickson, D. Proc 22nd Annu Meet. Vol. 542. ISMRM; 2014. Plug and Play Parallel Transmission at 7 and 9.4 Tesla based on Principles from MR Fingerprinting.

Figure 1.

Construction of the hybrid coil array. a) 3D schematic of the coil array, 8-channel transceive dipole array combined with an 8-channel Rx only birdcage array, with RF-shield. b) Splittable structure, ODU plugs to interface with the system. c) Mechanical design, 3D printed holders, and wires connecting the RF coil elements to the PCBs behind the RF-shield. d) transceive and Rx boards for the dipole or birdcage arrays behind the RF shields. e) circuit diagram of the matching and lattice balun for the dipole elements. f) RF measurement fixture on the transceive board to fine tune the lattice baluns. g) Hybrid coil array loaded by human in the MR scanner.

Figure 2.

Transmit properties of the 8-channel dipole coil array for different lengths and positions of shortening inductance. The array was loaded with an elliptical phantom representing the largest expected patient size, and driven such that the transmit field is maximized in the center voxel. a) power efficiency, transmit field in the center voxel per input power [$\mu\text{T}/\sqrt{\text{kW}}$]. b) SAR efficiency, transmit field in the center voxel per maximal 10-g SAR [$\mu\text{T}/\sqrt{\text{W/kg}}$]. c) Field homogeneity, ratio of maximal to the minimal transmit field in the central slice. d) Maximal coupling of the array [dB].

Figure 3.

SNR comparison body imaging at 3 T and 7 T using the hybrid body coil array. SNR maps are shown in dB. Body Coil: Body coil of the 3 T system is used in transceive mode. Local Coil Array: Body coil is used for transmit only, and a local receive coil array is used for reception at 3 T. Hybrid Coil Array: 8-channel dipole array is used for transceive and a 8-channel birdcage array is used in addition for Rx only at 7 T. The same imaging parameters were used for all experiments: $T_{aq}=26$ s, FOV= 420×420 mm, matrix size 128×128 , slice thickness 5 mm, TR=200 ms, TE=5 ms, FA=30°, BW=800 Hz/pixel.

Figure 4.

Transmit properties of the hybrid coil array for different body sizes (body mass index of 18, 23, and 28). B_1^+ : transmit efficiency in $\mu T/\sqrt{kW}$ for quadrature drive. $|S|$: scattering matrix of the 8 channel Tx dipole coil array in dB measured on the bench. Maps were obtained with PnP-MRF imaging experiments, $T_{aq}=32$ s, $FOV=420 \times 420$ mm, 128×128 matrix, slice thickness 5 mm.

Figure 5.

Receive properties of the hybrid coil array for different body sizes (body mass index of 18, 23, and 28). SNR: SNR in dB with optimal reconstruction. Ψ : Measured noise correlation matrix, of the 16 channel hybrid array. The first eight channels correspond to the dipole array, and the last eight channels to the birdcage coil array. Imaging parameters were: $T_{aq}=26$ s, FOV= 420×420 mm, matrix size 128×128 , slice thickness 5 mm, TR=200 ms, TE=5 ms, FA=30°, BW=800 Hz/pixel.

Figure 6.

Multiparametric maps obtained from PnP experiments at 7 T for three different body sizes (body mass index of 18, 23, and 28). Proton density weighted by the receive sensitivity profile (PD), longitudinal relaxation time (T_1), and transmit efficiency (B_1^+) maps for the quadrature and gradient mode were mapped. $T_{aq}=32$ s, 128×128 matrix, FOV= 420×420 mm, slice thickness 5 mm.

Table 1

Transmit properties of the 8-channel loop coil array for different widths of the loop elements, the lengths are 20 cm. The array was loaded with an elliptical phantom representing the largest expected patient size, and driven such that the transmit field is maximized in the center voxel. In the first column the power efficiency, transmit field in the center voxel per input power. In the second column SAR efficiency, transmit field in the center voxel per maximal 10-g SAR. In the third column field homogeneity, ratio of maximal to the minimal transmit field in the central slice. In the last column the maximal coupling of the array.

W [cm]	$\frac{B_1^+(0)}{\sqrt{P_{in}}} [\frac{\mu T}{\sqrt{kW}}]$	$\frac{B_1^+(0)}{\max(SAR_{10g})} [\frac{\mu T}{\sqrt{W/kg}}]$	$\frac{\max(B_1^+)}{\min(B_1^+)} _{z=0}$	max(S_{ij}) [dB]
5	3.5	0.19	500	-17
7.5	4.1	0.23	490	-16
10	4.2	0.23	160	-12
12.5	4.1	0.27	620	-9.2
15	3.2	0.29	490	-6.4

Table 2

Transmit properties of the 8-channel birdcage coil array for different lengths. The array was loaded with an elliptical phantom representing the largest expected patient size, and driven such that the transmit field is maximized in the center voxel. In the first column the power efficiency, transmit field in the center voxel per input power. In the second column SAR efficiency, transmit field in the center voxel per maximal 10-g SAR. In the third column field homogeneity, ratio of maximal to the minimal transmit field in the central slice. In the last column the maximal coupling of the array.

L [cm]	$\frac{B_1^+(0)}{\sqrt{P_{in}}} [\frac{\mu T}{\sqrt{kW}}]$	$\frac{B_1^+(0)}{\max(SAR_{10g})} [\frac{\mu T}{\sqrt{W/kg}}]$	$\frac{\max(B_1^+)}{\min(B_1^+)} _{z=0}$	max(S_{ij}) [dB]
15	4.6	0.30	49	-11
17.5	4.8	0.33	89	-11
20	4.9	0.36	123	-11
22.5	4.8	0.36	54	-11
25	4.6	0.35	29	-11

Table 3

Receive properties of different coil arrays. Optimal SNR, sum-of-squares SNR in the center of a body phantom in arbitrary units and maximal coupling are shown. The considered coil arrays are the winning transceive 8-channel dipole array and additional 8-channel dipole array, loop arrays of different widths, and birdcage array with different end-ring capacitors. The properties of the winning transceive coil array alone are shown for comparison.

	SNR_{opt}	SNR_{sos}	$\max(S_{ij})$ [dB]
dipole 8-ch	0.88	0.88	-20
dipole 16-ch	0.97	0.97	-11
dipole + loop W=5 cm	0.92	0.92	-15
dipole + loop W=10 cm	0.97	0.96	-11
dipole + loop W=12.5 cm	0.99	0.97	-9
dipole + loop W=15 cm	0.97	0.90	-7
dipole + birdcage, min coup	0.97	0.96	-12
dipole + birdcage, max SNR	1	0.83	-8

Table 4

Simulated maximal averaged 10g SAR values for quadrature, gradient, and MRF excitations for three different body models, to approximate the experimental loading conditions. The winning array design was simulated in its transmit configuration loaded by three different body models: a female body model, a male body model, and the male body model scaled by a factor of 1.1.

$\max(\text{SAR}_{10\text{g}})$ [W/kg]	quadrature	gradient	MRF
Female	0.11	0.19	0.14
Male	0.13	0.28	0.20
1.1*Male	0.16	0.20	0.19