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Traveling-wave meets standing-wave: A simulation study using pair-of-transverse-dipole-ring (PTDR) coils for adjustable longitudinal coverage in ultra-high field MRI

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Abstract:

1.

At ultrahigh fields (B_0 7T), it is challenging to cover a large field of view using single-row conventional RF coils (standing wave resonators) due to the limited physical dimensions. In contrast, traveling wave approaches can excite large fields of view even using a relatively simple hardware setup, but suffer from poor efficiency and high local specific absorption rate in non-imaged regions. In this study, we propose and numerically analyze a new coil which combines the concept of traveling wave and standing wave. The new coil consists of a Pair of Transverse Dipole Rings (PTDR) whose separation is adjusted according to the desired imaging coverage. The PTDR coil was validated using electromagnetic (EM) simulations in phantoms and human leg models, which showed that coverage can be as long as 60 cm. When the coverage of the PTDR coil was shortened to 20 cm to cover the knees only, its transmit and SAR efficiencies were 84% and 37% higher than those of the 50 cm coverage, respectively.

Keywords

High-field MRI; RF coil; traveling wave; standing-wave; transmit efficiency; SAR

2. Introduction:

Ultrahigh field (B_0 7T) MRI systems can offer higher signal-to-noise ratio (SNR), enhanced spatial resolution and higher susceptibility contrast for human imaging [1–3]. However, several challenges arise in RF excitation, such as inhomogeneous transmit fields (B_1^+), increased power requirements and high specific absorption rate (SAR) [2,3]. Unlike clinical 1.5T and 3T scanners, 7T scanners currently are not equipped with body coils. For MR imaging with relatively short coverage in the z-dimension (head, knees and prostate), local transmit coils such as microstrip and L/C loop resonators are commonly used [4–7].

These coils resonate at the Larmor frequency and generate standing wave RF fields. Due to their limited physical dimensions at ultrahigh fields, it is technically challenging to use them to image a long area such as the torso or the legs.

One solution to this large field of view (FOV) challenge is the travelling wave approach, as demonstrated in Brunner et al [8]. For 7T human scanners, the RF shield of the scanner bore (~60 cm diameter) forms a cylindrical waveguide that supports the lowest transverse electric field (TE11) mode. Given this waveguide effect, traveling wave RF fields along the z-direction can be generated using either remote antennae [8] or local loop resonators inside the bore [9,10]. Compared to the conventional resonators, however, traveling wave excitation suffers from poor transmit efficiency and high SAR in non-imaged regions [11].

If a pair of traveling wave antennas are positioned inside the bore and orientated face-to-face along the length of the bore, they may generate a standing wave in the space between them. In Brunner et al [8], the authors mentioned that a pair of patch antennas must be positioned outside the bore to avoid the standing wave effect. In Wiggins et al [10], the authors demonstrated that a pair of orthogonal loops can generate a standing wave in the space between them. However, while both patch antennas and loop resonators can generate standing waves across long FOVs in the bore, they are not well suited to this application. Specifically, transmit (B_1^+) fields derive from the component of the transverse magnetic field that is polarized in the correct direction for spin excitation, $B_1^+ = (B_x + iB_y)/2$. The patch antenna has large solid plates and can only be placed at the ends of the MR bore, so it is impossible for the RF signals from the two patch antennas to meet each other in the presence of a human body. Although the orthogonal loop resonator can be positioned inside the bore, it is a near-field resonator and is not efficient for transmitting RF power through wave propagation when it is not positioned in close proximity to the body. As shown in Wiggins et al [10], the B_1^+ field generated by eight loops is approximately 10 dB lower compared to a conventional traveling wave setup.

In this study, we propose a novel coil design combining traveling wave and standing wave approaches, aiming to obtain the advantages of both. The new coil consists of a **P**air of **T**ransverse **D**ipole **R**ings (PTDR), whose gap is adjusted to vary imaging coverage in the z-dimension. In contrast to these approaches, the PTDR coil has compact dimensions in the axial plane, so the rings can be placed close to the MR bore with the subject sitting inside the rings. They generate a standing wave region with high B_1^+ . In addition, transverse dipoles are radiative antennas and much more efficient for power delivery compared to loop resonators. The PTDR coil is first explained in concept and then validated by electromagnetic (EM) simulations in different loading conditions (empty bore, cylindrical phantom and human model). Coil performance of the PTDR coil is also compared to a conventional traveling wave setup. The new coil inherits advantages of both traveling wave and standing wave coils: it exhibits excellent efficiency over short regions (inherited from standing-wave), and it maintains the ability to cover longer areas (inherited from traveling-wave).

3. Materials and Methods:

3.1 Coil structure

Fig. 1A shows the structure and orientation of the PTDR coil. It comprises two rings, each of which consists of four transverse dipoles laid out end-to-end. The diameter of each ring is set to 50 cm to leave enough free space for the human body and other hardware (receive arrays or non-proton coils). Each dipole element is a half-wavelength antenna, and is shortened to 30 cm by inductors L_t and matched to $50\ \Omega$ using capacitor C_m [12], as shown in Fig. 1B. These dipoles are driven with sequenced 90° phase increments in this study to generate a circular polarized (CP) field. To minimize port-to-port coupling between elements of the two rings, one ring is rotated 45° with respect to the other.

The number of dipole elements in each ring is not limited to four. In this work, it is set to four to achieve a good compromise between the transmit efficiency and parallel transmission/reception capability. Note that when only one or two elements are employed, full-wavelength or two-wavelength dipoles might be a better choice to cover the whole volume.

3.2 Simulation setup

Numerical full-wave simulations of the electromagnetic (EM) fields were performed to explore the PTDR coil's performance and compare it with a conventional traveling wave setup. Simulations were performed using commercially available software (ANSYS HFSS and ANSYS Designer, ANSYS Electromagnetics, Canonsburg, PA, USA). Coil elements were tuned to the 298 MHz (proton Larmor frequency of 7T) and matched to $50\ \Omega$ (the characteristic impedance of RF chains).

The values of all lumped elements (capacitors and inductors) were determined using the 3-D EM and RF circuit co-simulation method. First, lumped elements were replaced with excitation ports, and the scattering (S-) parameter matrix of the RF coil was calculated using 3-D EM simulation. The S-parameter matrix was then exported to RF circuit software for the tuning and matching. The updated values of all the lumped elements were then pushed back to the EM simulation software to calculate the EM field distributions. Since no decoupling circuit is used in the PTDR coil, the RF circuit optimization only focused on minimizing the reflection coefficient of each coil element. Details about the simulation setup can be found in Kozlov et al [13,14]. The accuracy and reliability of the B_1 fields using this method has been previously demonstrated on phantoms [15].

Simulations were performed on a workstation with 64G RAM and 12 cores (1.9 GHz Intel Xeon Processor). All conductors and lumped elements (capacitors and inductors) were set to be ideal without ohmic loss. The entire simulated system was surrounded by a perfectly absorbing boundary which was placed >20 cm away from the edge of the bore. Manual initial mesh definition was used in ANSYS HFSS to accelerate simulation convergence and ensure reliability.

Transmit RF fields (B_I^+) were extracted from the simulated fields according to:
 $B_I^+ = (B_x + iB_y)/2$ [16], where B_x and B_y are the x- and y- components of the magnetic (H) field. Local SAR and the Poynting vector were calculated using the built-in Fields Calculator module in ANSYS HFSS. The Poynting vector represents the power flow and is defined as:

$$\vec{S} = \text{Re}\left\{\vec{E} \times \vec{H}^*\right\}/2.$$

3.3 Standing-wave region produced by two traveling-wave antennae

Fig. 2A shows the simulation model of a single ring in an empty waveguide (diameter 60 cm, length 300 cm). Each dipole was excited with 1 Watt. As expected, transverse dipoles in a single ring demonstrate typical traveling wave behavior characterized by Poynting vectors that are parallel to the axis of the magnet bore, as shown in Fig. 2B. Another characteristic of traveling wave behavior is the linear B_I^+ phase distribution along the z-direction [8], as shown in Fig. 2D. B_I^+ amplitude is quite low near the ring (Fig. 2C) because the magnetic field there is mostly oriented along z.

In the presence of another ring (Ring #2 in Fig. 2E), however, a standing wave region appears between the two rings. This is evidenced by the B_I phase distribution along the z-direction in the region, which is uniform rather than linear (Fig. 2H). Ring #2 is passive and positioned 40 cm away from the active Ring #1. Compared to a single ring, the B_I^+ amplitude in the standing-wave region is increased by approximately a factor of two (Fig. 2G vs Fig. 2C). Additionally, its extent can be easily adjusted by varying the distance between two rings.

B -field magnitudes and vectors are shown in Fig. 3, where B_{tot} is the total B -field and B_z is the z-component of B -field. Note that the B -field here is generated by the RF coil and includes x-, y- and z-components, which is different from the static magnetic field (B_0) and the RF field (B_I). It is found that the passive Ring #2 acts like a reflector, improving the B_{tot} amplitude in front of it and decreasing B_{tot} behind it (Fig. 3A vs. Fig. 3D). The other factor in B_I^+ enhancement with two rings is that the reflection effect increases the transverse magnetic field efficiency in the standing wave region. For a single ring, the B -field is mainly along z-direction near the bore (Fig. 3B), a typical feature of the TE11 mode. In the presence of the passive Ring #2, however, the B_z in the standing wave region is significantly reduced (Fig. 3E). Thus, the transverse magnetic field and the B_I^+ field are relatively stronger. This can also be seen in the comparison of B vector maps in Figs. 3C and 3F, where a certain fraction of the magnetic field in the standing wave region is rotated from the z-dimension to the x- or y-dimensions.

3.4. Driving one ring versus both rings

Fig. 4 compares B_I^+ and electric (E) fields that arise when driving one ring versus both rings in differential-current mode. In the differential-current mode, the two rings are driven with the same voltage amplitudes but with a 180° phase difference. Fig. 4D shows the phase arrangement when both rings are driven. When both rings are driven, the input power of

each dipole is set to 0.5 Watts to ensure the total input power is the same as that in Fig. 2. Driving both rings increased the B_1^+ field magnitude (~ 1.5 fold) and reduced the E -field, indicating higher transmit power efficiency and higher SAR efficiency in the standing wave region. The B_1^+ improvements can be attributed to the fact that the two rings increase the reflection effect and the transverse magnetic field efficiency, as shown in Fig. 5. The E -field reduction can be attributed to the fact that the two rings produce opposite E -fields in the standing wave region when driven in differential-current mode.

3.5. Simulation setups on phantom and human model

The transmit efficiency and local SAR distribution of the PTDR coil (when both rings are driven) was then evaluated and compared to a conventional traveling wave setup, as shown in Figs. 6A and 6B. For the traveling wave setup, crossed-dipoles were placed at the end of waveguide as the transmit coil [17]. The total input power was set to 8 Watts and equally divided to each coil element in CP mode. A cylindrical phantom with 16 cm diameter was placed at the center of MR bore to load the coils. The phantom length varied from 40 to 80 cm (with steps of 10 cm) and the ring distances varied from 20 cm to 60 cm accordingly. The values of conductivity (σ) and the relative permittivity (ϵ_r) of the phantom were set similar to those of human brain tissues: $\sigma = 0.6$ S/m and $\epsilon_r = 78$.

The PTDR coil's performance was also examined in a human model simulation. Figs. 6C and 6D show the human model and a PTDR coil with ring distances of 50 cm and 20 cm, respectively. The 50-cm-distance is expected to cover the whole lower legs including the knees and calves (Fig. 6C); while the 20-cm-distance is expected to cover the knees only (Fig. 6D). The human body model (ANSYS Electromagnetics, Canonsburg, PA, USA) consists of 300 + objects (bones, muscles, and organs) and 33 tissue types, and has 1 mm³ isotropic resolution.

4. Results

4.1 Phantom Simulation

Fig. 7A and 7B show the B_1^+ and 1-g local SAR distributions using the conventional traveling wave setup and the PTDR coil (with 8 Watts input). The maps are plotted in the central sagittal slice across different phantoms with a linear color scale. B_1^+ increased as the square root of the input RF power (P_{in}), but SAR increased linearly with P_{in} . Thus, power efficiency is defined as: $B_{1_{avg}}^+ / \sqrt{P_{in}}$ and SAR efficiency (also called safety excitation efficiency) [18] is defined as: $B_{1_{avg}}^+ / \sqrt{peakSAR_{1g}}$, where $B_{1_{avg}}^+$ is the mean $|B_1^+|$ over a targeted area and $peakSAR_{1g}$ is the highest 1-gram averaged SAR. Table 1 lists the transmit and SAR efficiencies over a 5×5 cm² square in the central sagittal slice and the SAR efficiency.

The transmit efficiency of the traveling wave setup varied little with phantoms of different lengths, from $0.1 \mu T / \sqrt{W}$ to $0.13 \mu T / \sqrt{W}$. B_1^+ fields were slightly asymmetric along the bore axis, stronger at the left side of Fig. 7A due to the fact that the excitation port is located on that end. Local SAR distributions using the traveling wave setup were severely asymmetric

along the bore axis, showing a steep decrease after 6 cm behind the left end of the phantom. This is because the energy deposition of the traveling wave is focused on the excitation port's closest end.

The transmit efficiency of the PTDR coil increased when the ring distance decreased (i.e., coverage decreased). For example, the efficiency of 20-cm-coverage was 2-times higher than that of the 60-cm-coverage case ($0.32 \mu\text{T}/\sqrt{W}$ vs. $0.16 \mu\text{T}/\sqrt{W}$). B_1^+ fields in the PTDR coil were quite uniform even for the 60-cm-coverage case. Unlike the traveling wave setup, local SAR distributions of the PTDR coil were symmetrical along the bore axis due to the balanced excitation from both sides. Although both the local SAR and B_1^+ increase with decreased ring distance, the SAR efficiency increased roughly with decreased ring distance.

Compared to the conventional traveling wave setup, the PTDR coil exhibited power efficiency improvements from 1.2-fold (80-cm-long phantom) to 3.2-fold (40-cm-long phantom), and SAR efficiency improvements from 1.6-fold (80-cm-long phantom) to 2.3-fold (40-cm-long phantom). Although the local SAR of PTDR coil is higher than that of the traveling wave coil with the same input power, it produces much higher B_1^+ fields and thus has a higher SAR efficiency (defined as: $B_{1_avg}^+ / \sqrt{\text{peakSAR}_{1g}}$).

4.2. Human Leg Simulation

Fig. 8A-C shows the B_1^+ maps in one human leg with ring distances of 50 and 20 cm. Fig. 8D shows the 1g-SAR distribution in the central coronal plane of the human model for an average B_1^+ in the knees of $1 \mu\text{T}$

For the ring distance of 50 cm, B_1^+ was relatively uniform over the whole lower legs (including the knees and calves). The average B_1^+ over the whole lower leg was $0.37 \mu\text{T}$. When the ring distance was shortened to 20 cm (aiming to cover the knees only), the average B_1^+ over the knees was $0.68 \mu\text{T}$, indicating that 30% of the RF power is needed to achieve the same B_1^+ strength in the knees. Note that a portion of the RF power is still deposited beyond the targeted area (such as thighs and hips) although most of it is concentrated in the rings' gap area. The SAR distributions depend on the tissue geometry and EM properties, and are thus different from those in phantoms. The highest SAR in both cases is located in one leg when driven in CP mode. For the same B_1^+ in the knee area ($1 \mu\text{T}$), the highest local SAR is 2.8 W/Kg when the ring's gap is 20 cm, while the highest local SAR is 5.2 W/Kg when the ring's gap is 50 cm. This indicates 37% more B_1^+ in the knee area can be obtained with the same SAR limits by shortening the gap to cover only the anatomy of interest.

One of main challenges in high-field MRI is inhomogeneous B_1 fields due to RF wave behavior, especially for transverse slices. The most widely-accepted approach to solve this problem is parallel transmission, which enables tailoring the amplitudes and phases of RF waveforms for individual channels [19–23]. Fig. 9 shows central axial B_1 fields using individual elements of the PTDR coil and their combination in CP mode (same amplitude and phase arrangements as those in Figs. 4 and 5). Their distinct patterns suggest that they are effective for the parallel transmission at 7T. The PTDR coil could also provide additional

spatial B_1 diversity that is complementary to longitudinal dipoles [24] due to their intrinsic geometric decoupling.

5. Discussion:

In this study, we proposed and numerically analyzed a new coil (PTDR) which combines the concepts of traveling-wave and standing-wave, and has adjustable coverage along the z-direction. Consequently, it exhibits excellent transmit efficiency and SAR efficiency over short regions and maintains the ability to cover long z-FOVs. Phantom simulations showed that PTDR coils achieved up to 60 cm coverage, which is a reasonable upper limit since gradient linearity along the z-dimension is typically 50 cm. In the current design, the ring distance has to be at least 10 cm to avoid the strong coupling between dipoles of different rings. When the coverage of the PTDR coil was shortened to 20 cm, its transmit efficiency and SAR efficiency can be improved by 100% and 44% compared to the 60-cm-coverage case, respectively. Similar simulation results were also found when the PTDR coil was used for the knee only (short coverage) and the whole lower leg (long coverage). We limited our comparisons to a traveling-wave coil because we know of no common design that covers a volume as long as 60 cm. In this work, the PTDR coil was driven in “birdcage” mode and the axial B_1^+ field maps had a bright center (Fig. 9), similar to that of a conventional birdcage coil. However, each element of the PTDR coil has a distinct B_1^+ pattern, so uniform flip angles may be achieved using parallel transmission in an RF shimming mode or with tailored spokes pulses [21, 22].

By rotating the ring, the isolation between dipoles in the two rings was acceptable across different distances, with S_{21} better than -14 dB ($<4\%$ power cross-talk). Without rotating the ring, adjacent dipoles in the two rings exhibit strong coupling and thus may decrease the coil performance. The worst port-to-port isolation occurred for adjacent dipoles in the same ring, with S_{21} of about -7 dB ($\sim 20\%$ power cross-talk). In either a multiple-channel transmit system or a single-channel transmit system with power splitting networks, the cross-talk transmit power would be absorbed by protection circuits and wasted, consequently decreasing transmit efficiency. Therefore, if the port isolation can be improved by appropriate decoupling treatments, the transmit efficiency should increase.

A key point of the PTDR method is the use of transverse dipoles. They have a strong reflection effect that is desirable to reinforce the RF fields in the imaging region and increase the transverse magnetic field efficiency. Another advantage of transverse dipoles is that they can be placed close to the bore to leave enough free space for the subject and other coils. It should be noted that the reflection effect of transverse dipoles is not perfect and some RF power is still deposited outside the imaging region (Fig. 5), where it is wasted. One way to reduce this power waste could be the use of appropriate shielding or Yagi-Uda reflectors [25] placed behind the dipole rings. An alternative method would be to fold or bend the dipoles toward each other to restrict more power in the targeted imaging region. Using these methods, the power flow to unwanted areas could be reduced and the transmit efficiency could be further improved.

In conclusion, we proposed a novel PTDR RF coil for ultrahigh field MRI, and evaluated it in simulations. The results showed that the advantages of both standing-wave and traveling-wave excitation can be obtained using PTDR coils; specifically, PTDR coils exhibit high efficiency over short regions while maintaining the ability to cover longer areas. Further studies will be focused on constructing this PTDR coil and validating its performance through MR experiments.

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7. References

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Fig. 1.

A: Simulation model of the proposed **P**air of **T**ransverse **D**ipole **R**ings (PTDR) coil. It is made up of two rings, each of which comprises four identical transverse dipoles (30 cm long and 1 cm wide) which are placed head-to-head and driven with 90° phase increments to generate a combined CP traveling wave field. **B:** Diagram of a dipole element.

Fig. 2.

A: Simulation model of a single dipole ring. **B-D:** Simulated Poynting vector and B_1^+ distributions result when the dipole elements in Fig. 2A are excited with 1 Watt each. **E:** Simulation model of two dipole rings. **F-H:** Simulated Poynting vector and B_1^+ distributions result when the dipole elements of Ring 1 in Fig. 2E were excited with 1 Watt each, but the dipoles in Ring 2 were terminated with 50 Ω . The arrow colors in Figs. 2B and 2F represent the magnitude, including x-, y- and z- components. The sudden transition from blue to orange in Figs. 2D and 2H is due to phase wrapping.

Fig. 3.

A-C: Simulated B_{tot} magnitude, B_z magnitude and B -field vector result when the dipole elements are excited with 1 Watt each (without another Ring). **D-F:** Simulated B_{tot} magnitude, B_z magnitude and B -field vector result when the dipole elements of Ring #1 were excited with 1 Watt each, but the dipoles in Ring #2 were terminated with 50 Ω .

Fig. 4.

A: Phase arrangement when one ring is driven with 1 Watt in each dipole. **B and C:** Simulated B_1^+ and E -field distribution using the setup in Fig. 4A. **D:** Phase arrangement when both rings (with the same diameter) are driven, where each dipole is driven with 0.5 Watts. These dipoles are driven with sequenced 90° phase increments in this study to generate a circular polarized (CP) field. To minimize port-to-port coupling between elements of the two rings, one ring is rotated 45° with respect to the other. **E and F:** Simulated B_1^+ and E -field distribution using the setup in Fig. 3D. These simulations were performed with an empty bore without a phantom or human model in it.

Fig. 5.
A-C: Simulated B_{tot} magnitude, B_z magnitude and B -field vector result when one ring is driven. **D-F:** Simulated B_{tot} magnitude, B_z magnitude and B -field vector result when both rings (with the same diameter) are driven.

Fig. 6.

A: Simulation model of traveling wave setup on phantom. **B:** Simulation model of the PTDR coil on phantom. **C:** Simulation model of the PTDR coil covering the whole lower legs (ring distance=50 cm). **D:** Simulation model of the PTDR coil covering the knees only (ring distance=20 cm).

Fig. 7.

Comparison of the proposed PTDR coil with a traveling-wave alone setup. **A:** B_1^+ distributions in the central sagittal slice of the phantom. **B:** Local SAR distributions in the central sagittal slice of the phantom with the same input power. In both cases, the total input power was set to 8 Watts and equally divided across coil elements. A cylindrical phantom with 16 cm diameter was placed at the center of MR bore to load the coils. The phantom length varied from 40 to 80 cm (with steps of 10 cm) and the ring distances were varied from 20 cm to 60 cm accordingly.

Fig. 8.

Comparison of PTDR coils with 50-cm and 20-cm gaps covering the whole lower legs (50 cm) or the knees only (20 cm). **A-C:** B_1^+ maps in the leg (sagittal, coronal and axial slices). **D:** 1g-SAR distribution in the central coronal plane of the human model when the average B_1^+ in the knees equals 1 μ T. The peakSAR_{10g} is much lower with the ring's gap of 20 cm, partly because less RF power is needed due to the greater B_1^+ efficiency, partly because less RF power is deposited in non-imaged regions.

Fig. 9.

Individual B_1 fields of the PTDR coil in the phantom, midway between the two rings. The distinct patterns suggest that they may be effective for parallel transmission and parallel imaging at 7T. The phantom is 40-cm long and the ring distance is 20 cm. Each dipole is driven with 1W input.

Table 1

Transmit efficiency over a $5 \times 5 \text{ cm}^2$ square in the central sagittal slice and SAR efficiency using the traveling wave (TW) setup and the PTDR coil.

Transmit efficiency and SAR efficiency of traveling wave (TW) setup and PTDR coil			Phantom length/Ring distance [cm]				
			40/20	50/30	60/40	70/50	80/60
TW	$B_{1_avg}^+ / \sqrt{P_{in}}$	$[\mu\text{T}/(\text{W})^{1/2}]$	0.10	0.11	0.12	0.13	0.13
	$B_1^+ / \sqrt{\text{peakSAR}_{1g}}$	$[\mu\text{T}/(\text{W/Kg})^{1/2}]$	0.59	0.66	0.68	0.65	0.61
PTDR	$B_{1_avg}^+ / \sqrt{P_{in}}$	$[\mu\text{T}/(\text{W})^{1/2}]$	0.32	0.29	0.24	0.20	0.16
	$B_1^+ / \sqrt{\text{peakSAR}_{1g}}$	$[\mu\text{T}/(\text{W/Kg})^{1/2}]$	1.37	1.17	1.24	1.22	0.95