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Comparison of radiofrequency body coils for MRI at 3 Tesla: a simulation study using parallel transmission on various anatomical targets

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Abstract

The performance of multichannel transmit coil layouts and parallel transmission (pTx) radiofrequency (RF) pulse design was evaluated with respect to transmit B1 (B1+) homogeneity and Specific Absorption Rate (SAR) at 3 Tesla for a whole body coil. Five specific coils were modeled and compared: a 32-rung birdcage body coil (driven either in a fixed quadrature mode or a two-channel transmit mode), two single-ring stripline arrays (with either 8 or 16 elements), and two multi-ring stripline arrays (with 2 or 3 identical rings, stacked in the z-axis and each comprising eight azimuthally distributed elements). Three anatomical targets were considered, each defined by a 3D volume representative of a meaningful region of interest (ROI) in routine clinical applications. For a given anatomical target, global or local SAR controlled pTx pulses were designed to homogenize RF excitation within the ROI. At the B1+ homogeneity achieved by the quadrature driven birdcage design, pTx pulses with multichannel transmit coils achieved up to ~8 fold reduction in local and global SAR. When used for imaging head and cervical spine or imaging thoracic spine, the double-ring array outperformed all coils including the single-ring arrays. While the advantage of the double-ring array became much less pronounced for pelvic imaging with a substantially larger ROI, the pTx approach still provided significant gains over the quadrature birdcage coil. For all design scenarios, using the 3-ring array did not necessarily improve the RF performance. Our results suggest that pTx pulses with multichannel transmit coils can reduce local and global SAR substantially for body coils while attaining improved B1+ homogeneity, particularly for a “z-stacked” double-ring design with coil elements arranged on two transaxial rings.

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Keywords

RF array design; body MR imaging; parallel transmission; high field MRI

Introduction

During the last decade, the fast growing interest to use MR scanners operating at high (3 Tesla (T) – 7T) and ultra-high (7T and higher) magnetic fields has triggered a large body of work aimed at using multichannel or parallel RF transmission (pTx) (1,2) to tackle the issue of transmit B1 (B1+) inhomogeneity arising from complex interactions of shortened RF wavelengths with dielectric and lossy biological tissues (3). The pTx technology relies on a transmit RF array built with multiple coil elements that can be driven by independent transmit channels (4). Initial pTx works, including B1+ shimming (5–8), transmit SENSE (1,2,9–11), multi-spoke (12–14) or k_T point (15) pulse designs, have mostly been performed using a head or body RF array with coil elements only distributed in a single-ring structure within the transaxial plane. Meanwhile, there is constant interest in seeking optimal RF array design for pTx. Efforts have been taken to characterize the impact of the number of transmit channels (16,17), or to study the ultimate global Specific Absorption Rate (SAR) in multichannel transmission (18,19).

It has been also shown that either experimentally or by simulations that distributing transmit coil elements in multiple transaxial rings stacked along the z axis (i.e., the direction of the main magnetic field) could further enhance pTx RF pulse performances by improving excitation homogeneity, or reducing SAR, or both (20–26). In a recent study targeting body imaging at 3T, Guerin et al. (25) compared the performance of several body RF arrays built with loop-shaped coil elements, including single-, two-, three- and four-ring arrays where rings of multiple coil elements were stacked along the z axis. Their results using single-slice pTx pulse design targeting liver imaging showed that distributing RF coil elements along the z axis provided substantial advantages in terms of excitation homogenization and peak local SAR reduction, although at the cost of increased average or peak RF power (25). However, these results cannot be used to infer what would happen if a different type of transmit elements (e.g., stripline elements) were used, a different type of pTx pulses (e.g., slab-selective pulses) were desired, or a different anatomical target (e.g., brain) was excited.

In this study, we further explored at 3T the impact of the geometric design of multi-channel body RF coils on excitation homogeneity and peak local SAR, focusing on several aspects that, to our knowledge, have not been previously investigated. Specifically, for a given RF coil design, we considered three anatomical targets: 1) head and cervical spine (head/C-spine), 2) thoracic spine (T-spine), and 3) pelvis. Each target was defined by a 3D volume representative of a meaningful region of interest (ROI) in routine clinical applications at 3T. Furthermore, we simulated RF arrays made of stripline elements (4,27) (instead of loops) and designed pTx RF pulses to homogenize excitation within the entire ROI (as opposed to a single slice). The RF arrays modeled in this study include two single-ring arrays consisting of 8 or 16 coil elements azimuthally distributed within the transaxial plane, and two multi-ring arrays comprising 2 or 3 rings stacked along the z axis with 8 coil elements per ring. In

order to include in the comparison a conventional body RF coil at 3T, we also modeled a vendor provided standard two-port birdcage body coil and drove it in a fixed quadrature mode (with same RF amplitude but a 90° RF phase difference between the two ports) or a two-channel transmit mode with independent RF phase and amplitude inputs to the two ports.

Methods

Electromagnetic modeling

In this study, five 3T body RF coils were modeled and compared: 1) a two-port fed, 32-rung shielded high-pass birdcage (BC) coil; 2) a single-ring array consisting of 8 stripline elements, referred to as “SL 1×8”; 3) a single-ring array composed of 16 stripline elements, referred to as “SL 1×16”; 4) a two-ring array consisting of two rings of 8 stripline elements each, referred to as “SL 2×8”; and 5) a three-ring array comprising three rings of 8 stripline elements each, referred to as “SL 3×8”. Within each of the 8- or 16-elements rings, the stripline elements were evenly and azimuthally distributed within the transaxial plane. For multi-ring arrays, the rings were stacked along the z axis without overlapping.

The BC coil (28,29) considered was an inductively coupled coil, of which the 32 rungs were aligned in the z direction and distributed azimuthally on a cylindrical surface of 704 mm in diameter. All rungs measured 500 mm long and 20 mm wide, and were enclosed by a coaxed RF shield consisting of a cylindrical copper sheet measuring 744 mm in diameter and 1200 mm in length. The coil came with two RF feeding ports, azimuthally separated in space by 90° (see Fig. 1). Two transmit modes were considered: a traditional quadrature mode (referred to as “BC 90°”) and a two-channel transmit mode (referred to as “BC 1×2”). In the quadrature mode, the two ports were driven with same RF amplitude but a fixed 90° phase shift so as to create a circularly polarized (CP) RF field in vacuum and at the coil center, whereas in the two-channel transmit mode, the two ports were driven by two independent RF channels allowing port-specific adjustment of RF waveforms. Note that in either mode, driving each port will energize all 32 rungs.

Both “SL 1×8” and “SL 1×16” arrays had a conventional single-ring structure with all the coil elements at the same level along the Z direction and evenly positioned on a cylindrical surface. The azimuthal angle spanned by any two neighboring elements was 45° and 22.5° for the “SL 1×8” and “SL 1×16” arrays, respectively.

The “SL 2×8” and “SL 3×8” arrays, however, were designed to have a multi-ring (or z-stacked) structure where the coil elements were evenly split into two or three rings along the z-axis. For both arrays, each ring consisted of 8 elements which were evenly and azimuthally distributed with a constant azimuthal angle of 45° between adjacent elements. Furthermore, in order to reduce coil coupling between the rings, a 10-mm gap was introduced between any two adjacent rings along the z direction, and an azimuthal 22.5° rotation was applied to one ring relative to any adjacent ring.

All four stripline arrays had the same diameter as the BC coil, i.e., the cylindrical surface on which the stripline elements were placed was 704 mm in diameter. All stripline elements

were made of copper stripes measuring 20 mm in width. The length of the elements, however, was different depending on the coil type – the length of the elements was 450 mm for both “SL 1×8” and “SL 1×16” arrays, but was 220 mm for the “SL 2×8” array and 145 mm for the “SL 3×8” array. The RF shield structure described in the BC coil was also used to enclose the stripline elements and was connected to the ground line of the coil. The dielectric material used in between the coil elements and the ground was air. Outlines of the five coil structures are shown in Fig. 1 (the same BC coil was used for both “BC 90°” and “BC 1×2”).

Electromagnetic (EM) simulations were run in a commercial software (Remcom, USA) using the Finite Difference Time Domain (FDTD) algorithm to solve Maxwell’s equations. All coils were loaded with a human whole body model (Duke, Virtual Family (30)). This tissue model, consisting of 545,674 tissue voxels and 76 tissue types, was defined on a grid with 5 mm isotropic resolution. Each of the three anatomical targets (i.e., head/C-spine, T-spine and pelvis) was defined by a cubical ROI covering an anatomical region typically targeted in routine clinical MRI sessions. The size of the ROI along x (mediolateral direction), y (anteroposterior direction) and z (superoinferior direction) axes was $20 \times 24 \times 41 \text{ cm}^3$ for head/C-spine, $4 \times 22 \times 40 \text{ cm}^3$ for T-spine and $32 \times 20 \times 40 \text{ cm}^3$ for pelvis. For each anatomical target, the ROI was centered at the isocenter of the RF coil along the z-axis; all voxels falling outside the ROI were ignored for RF pulse design (see Fig. 2).

For each stripline array and for each given anatomical target, the EM field maps of individual coil elements were simulated by driving one element at a time in the presence of all other elements. Using EM simulations, each element was tuned to the proton Larmor frequency at 3T (i.e., 128 MHz) through iterative adjustments of the value of a lumped tuning capacitor. As in previous studies (31–33), the matching circuitry was not included in the modeling for simplicity. Decoupling between coil elements was simply the result of inter-element distance and was found to always be at least –20 dB within each of the four stripline arrays. To reduce the scaling effect of element-dependent quality factors, the simulated EM fields of each element were normalized to the corresponding net input RF power.

For the BC coil operated in the two-channel transmit mode (i.e., “BC 1×2”), the EM field of each port was simulated by driving one port at a time. The EM field of the BC coil, when operated in the standard quadrature CP mode (i.e., “BC 90°”), was obtained by driving the two ports simultaneously with the same RF magnitude, but with a 90° relative phase shift. The flip angle distribution and SAR derived for this CP mode were used as a reference baseline against which the RF performance of the other coil structures investigated in the present study was compared.

For each combination of coil design and anatomical target, except when using “BC 90°”, the normalized multi-channel electric field maps were utilized along with the tissue electric properties (i.e., conductivity and mass density) to calculate the global SAR matrix (2) and the local 10-gram SAR (10g SAR) matrices (34). A voxel-specific 10g SAR matrix was derived for each tissue voxel based on a fast region growth method (35) in combination with an advanced algorithm (36) to ensure that the cubic neighboring tissue volume over which

the SAR is averaged corresponds to exactly 10 grams of tissue. This resulted in a total number of 545,674 10g SAR matrices, which were subsequently compressed into a largely reduced number of virtual observation points (VOPs) (37) for use in the local SAR controlled pulse design as described later. Table 1 presents the parameters utilized to create the VOPs for each pTx pulse design scenario and the resulting number of VOPs.

pTx RF pulse design

In the present study, slab-selective pTx spoke RF pulses (38,39) were designed using single- or two-spoke pulses to achieve a uniform flip angle (FA) distribution within the ROI. In the case of a single-spoke pulse design which is also referred to as RF shimming (4,5,7,8,40–43), a traditional slab-selective RF excitation module consisting of a slab-selective shaped RF pulse (e.g., a sinc pulse) and a slab-selective gradient was employed, while the same RF pulse shape was multiplied by a channel-specific RF shim value (a complex number representing RF phase and magnitude). In the two-spoke pulse design, two such slab-selective RF excitation modules were utilized, and, more importantly, inter-pulse gradient blips were calculated to define the placement of the spoke trajectories in the excitation k-space (44).

For each pTx enabled scenario, the channel-specific and spoke-specific RF shim values were calculated by formulating in the image domain (10) a constrained minimization to minimize either global or local SAR while satisfying predefined excitation fidelity:

$$\hat{w} = \underset{w}{\operatorname{argmin}} \operatorname{SAR}(w), \text{ s.t. } \operatorname{RMSE}(w) \leq \varepsilon \quad [1]$$

where w is a complex-valued vector concatenating the RF shim values, $\operatorname{SAR}(w)$ the cost function formulated depending on which SAR type to be minimized, $\operatorname{RMSE}(w)$ a scalar denoting the Root Mean Square excitation Errors (RMSE), and ε a number prescribing excitation error. The RMSE was defined in a Magnitude Least Squares (MLS) sense (45)

assuming small-tip-angle excitation (46): $\operatorname{RMSE}(w) = (\|Aw - d\|_2^2 / N_u)^{1/2}$, where A is a complex-valued system matrix involving the B1+ maps of individual RF channels inside the ROI, d a vector representing the magnitude of the desired transverse magnetization, and N_u the number of voxels within the ROI.

For both RF shimming and two-spoke design, the system matrix A was constructed as $A = [S_1 A_0, S_2 A_0, \dots, S_C A_0]$ where C is the number of channels, $S_c = \operatorname{diag}\{S_c(r_u)\}$ is a diagonal matrix with its diagonal elements being the B1+ map s_c of the c -th channel at spatial location r_u (with $u=1, 2, \dots, N_u$), and A_0 is a matrix involving Fourier kernel associated with the k-space placement of the spokes. The u -th spatial and v -th spoke element of matrix A_0 was further expressed as $a_{uv} = i\gamma f \cdot \exp(ir_u \cdot k_v)$ assuming no correction for main field inhomogeneity, where $i = \sqrt{-1}$ is the imaginary unit, γ the gyromagnetic ratio, f the time integration of the sub-pulse shape, and k_v the k-space position (with $v=1, 2, \dots, N_v$ and N_v being the number of spokes). The size of the overall system matrix A was therefore $N_u \times C$ for RF shimming (where $N_v=1$) and $N_u \times (2C)$ for two-spoke pulse design (where $N_v=2$).

Correspondingly, the vector w was given by $w = [w_{1,1}, w_{2,1}, \dots, w_{N_v-1,C}, w_{N_v,C}]^T$ where $w_{v,c}$ is the RF shim value for spoke v and channel c , and the vector d was written as $d = [d_1, d_2, \dots, d_{N_u}]^T$ where d_u denotes the magnitude of the desired transverse magnetization at the u -th spatial location.

The cost function $\text{SAR}(w)$ was formulated to control either global SAR or peak 10g SAR.

For global SAR control, $\text{SAR}(w) = \|S_0 w\|_2^2$ was written in a quadratic form explicitly in w , where S_0 is a matrix created based on the global SAR matrix (2). For peak 10g SAR control,

$\text{SAR}(w) = \sum_{n=1}^{N_{\text{VOP}}} \alpha_n \|S_n w\|_2^2$ was formulated as a weighted sum of all local SAR values estimated by VOPs, where N_{VOP} is the number of VOPs, S_n is a matrix constructed based on the n -th VOP, and α_n is a weighting factor used to weight the local SAR estimated by the n -th VOP. We constructed S_0 and S_n matrices following the recipe described in the Appendix of Sbrizzi et al. (47).

Following Lee et al. (48), we sought to solve the constrained minimization in Eq. [1] by casting it into a regularized MLS problem,

$$\hat{w} = \underset{w}{\operatorname{argmin}} [\text{RMSE}^2(w) + \lambda \cdot \text{SAR}(w)], \quad [2]$$

where λ is the regularization parameter. The way Eq. [2] was solved was dependent on which SAR type was controlled. Specifically, for global SAR controlled pulse design, a number of RF shim solutions were first calculated by varying the regularization parameter λ in Eq. [2], and the optimum solution was then chosen as the one that yielded the minimum global SAR among those satisfying $\text{RMSE}(w) \leq \varepsilon$. For each given λ value, Eq. [2] was solved using the variable exchange algorithm (45).

However, for local SAR controlled pulse design where $\text{SAR}(w) = \sum_{n=1}^{N_{\text{VOP}}} \alpha_n \|S_n w\|_2^2$, the optimum RF shim solution was obtained through an iterative approach as described in Lee et al. (48) and Sbrizzi et al. (47). Briefly, we started with equal weightings to all VOPs and iteratively updated the weighting factors to reduce the peak local SAR estimated by VOPs. In each iteration, Eq. [2] for a given set of weighting factors was solved in a way similar to that described above for the global SAR controlled design, and an RF shim solution was obtained that gave rise to the least peak 10g SAR while satisfying $\text{RMSE}(w) \leq \varepsilon$. The weighting factor of this solution's peak local SAR was then increased to impose more constraint and to reduce the peak local SAR in the next iteration. The iterative process was terminated and the optimum RF shim solution returned when the peak local SAR estimated by VOPs did not decrease or the predefined maximum number of iterations were reached.

Furthermore, in the two-spoke pulse design with global SAR control, the spoke placement was optimized as described below in order to reduce the impact of the spoke placement on final RF performance, thereby leaving the coil design to be a dominant factor. Specifically, the spoke placement optimization was conducted for each λ value through an exhaustive search. The optimum placement was then determined as the one among 3312 candidates that gave rise to the minimum cost evaluated by $\text{RMSE}^2(w) + \lambda \cdot \text{SAR}(w)$. The placement candidates were obtained by gridding the k -space based on spherical coordinates (i.e., k_r, k_θ ,

k_ϕ) and restricting the two spokes to be symmetric with respect to the k-space origin. The gridding of the k-space was defined by incrementing the radius, k_r , from 0 to 10 m^{-1} in a nonlinear fashion — the corresponding field of excitation was incremented by 0.1 m between 0.1 and 1 m, by 0.2 m between 1.2 and 2 m, and by 0.4 m between 2.4 and 5 m; infinite field of excitation corresponding to $k_r = 0$ was also considered so that both spokes could be placed at the k-space origin. Both polar (k_θ) and azimuthal (k_ϕ) angles were incremented from 0° to 180° in steps of 15° . The optimum placement per lambda value was found once the pulse design problem had been solved for all placement candidates. The same λ -specific optimum spoke locations were also utilized to design two-spoke local SAR controlled pulses.

All pulse designs targeted uniform excitation within the ROI with nominal flip angles of 10 degrees. The sub-pulse shape was a 1-ms Hanning filtered sinc pulse with bandwidth time product = 10 and defined with a 10- μs dwell time, yielding a pulse duration of 1 ms for one-spoke and 2 ms for two-spoke pulses without including the ramps of the slab-selective gradient.

RF performance evaluation

L-curves quantifying the tradeoff between SAR values (either peak 10g SAR or global SAR) and RMSE were generated for each design scenario by prescribing several excitation errors ε in Eq. [1]. The global SAR was calculated using the corresponding global SAR matrix. Note that instead of using the VOPs (which were only used for designing local SAR controlled pulses), we derived the local SAR using all of the 545,674 10g SAR matrices to avoid SAR overestimation. All SAR numbers were calculated by averaging the specific energy absorption of a single RF excitation over a 2-ms time window.

In addition, more quantitative analyses were performed to further compare the coil performance using local SAR regularized pulse design. Two comparison criteria were considered: 1) resulting peak local SAR values while achieving similar excitation fidelity and 2) resulting excitation fidelity at a common peak local SAR. For comparison using the first criterion, the single RMSE value achievable with the “BC 90°” (i.e., the fixed quadrature mode of the BC coil) was utilized as the RMSE reference, and RF solutions yielding a comparable RMSE were chosen for all the pTx enabled pulse design scenarios (i.e., one- and two-spoke pulse designs using “BC 1×2” and the other four stripline arrays). For comparison with the second criterion, the peak 10g SAR of the RF solution corresponding to the best excitation fidelity (i.e., the least RMSE) achievable when designing two-spoke pulses using “SL 2×8” was chosen as the 10g SAR reference, and the RF solutions leading to a comparable 10g SAR were chosen for all the pTx enabled pulse design scenarios.

In these comparisons, the 10g SAR distribution and the FA map within the whole body for each design scenario were presented. The 3D FA maps were obtained by Bloch simulations of the designed pTx pulses. In addition, the Coefficient of Variation (CV), defined as the standard deviation divided by the mean of the FA distribution (i.e. $\text{CV} = \text{std}(\text{FA})/\text{mean}(\text{FA})$) was calculated to quantify the FA inhomogeneity within the ROI. Note that all Bloch simulations assumed no off resonance effect (i.e. $B_0=0$) and were obtained in the absence

of slab-selective gradient to mimic a non-selective excitation, so as to eliminate the potential impact of slab profile imperfections on the calculated CV, a bias that could otherwise affect result interpretation.

Results

For a given λ value in the two-spoke pulse design with global SAR control, the resulting RF performance was largely dependent on the k-space placement of the two spokes utilized in the pulse calculation. For example, when targeting T-spine (Fig. 3), the RF solutions resulting from the 3312 k-space placement candidates spread out in the SAR vs. RMSE space and presented a cloud pattern with large extremity to extremity distance especially when the resulting RMSE values were relatively small.

Comparing the L-curves of the controlled SAR quantity vs. the excitation error, the “SL 2×8” stripline array with 16 elements distributed in 2 rings gave rise to more favorable excitation fidelity vs. SAR tradeoffs than either of the two single-ring stripline arrays (i.e., “SL 1×8” or “SL 1×16”) when designing pulses for head/C-spine (Figs. 4a and 5a) or T-spine imaging (Figs. 4b and 5b). This was true for both one-spoke (upper row, Fig. 4 and 5) and two-spoke design (lower row, Fig. 4 and 5). However, for pelvic imaging with a substantially larger ROI, the advantage of the double-ring stripline array “SL 2×8” became much less pronounced (Figs. 4c and 5c).

Figs. 4 and 5 illustrate the L-curves for RMSE vs. the SAR quantity that was specifically controlled in pulse design as well as the impact of this specific design on the behavior of RMSE vs. the *uncontrolled* SAR parameter, for example, the behavior of RMSE vs. peak 10g SAR when global SAR was specifically controlled (Fig. 4) or *vice versa* (Fig. 5). Controlling one SAR parameter significantly improved the other as well.

For all design scenarios, further increasing the number of coil elements to 24 while distributing the elements in 3 rings (i.e. using the stripline array “SL 3×8”) did not increase the RF performance in terms of FA homogenization and SAR reduction, as compared to the “SL 2×8” array (Figs. 4 and 5). In most cases, it even degraded the RF performance (e.g., Fig. 5a).

Using a multielement stripline array always led to better achievable excitation fidelity (i.e., smaller least RMSE) than the BC coil (Figs. 4 and 5). Comparing the two transmit modes of the BC coil, the use of the two-channel transmit mode, “BC 1×2”, resulted in better RF performance than the fixed quadrature mode, “BC 90°”; this was observed for all three anatomical targets and for both global and peak local SAR regularization, especially when designing two-spoke pulses (bottom row in Figs. 4 and 5).

Figs. 6 and 7 illustrate quantitative comparisons for one- and two-spoke pulse designs, respectively, at an RMSE of ~0.04 for local SAR regularized pulses designed for T-spine imaging. This RMSE was the level of excitation fidelity achievable by standard mode of the BC coil (“BC 90°”). For one-spoke design, the two-channel transmit mode of the body coil, i.e. “BC 1×2”, reduced the local SAR to 28% of the value attained by the “BC 90°” mode, and, in this case, the double-ring stripline array “SL 2×8” performed only marginally better

(Figs. 6). For two-spoke pulse design (Fig. 7), peak 10g SAR was lower substantially for all coils relative to the one-spoke design (Fig. 6), but significantly more so for the double-ring stripline array “SL 2×8” which lowered peak 10g SAR to 22% of the level attained by the two-spoke pulse for the “BC 1×2” coil. The two ring “SL 2×8” outperformed all other coils, including the single-ring 16-channel design, achieving peak 10g SAR levels of ~75% of the “SL 1×16” (Figs. 6 and 7). Looking at extremes, both local and global SAR values were highest for the standard “BC 90°” coil and, with local SAR control, both local SAR and global SAR were reduced approximately 8-fold for the “SL 2×8” array and two-spoke pulse design.

In addition, when resulting in a similar peak 10g SAR, the use of the double-ring array “SL 2×8” decreased the RMSE (thus increased excitation fidelity) by up to 40% for one-spoke pulse design and 67% for two-spoke pulse design, as compared to the two single-ring stripline arrays (Figs. 8 and 9). With this comparison criterion, “SL 2×8” also reduced the RMSE by up to 40% and 67% as compared to the “SL 3×8” array, and 18% and 78% to the “BC 1×2” case, respectively. This reduction in RMSE in turn translated into an improved FA homogeneity as well as a closer-to-nominal FA inside the ROI for the “SL 2×8” array, as evidenced by the CV and mean FA calculations (bottom row in Figs. 8 and 9). Furthermore, using the “SL 2×8” array to design two-spoke pulses led to simultaneous improvement in excitation fidelity and reduction in SAR as compared to the standard mode of the BC coil (i.e., “BC 90°”).

Discussion

We have evaluated the performance of several body transmit RF coils at 3T, including a conventional birdcage structure driven in either a fixed quadrature mode or a dual-channel mode, as well as several arrays of stripline elements organized in either one ring of 16 elements or 2 or 3 rings (along the z-axis) of 8 elements each. The coil comparison focused on excitation fidelity under global or local SAR control, considering three standard imaging targets (head/C-spine, T-spine and pelvis) with either single- or two-spoke RF pulses. Our results indicate that, overall, increasing the number of transmit channels allows for a more uniform excitation with the lowest possible RMSE values (leftmost end of L-curves) always obtained with either 16- or 24-element coils, although the results are less uniform when one considers the best tradeoff (i.e. around the corner of L-curves) between excitation fidelity and either local peak SAR or global SAR values. Noticeably, in the head/C-spine and T-spine targets, the “SL 2×8” array consistently provided substantially better tradeoffs than all other coils, and this was observed for both local peak SAR and global SAR control cases, and with one- and two-spoke RF pulses. In the same targets, the “SL 1×8” and “SL 1×16” coils behave in a similar way, whereas the “SL 3×8” results were found between the latter and the “SL 2×8”. Importantly, the fact that the gains obtained with the “SL 2×8” coil were not matched by “SL 1×16” despite a same number of coil elements (i.e. 16), strongly suggests that the distribution along the z-axis of the two rings of 8 elements plays a crucial role in attaining the superior performance of the “SL 2×8” coil. Similar results were observed in the human brain at 7T when comparing different RF head array designs in the context of pTx multiband pulse design (49,50) – only when distributing 16 coil elements in

two rings was the RF performance improved as compared to a single-ring 8-element array design (26).

The results obtained in the pelvis significantly differed from those in the other targets, with a remarkable similarity between the “SL 1×16”, “SL 2×8” and “SL 3×8” arrays. Interestingly, the “BC 1×2” coil, within the range of RMSE that it could achieve in the pelvis, resulted in smaller SAR values than the stripline arrays. Further investigation would help to determine if these observations are mostly the result of the larger size of the pelvis target.

Our results for the birdcage coil suggest that the two-channel transmit mode can be used to effectively improve excitation homogeneity and reduce SAR, which we mainly attributed to the increased degrees of freedom in RF control. These results are consistent with previous studies that have experimentally demonstrated that two-channel transmission is advantageous over conventional single-channel transmission at 3T (51,52).

The main purpose of the current study was to find an optimal coil layout for an RF body array at 3T designed with stripline coil elements. Including a matching circuitry and simultaneously tuning and matching the many coil elements of each array using EM simulation would have yielded an extremely large computational burden, leading to prohibitively long EM simulations considering the local computational resources that were available at the time of our study. After performing some comparisons with and without matching, it was found that the latter had only very limited impact on the pattern of EM fields inside the body. For example, for an element picked from the “SL 1×16” array when targeting the T-spine, the correlation coefficients between EM fields simulated with and without matching were 0.96/0.92 (real part/imaginary part) for the B1+ fields, and 0.90/0.95 for the electric fields. Based on this high similarity between EM field patterns, which logically derives from current patterns being overall retained along the resonant element, it was decided not to include matching circuitry as was also done in other studies (31–33). In future work, co-simulation approaches (53,54) could be considered to incorporate coil matching in the simulations.

Regarding coupling between coil elements, we found that the natural decoupling achieved by the relatively large inter-element distance was at least –20 dB or more within each array considered in this study, which, based on our experience with pTx pulse design using stripline arrays at 7T is satisfactory for pTX applications (4,5,49). As a result, we did not add any decoupling network in our stripline RF arrays. However, if Bloch simulations were to be run taking into account mutual coupling between array elements, the mutual coupling matrix could be incorporated into the pulse design as described by Pang and Zhang (55) to improve further excitation accuracy.

Intuitively, one could have expected the best RF performance to be obtained with the “SL 3×8” array, given its larger number of channels potentially allowing for more degrees of freedom in pulse design. However, our data suggest that the “SL 3×8” array only provides comparable, yet sometimes not as good, RF performance when compared to “SL 2×8”. In fact, further analysis showed that the B1+ field patterns for the 24 elements of “SL 3×8” did not distinguish as strongly from each other within the ROIs as they did for the 16 elements

of “SL 2×8”, based on a principal component analyses (56) of the B1+ maps using the singular value decomposition approach described in Guerin et al. (25). The maximum cumulative sum of normalized singular values calculated for the “SL 3×8” array was found not to be greater than that for the “SL 2×8” array (i.e., 2.9 vs 3.1 for head/C-spine, 2.8 vs 3.2 for T-spine, and 4.3 vs 4.9 for pelvis). In addition, the elements in the “SL 3×8” array are shorter (145 vs 220 mm) and their B1+ fields attenuate more rapidly towards the medial z-axis of the body. Consequently, the “SL 3×8” array is likely to require higher RF power when trying to achieve the same average flip angle in the middle of the body; to some extent, this may explain the higher peak local SAR values observed with the “SL 3×8” array when the ROI was positioned in a “deep” location inside the body (e.g., in the T-spine case). These results indicate that EM field spatial distribution of individual coil elements is as important as the total channel count in ultimately determining the RF performance of a given coil array design.

Unlike conventional single-slice two-spoke pulse design (14,57) which restricts the center of each spoke to the $k_z=0$ plane, our two-spoke pulse design allows the two spokes to move along the k_z dimension, additionally to the k_x-k_y plane, thereby increasing the degrees of freedom to achieve a more homogeneous B1+ field in a 3D ROI. This is similar to the 3D gradient blips proposed with the k_T point pulse design for non-selective RF excitation (15).

In the current work, the spoke location for all two-spoke pulse designs with global SAR control was determined through an extensive search, which was previously shown capable of producing pulses of improved excitation performance with total RF power constraint (14,38). In Fig. 3, each cluster with a same color indicates a large variation in excitation fidelity and the resultant global SAR when varying the spoke location for a specific λ ; similar behavior of the RF performance in relation to the spoke location is also observed when designing multi-slab total-RF-power-constrained pulses for cerebral time of flight angiography at 7T (38), demonstrating the importance of optimizing the spokes location in pTx pulse design. Alternatively, more advanced algorithms for joint design of spoke trajectories and RF pulses (58–62) can be considered to optimize the spoke locations.

One limitation of the current study is that the spoke locations utilized for the two-spoke pulse design with local SAR control were borrowed from the k-space optimization for global SAR controlled pulse design and therefore may be suboptimal. However, it is reasonable to expect that these spoke locations would lead to better RF performance than an arbitrarily defined placement; this is supported by our initial observation that the global SAR and the peak 10g SAR of these pTx pulses are highly correlated – the pulse design parameters leading to a reduction in one are likely to also decrease the other. This was also confirmed by the superior performance of the designed pulses in comparison to those obtained with a few other randomly chosen spoke locations. Considering that solving the local-SAR-regularized MLS problem through an exhaustive search over a large number of placement candidates would be prohibitively long with our current pulse design implementation and computational capability, borrowing the optimum spoke placement from the global SAR regularized pulse design served as an attractive workaround.

For this initial study, it was considered reasonable to choose a medium-sized male adult model to load the coil arrays in EM simulations. With this model, our results suggest that the “SL 2×8” array represents the most optimal coil design out of the four arrays compared. Further investigation would be necessary to determine the potential impact, on these results, of using human models of different size and gender.

In summary, we have demonstrated based on electromagnetic simulations of various body transmit RF coils at 3T that pTx spoke pulse designs with multichannel transmit coils can achieve volumetric excitation homogenization simultaneously with substantial local and global SAR reduction compared to a standard single-channel body coil design. The gains in both B1+ homogeneity and SAR parameters are substantial even for the body coil driven in a two-channel transmit mode but are most pronounced for the double-ring RF array design with coil elements arranged on two transaxial rings stacked in the z direction.

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Abbreviations used

pTx	parallel transmission
RF	radiofrequency
B1+	transmit B1 field
SAR	Specific Absorption Rate
ROI	Region Of Interest
SENSE	SENSitivity Encoding
BC	birdcage
CP	Circularly Polarized
EM	electromagnetic
FDTD	Finite Difference Time Domain
VOP	Virtual Observation Point
FA	Flip Angle
RMSE	Root Mean Square Error
CV	Coefficient of Variation
B0	static magnetic field inhomogeneity
C-spine	cervical spine
T-spine	thoracic spine.

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Figure 1.

Outline of the five coil structures modelled and compared. The birdcage (BC) coil is a 32-rung shielded high pass and inductively coupled coil, of which the two RF feeding ports are marked with yellow arrows. The other four coils are all stripline (SL) arrays, which consist of a number of stripline elements arranged either in a conventional single-ring manner (i.e., SL 1×8 and SL 1×16), or in a multi-ring fashion (i.e., SL 2×8 and SL 3×8).

HEAD/C-SPINE **T-SPINE** **PELVIS**

Figure 2.

The region of interest (ROI) defined for the three anatomical targets considered. To facilitate the visualization of these volumes, anatomical structures are made visible through a plane arbitrarily chosen inside each cubical ROI. Color scale is arbitrary.

Figure 3.

Optimization of k-space spoke placement in global SAR controlled two-spoke pTx pulse design targeting the T-spine. (a)–(e): For each coil, each dot shown in the global SAR vs. RMSE space represents an RF solution found for a specific combination of a λ value and a k-space placement, and the RF solutions calculated for the same λ are all in a same color; the black L-shaped curve traces the optimal RF solutions for individual λ values, and the corresponding k-space placements were selected for producing two-spoke pTx pulses. (f): Curves tracing the optimal RF solutions are displayed for all coils (y-axis scaled differently from (a)–(e)).

Figure 4.

L-curves quantifying tradeoffs between excitation error (measured by RMSE) and the global SAR when using the pTx enabled coils to design *global* SAR regularized one-spoke (top row) and two-spoke (bottom row) pulses for head/C-spine (a), T-spine (b) and pelvis (c). In each case, the corresponding peak 10g SAR (which was not controlled in the pulse design) is also shown. The single dot in each plot refers to the constant RF performance of the birdcage (BC) coil when operated in the fixed quadrature mode.

Figure 5.

L-curves quantifying tradeoffs between excitation error (measured by RMSE) and the peak 10g SAR when using the pTx enabled coils to design *local* SAR regularized one-spoke (top row) and two-spoke (bottom row) pulses for head/C-spine (a), T-spine (b) and pelvis (c). In each case, the corresponding global SAR (which was not controlled in the pulse design) is also shown. The single dot in each plot refers to the constant RF performance of the birdcage (BC) coil when operated in the fixed quadrature mode.

Figure 6.

Comparison of peak 10g SAR at a similar level of excitation fidelity when designing local SAR regularized *one*-spoke pulses for T-spine imaging. Top row presents the coronal and sagittal maximum intensity projection (MIP) of the 10g SAR distributions within the whole body. Although the reference RMSE was set to 0.04, the performance corresponding to RMSE=0.045 was chosen for the “BC 90°” (i.e. the pTx mode of the birdcage (BC) coil) because this was the minimum RMSE that could be achieved by “BC 90°” when designing local SAR regularized one-spoke pulses.

Figure 7. Comparison of peak 10g SAR at a similar level of excitation fidelity when designing local SAR regularized *two*-spoke pulses for T-spine imaging. Top row presents the coronal and sagittal maximum intensity projection (MIP) of the 10g SAR distributions within the whole body.

Figure 8. Comparison of excitation fidelity at similar peak 10g SAR when designing local SAR regularized *one*-spoke pulses for T-spine imaging. The reference local SAR was set to 6.11 W/kg. The corresponding flip angle distributions in the central coronal and central sagittal slice are presented, as well as some other helpful metrics are reported.

Figure 9.

Comparison of excitation fidelity at similar peak 10g SAR when designing local SAR regularized *two*-spoke pulses for T-spine imaging. The reference local SAR was set to 6.11 W/kg. The corresponding flip angle distributions in the central coronal and central sagittal slice are presented, as well as some other helpful metrics are reported.

TABLE I

Creation of virtual observation points

		BC 1×2	SL 1×8	SL 1×16	SL 2×8	SL 3×8
HEAD	Absolute Upper Bound ¹	0.02 W/kg				
	Relative Upper Bound ²	12.5%	4.3%	5.3%	4.8%	8.8%
	Number of VOPs	16	128	129	33	36
SPINE	Absolute Upper Bound	0.1 W/kg				
	Relative Upper Bound	74%	13%	10%	11%	15%
	Number of VOPs	1	8	14	13	10
PELVIS	Absolute Upper Bound	0.07 W/kg				
	Relative Upper Bound	51%	9.4%	8.2%	7.6%	12.1%
	Number of VOPs	2	18	31	28	9

¹ Absolute upper bound gives the maximum possible SAR overestimation when using the created VOPs to estimate the peak 10 g SAR for a total RF input power of 1 W. For a given anatomical target, this parameter was kept constant across coils to provide comparable accuracy in SAR control during pulse calculation.

² Relative upper bound is absolute upper bound divided by the highest possible 10 g SAR for a total RF input power of 1 W. This parameter was specified to create VOPs and was adjusted to ensure a constant absolute upper bound for a given anatomical target.