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Parallel transmission to reduce absorbed power around deep brain stimulation devices in MRI: Impact of number and arrangement of transmit channels

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Abstract

Purpose: To assess the mean and variance performance of parallel transmission (pTx) coils for reduction of the absorbed power around electrodes (APAE) in patients implanted with deep brain stimulation devices (DBS).

Methods: We simulated four pTx coils (8 and 16 channels, head and body coils) and a birdcage body coil. We characterized the radiofrequency safety risk using the absorbed power around the DBS electrodes (APAE), which is the integral of the deposited power (in Watts) in a small cylindrical volume of brain tissue surrounding the electrode tips. We assessed the APAE mean and variance by simulation of five realistic DBS patient models that include the full DBS implant length, extracranial loops and the implanted pulse generator.

Results: PTx coils with 8 (16) channels were able to reduce the APAE by >18X (>169X) compared to the birdcage coil in average for all patient models, at no cost in term of flip-angle uniformity nor global SAR. Moreover, local pTx coils performed significantly better than body arrays.

Conclusions: PTx is a possible solution to the problem of radiofrequency heating of DBS patients when performing MRI, but the large inter-patient variability of the APAE indicates that patient-specific safety monitoring may be needed.

Introduction

Deep brain stimulation (DBS) is an established therapeutic solution for treatment resistant movement disorders (1–3), epilepsy (4–7), and has been proposed for the treatment of psychiatric disorders (8–15). Unfortunately, deployment of DBS in a wide array of mental disorders is hindered by a lack of understanding of the mechanisms of action of DBS (16–

19). MRI is an ideal imaging modality to shed light on such mechanisms. Unfortunately, MRI is not widely used in this patient cohort because of the potential safety risks, including heating of tissue due to radio-frequency (RF) currents induced on the DBS implant (20,21). This effect has been characterized in multiple experimental and simulation studies (22–35). Parallel transmission (pTx) has been proposed to mitigate this problem as pTx coils and pulses offers additional transmit degrees-of-freedom that can be used to shape the B_1^+ distribution – which controls the flip-angle – as well as electric fields induced in the patient – which cause heating. For example, Etezadi-Amoli et al built a 4-channel planar pTx system and showed that it is possible to generate transmit waveform combinations minimizing the current induced in a guidewire equipped with a current sensor (36). Eryaman et al used a birdcage coil driven at its two decoupled ports and found the shim weights that minimized the temperature increase at the tip of a simple copper wire implanted subcutaneously in an anesthetized pig (30). They also showed that the relative phase difference between the shim weights could be optimized by minimization of the flip-angle artifact induced by the implant on a gradient-echo image. In 2013, Gudino et al reproduced the 4-channel planar pTx coil setup of Etezadi-Amoli et al, but this time used a coil driving function derived from an electromagnetic (EM) simulation of the implant to find the complex-valued shim weights yielding the minimum induced temperature at the tip of a guidewire (37). In addition to these experimental studies, Eryaman et al (38), Guerin et al (39) and McElcheran et al (40) performed electromagnetic simulation of DBS patients in pTx coils also indicating the potential of pTx to reduce the RF MRI safety concern in this patient cohort.

In this work, we use our recently developed virtual DBS population of five clinically-derived, realistic DBS body models and systematically (i) assess the potential of pTx to reduce the lead-tip heating risk and (ii) evaluated the variance of the power absorbed around the DBS electrodes in this patient cohort. We simulated 3 Tesla (123 MHz) MRI because it is likely to be the field strength of choice for functional studies of the mechanisms of action of DBS therapy in psychiatric disorders (41).

METHODS

Computational models of DBS human body and pTx coil

We simulated the five DBS patient models shown in Fig. 1A, which were created as explained in detail in Ref. (42). {Guerin, 2018 #1438} All patients are implanted bilaterally (left and right). Janis, Kurt and Michael (fictitious names) have a single IPG driving both the left and right leads. Freddie and Angus have two IPGs each driving the left and right leads independently. In addition, Angus has two abandoned leads not connected to any IPG. In all cases, we built a model of a generic DBS implant around the geometrical path, which comprise four electrodes (1.27 mm diameter, 1.5 mm long) each connected to the IPG using straight wires. The IPG was modeled as a rectangular box with size 1”×1”×0.25”. The CT volumes were segmented into three classes (internal air, soft tissue and bone) using 3D Slicer (43) followed by manual cleanup (42). Segmented tissue voxel volumes were then meshed into 2-manifold, watertight surface meshes using a previously described procedure (44). The final mesh and coil model were imported in Ansys Electronics (Canonsburg, PA).

We modeled five radio-frequency (RF) coils shown in Fig. 1B. The birdcage coil (BC) is a high-pass, 32-rung coil driven at two ports located at 0° (port 1) and 90° (port 2). The BC coil was driven in quadrature mode (equal amplitudes on ports 1 and 2, 90° phase difference) as well as in pTx mode, whereby the waveforms played on ports 1 and port 2 are optimized individually. We also simulated four pTx coils: two 8-channel coils (one for body imaging, one for dedicated head imaging) and two 16-channel coils (one for body, one for head imaging). The 16-channel coils were arranged in two rows of eight channels because in a previous study (not specific to DBS) we found that such arrangement leads to a beneficial power vs flip-angle tradeoff compared to a single row of 16 channels (45). The body pTx coils comprised rectangular loops on a cylindrical former. The head pTx coil loops had a curved shape bending toward the head in the +z direction in order to stay as close to the body as possible, thus maximizing transmit efficiency (Fig. 1B).

Electromagnetic simulation

We computed the electromagnetic (EM) field induced by each coil in the five patient models using a co-simulation process based on Ansys Electronics, previously described in detail in Ref. (46,47). The body models were centered as would be done in practice by placing the spot between the eyes at the coil isocenter. The values of the tuning capacitors of the BC coil were known from the physical coil and were modeled as such in the Ansys Electronics field simulation step; however, ports 1 and 2 were modeled as lumped ports. For the pTx coils, tuning capacitors were not known in advance and were therefore modeled as lumped ports in the co-simulation process. Specifically, there were eight tuning capacitors per channel for the 1/8-body design and four tuning capacitors per channel in the 1/8-head, 2/8-body and 2/8-head designs. Maximum tetrahedron and triangle edge length constraints were imposed in regions of the simulation domains where the electric field was known to vary rapidly. Triangle edge lengths were constrained to <8 mm, <50 mm and <30 mm on the coil, shield and IPG surfaces, respectively. Tetrahedron edge lengths were constrained to <15 mm inside the body volume (all tissue classes), except in a rectangular volume surrounding both electrode tips and in smaller cylinders closely surrounding the electrode tips, where the maximum edge length constraints were set to 3 mm and 0.5 mm, respectively (see Ref. (42) for additional details). Mesh operations were applied during the initial meshing of the model by Ansys Electronics and then refined iteratively until reaching convergence of the S-parameters (5% convergence tolerance) as well as the APAE metric (3% convergence tolerance). The APAE is the absorbed power around the electrodes, is expressed in Watts and is integrated over small cylindrical volumes surrounding the left and right electrodes (42).

For each coil and body model simulation, the output of the field simulation was (i) the EM fields created by the individual ports and (ii) the scattering matrix (S-matrix) of the coil. The S-matrix was loaded in the Ansys Electronic circuit simulator, where tuning and matching capacitors were reinstated. Tuning was performed at 123 MHz (3 Tesla) and matching to <-20 dB for all channels. We did not attempt to decouple the transmit channels of the pTx coils, as would be done with a physical coil; instead, we assumed ideal decoupling across all channels, which can be done in practice using inductive or capacitive decoupling (48) (49–51) or a special decoupling matrix placed between the RF power amplifiers and the coil (52). In our simulations, ideal decoupling was achieved by electrically isolating each channel of a

coil in turn. This was done by assigning the correct tuning and matching capacitor values to the channel under consideration but assigning very small capacitor values (large impedance) to the other channels, thus preventing current from flowing and yielding negligible coupling with the active channel (<50 dB).

The tetrahedron mesh using internally by Ansys Electronics is efficient for solving Maxwell's equations but is not easy to manipulate for post-processing; therefore the EM fields were exported on a voxel grid. The magnetic (B) and electric (E) field values were exported on a 2 mm isotropic resolution grid covering the entire body model. In addition, E fields were exported on a 0.1 mm isotropic voxel grid covering the tip of the left and right electrodes (and, for Angus, the two abandoned lead tips). We selected such high spatial resolution because the E field is known to vary extremely rapidly around the DBS electrodes, therefore accurate prediction of RF safety metrics in this region requires a much higher resolution than typically used when evaluating SAR in patients without implants (35). The magnitude and phase values of the B field were used to compute the B_1^+ maps and the magnitude values of the electric field were used to compute global SAR and the APAE matrix (see details below). The E-fields and B_1^+ maps of the individual channels were then combined by linear superposition using the multi-channel RF pulse as weight factors (see next section).

Pulse design

PTx RF pulses were optimized as follows:

$$\begin{aligned} \min_{\mathbf{x}} \|\mathbf{M}(\mathbf{x}) - \mathbf{M}_{tar}\|^2 \quad [1] \\ s.t. \\ a) \mathbf{x} \mathbf{G} \mathbf{x}^H \leq GSAR_{max} \\ b) \mathbf{x}(\mathbf{P}_L + \mathbf{P}_R) \mathbf{x}^H \leq P_{max}, \end{aligned}$$

Where $\mathbf{x}$ is the RF waveform played on all the transmit channels stacked in a single column vector, $\mathbf{M}(\mathbf{x})$ is the magnetization distribution produced by the RF pulse $\mathbf{x}$ and the B_1^+ maps of the different transmit channels and $\mathbf{M}_{tar}$ is the target magnetization amplitude given by $\mathbf{M}_{tar} = \sin(FA)$, where FA is the target flip-angle (equal to 10° in this work). Since we consider small tip-angle pulses, $\mathbf{M}(\mathbf{x})$ is a linear function of $\mathbf{x}$:

$$\mathbf{M}(\mathbf{x}) = \mathbf{A} \mathbf{x} ; [\mathbf{A}]_{i,(jc)} = i\gamma M_0(\mathbf{r}_i) B_{1+,c}(\mathbf{r}_i) \exp\{-2\pi i \mathbf{k}(j\Delta t) \cdot \mathbf{r}_i\}, \quad [2]$$

Where i, j and c are the spatial coordinate, time and channel indices, respectively; γ is the gyromagnetic ratio and $M_0(\mathbf{r}_i)$ is the steady-state magnetization at location $\mathbf{r}_i$. $B_{1+,c}(\mathbf{r}_i)$ is the complex B_1^+ map of channel c at location $\mathbf{r}_i$ and $\mathbf{k}(j\Delta t)$ is the transmit k-space location at time j Δt , which depends on the gradient waveforms played during the excitation pulse (53–55). We designed slice-selective B1-shimming pulses with a single spoke, therefore $\mathbf{k}(t)$ only

has one non-zero component representing fast traversal in the slice-select direction. The objective function in Eq. [1] is a magnitude least-squares term since this leads to superior flip-angle uniformity compared to least-squares design (56).

Constraint a) in Eq. [1] limits the global SAR induced in the entire body model. The limit GSAR_{max} was set to the global SAR value of the BC coil when driven in quadrature mode (depends on the specific patient model). $\mathbf{G}$ is the global SAR matrix computed as:

$$\mathbf{G} = \frac{1}{N} \sum_{i=1}^N \frac{\sigma(\mathbf{r}_i)}{2\rho(\mathbf{r}_i)} \mathbf{E}(\mathbf{r}_i) \mathbf{E}(\mathbf{r}_i)^H, \quad [3]$$

Where $\sigma(\mathbf{r}_i)$ and $\rho(\mathbf{r}_i)$ are the conductivity and density values at location $\mathbf{r}_i$, respectively; and $\mathbf{E}$ is the $C \times 3$ matrix of the x, y and z components of the electric fields induced by the C channels of the coil at location $\mathbf{r}_i$. N is number of voxels in the body model. The global SAR matrix computation spatial mask excluded voxels corresponding to the DBS implant as well as voxels located in spheres of diameter 3 cm centered on the left and right DBS electrodes.

Constraint b) is the APAE constraint that limits the power absorbed in the tissues surrounding the left and right electrode to P_{max} . The power matrix $\mathbf{P}_R$ was computed as:

$$\mathbf{P}_R = \sum_{i=1}^M \sigma(\mathbf{r}_i) \mathbf{E}(\mathbf{r}_i) \mathbf{E}(\mathbf{r}_i)^H, \quad [4]$$

Where M is the number of voxels in the APAE mask, which is a cylinder tightly surrounding the right electrode (6 mm diameter, 24 mm length). The power matrix $\mathbf{P}_L$ (left electrode) was computed in a similar fashion (for Angus, two additional power matrices were computed corresponding to the left and right abandoned leads, which were also included in constraint b)). In addition, maximum RF voltage was limited to 250 V on every channel for the pTx coils.

We used this pulse design framework for the excitation of a transverse slice at isocenter, a transverse slice located at isocenter +4 cm in the Z (=B0) direction and a coronal slice passing through the left and right DBS electrodes tips (all slices have a thickness of 5 mm). For each slice orientation and each patient model, the pulse design B_1^+ computational mask included the DBS leads but excluded the extension cables. This is because DBS leads are intra-cranial and therefore cause flip-angle artifacts located toward the center of the slice that should be minimized in the pulse design process. In contrast, extension cables are located right below the skin and create a flip-angle artifact at the edge of the slice that is less important to correct.

Statistical assessment of coil-coil differences

To objectively assess coil performance, we use the L-curve formalism whereby we plot the APAE as a function of the flip-angle (FA) error. For each patient and coil model, L-curves

were obtained by designing several RF pulses with varying maximum APAE constraint values, while keeping the global SAR constant and equal to that associated with the quadrature BC coil (BC-quad). Thus, each point on the curve is a different optimized RF pulse tradeoff between APAE and FA error. We tested the statistical significance of pairwise coil comparison in the APAE at constant FA error metric as well as the FA error at constant APAE metric using two-tailed paired t-tests. Coil performance means and variances were calculated across the five patient models.

Use of generic DBS models to control the APAE

We also designed pulses using the B1+ maps of the model Janis, however using APAE matrix associated with the other body models. This was done to assess whether APAE control can be done with pTx using generic DBS patient models.

RESULTS:

Fig. 2 shows APAE vs FA error L-curves for the design of a transverse slice excitation located at isocenter in the Janis model with (Fig. 2B) and without (Fig. 2A) a global SAR constraint. These L-curves show that the ranking of coils depends on whether or not a global SAR constraint is active. Moreover, the global SAR vs FA error L-curve of Fig. 2A shows that constraining global SAR is crucial: Otherwise the pulse design algorithm generates excellent FA distributions at very small APAE, however at the cost of a 38-fold increase in the global SAR as compared with BC-quad (2/8-body), which is unacceptable (Supporting Information Fig. S1). The lower row of the 2/8-body array overlaps the most with the DBS implant and therefore driving these channels tends to create large APAE at the lead tip. In other words, the B_1^+ sensitivity per unit square-root APAE is smaller for the lower row than for the upper row of the 2/8-body coil. In contrast, the B_1^+ sensitivity per unit square-root APAE for the 1/8-body array is intermediate (between that of the upper and lower rows of 2/8-body). Since global SAR varies quadratically with the RF amplitude, a lower B_1^+ sensitivity per unit square-root APAE for the lower row of the 2/8-body array results in a disproportionately large global SAR compared to 1/8-body (this can be seen visually on Supporting Information Fig. S1 & S2, where SAR is much greater in the lower part of the model, i.e. the neck and shoulders).

In Fig. 3, we show the APAE vs FA error L-curves for all coils and all patients for the transverse slices design off isocenter (similar results for the transverse slice at isocenter and the coronal slice are shown in Supporting Information Figures S3 & S4). For the transverse slice at isocenter, the coil ranking is 2/8-head > 1/8-head $\approx$ 2/8-body > 1/8-body > BC-pTx > BC-quad for most patient models (the symbol “A>B” denotes “coil A performs better than coil B”). For the transverse slice off isocenter, the performance of 2/8-body became comparable to that of 2/8-head, resulting in the following ranking: 2/8-head $\approx$ 2/8-body > 1/8-head > 1/8-body > BC-pTx > BC-quad. Finally, for the coronal slice imaging target, the ranking was similar than that for the transverse slice at isocenter (2/8-head > 1/8-head $\approx$ 2/8-body > 1/8-body > BC-pTx > BC-quad). These results show that it is beneficial to use 16 channels instead of 8 and that using a head coil as opposed to a body coil is beneficial for

achieving low APAE at good FA, which expected since local coils cover a smaller extent of the DBS implant than body coils.

FA maps (at constant APAE), absorbed power maximum intensity projection maps (at constant FA error) and APAE reduction factors compared to the BC coil are shown in Figs. 4, 5 and 6 for the design of a transverse slice off isocenter. APAE reduction factors for all coils and body models and design of a transverse slice at isocenter and a coronal slice are shown in Supporting Information Figures S5 & S6, respectively. Maps of the absorbed power in mW show the dramatic reduction of APAE that can be expected when using pTx in this patient cohort. APAE reduction factors were as high as 330X (Fig. 6), 258X (Fig. S5) and 169X (Fig. S6) compared to BC-quad, at constant FA error and global SAR. In average across body models, the APAE levels at constant FA error for the BC-pTx, 1/8-body, 1/8-head, 2/8-body and 2/8-head coils were 46%, 37%, 6%, <1% and <1% that of the BC-quad APAE level, respectively, for design of a transverse slice off isocenter ($Z=4$ cm). In other words, using either an 8-channel head coil or 16 channels arranged either on a body or head former led to a reduction of the APAE of more than 94% compared to the BC-quad at constant FA uniformity and global SAR. For design of a transverse slice at isocenter, the APAE levels of the BC-pTx, 1/8-body, 1/8-head, 2/8-body and 2/8-head coils were 49%, 57%, 17%, 5% and 1% than that of BC-quad in average across patients. For the design of a coronal slice, the APAE levels (in the same order of coils) were 44%, 4%, 1%, <1% and <1% than that of BC-quad at constant FA error and global SAR. In short, pTx provides large reductions in APAE. For example, >169X reduction in the APAE can be achieved using pTx with 16 channels for all slice orientations, without penalty in term of FA uniformity or global SAR.

Despite the impressive performance of pTx coils, there was a tremendous variability of the APAE in the different patient models. The BC-quad APAE was ~2.2, ~0.3, ~7.0, ~1.6 and ~3.4 mW for Kurt, Janis, Freddie, Michael and Angus, respectively (this is relatively constant across imaging slice orientations), which represents a variability of ~88% of the APAE across patient models (standard deviation divided by mean). In contrast, the FA error variability across patient models for BC-quad was ~8%. In addition to the large APAE BC-quad variability across patients, the ability of pTx to reduce APAE even when compared to the BC-quad APAE level also varied significantly. For example, as shown in Fig. 6, the impact of pTx on APAE seemed more pronounced for Freddie than for the other patient models. Figs. 7B quantify this more rigorously by showing the mean $\pm$ standard deviation of the APAE levels at constant FA error for all coils, patient models (design of a transverse slice at isocenter + 4 cm. APAE reduction data at constant FA error for all coils and body models and the transverse slice at isocenter and for a coronal slice are shown in Supporting Information Figures S5 & S6). Fig. 7A, S7A & S8A also show the the mean $\pm$ standard deviation of the FA error at constant APAE levels for the transverse slice at isocenter + 4 cm, transverse slice at isocenter and the coronal slice, respectively.

Because of the large variability of APAE across patient models, not all coil comparisons were statistically significant. Table 1 shows P-values and statistical power levels for all pairwise coil comparisons in the APAE metric at constant FA error, for the design of a transverse slice off isocenter ($Z=4$ cm). P-values and power of coil-coil comparisons for the

transverse slice at isocenter and the coronal slice are shown in Supporting Information Tables S1 & S2, respectively. The light grey cells indicate coil-coil differences with P-values smaller than 10% and the dark grey cells indicate P-values smaller than 5%. In average across all body models, the 1/8-head, 2/8-body and 2/8-head coils create significantly lower APAE at constant FA error than the BC-quad coil at the 5% significance level. The statistical power of these comparisons is relatively low however (~56%), which indicates that more body models should be simulated to confirm this conclusion. The relations 1/8-head > 1/8-body, 2/8-body > 1/8-body and 2/8-head > 1/8-body are more robust (P-value<5%, power~70%). This suggests that head coils are better suited than body coils for reduction of the heating risk in DBS patients and that, if possible, a 2/8-head coil should be used.

Fig. 8 shows L-curve results for pulses designed with the B_1^+ maps of Janis but the APAE matrices of the other models (in the APAE reported on the Y-axis of Fig. 8 is the true APAE computed using Janis' E-fields however). Using the APAE matrices of Kurt, Michael, Freddie or Angus did not allow control of the APAE in Janis for any of the pTx coil simulated as the resulting L-curves were always quite far from the ultimate design strategy (use of Janis' APAE matrix and B_1^+ maps). In order to assess whether an ensemble of generic body models could lead to a robust control of the APAE in Janis, we also designed pulses while controlling the APAE in Kurt, Michael, Freddie and Angus simultaneously (black dash line). This was done by using four APAE constraints in Eq. 1, instead of a single one as in the designs presented so far. The results show that this is not a useful strategy either, as the APAE vs FA error are always sub-optimal and sometimes worse than the simple unconstrained solution.

DISCUSSION:

In this work, we computed the power absorbed in soft tissues surrounding the DBS implant tip (absorbed power around the electrodes, APAE) as a metric reflecting RF heating risk in DBS patients at 3 Tesla. We compared pTx transmission to the typical quadrature birdcage (BC-quad) coil excitation. The expected variability of the APAE within the patient population was assessed using five realistic DBS anatomical body models that were derived from clinical CT data. Thus, we could quantify not only the average APAE reduction due to pTx, but also its variability. We found that pTx coils are able to dramatically reduce the APAE compared to a BC body coil. Our statistical analysis of the APAE vs FA error tradeoffs across body models shows that head-dedicated pTx coils (1/8-head and 2/8-head) consistently outperform body pTx arrays (1/8-body and 2/8-body). When using 8 transmit channels, the APAE was reduced, in average across all patient models, by 29X, 235X and 18X compared to BC-quad when designing a transverse slice off isocenter ($Z=4$ cm), a coronal slice and a transverse slice at isocenter, respectively. When using 16 channels, the average APAE reduction factors at constant FA errors were even greater: 330X, 258X, and 169X for the transverse slice off isocenter ($Z=4$ cm), the coronal slice, and the transverse slice at isocenter design problems, respectively. Such APAE reductions came at no penalty in terms of flip-angle uniformity or global SAR. We point out that this study includes body models with one and two IPGs, and even one model (Angus) with two abandoned DBS leads in addition to the two functional leads. The DBS patients modeled in this work have very different extra-cranial DBS cable loop patterns and geometrical paths in the head, neck and

upper torso; which gives some level of confidence that our coil ranking and average APAE reduction predictions are robust across a large segment of the DBS patient population.

Despite these results, statistically robust pairwise coil differences that are valid for the entire DBS population were difficult to establish because of the large variability of the APAE metric. This is due to the fact that the APAE depends on the electric field, which is much more variable across body models than the magnetic field. Using five body models gave us a good estimate of the APAE variability and enough statistical power to draw definite conclusion on the following coil ranking: 1/8-head > 1/8-body, 2/8-body > 1/8-body and 2/8-head > 1/8-body (P-value<5%, power≈70%). Two other coil relationships were almost significant, but did not pass the strict 5% p-value test threshold: 2/8-body > 1/8-head (p=5.9%, power=51.3%) and 2/8-head > 1/8-head (p=6.2%, power=49.9%). In other words, head arrays are better than body arrays for reducing the heating risk in DBS patients and 16 channels arranged in 2 rows (2/8-body) is better than a single row of 8 channels.

The statistical power of our coil comparisons was limited by the number (N=5) of body models available. Based on the APAE variance estimates obtained in this work, we predict that using N=9 patient models would give us >80% statistical power for all coil-coil comparisons for the transverse slice off isocenter design problem, except for BC-quad / BC-pTx, BC-pTx / 1/8-body, 2/8-body / 2/8-head. For the design of a coronal slice, N=11 would give us >80% power for all coil-coil comparisons except for BC-quad / BC-pTx, 1/8-head / 2/8-body and 1/8-head / 2/8-head. In other words, using twice as many patient models would significantly increase the power of our statistical comparisons. We also point out that our statistical testing was two-tailed, which is somewhat conservative because this is equivalent to assuming that we did not know in advance which coil performs better than others. In reality, it is safe to assume that pTx coils outperform BC-quad, which would justify the use of one-tail t-tests thus improving statistical significance (when doing so we find that the 1/8-body > BC-quad, 1/8-head > BC-quad, 2/8-body > BC-quad and 2/8-head > BC-quad are all significant with P<3.5% and power>64%). In addition, pTx coils were able to significantly reduce the flip-angle artifact around the DBS lead. Statistical analysis of coil-coil comparisons in this metric are shown in the Supporting Information Tables S3, S4 & S5.

The focus of this study was the characterization of pTx performance in DBS patients as compared with the BC coil. Another interesting comparison is that with the unconstrained case, whereby the APAE metric is not considered in the pulse design process. This data point is actually present in all the L-curves presented in this work as the first point of the L-curves corresponds to the unconstrained case. As shown in Fig. 3, S3 and S4, for most slice orientation/body model/coil combinations, the APAE vs FA error tradeoff was better for pTx coils than the birdcage coil, even when not constraining the APAE in the pulse design process. The exceptions are for the 1/8-head, 2/8-body and 2/8-head pTx coils for design of a transverse slice at isocenter (Fig. S3). Moreover, for all coils, body models and slice orientations, constraining the APAE in the pulse design yielded a dramatic reduction of the APAE at the cost of a modest worsening of the FA excitation quality.

In this work, we quantified the risk of RF-induced heating of tissue using the APAE metric. We chose this metric because it is more tightly correlated to the peak induced temperature

than the non-averaged and averaged peak SAR and does not require a long and complex temperature solve. Peak SAR is arguably not a natural metric to use when quantifying RF heating in DBS patients because the unaveraged SAR distribution decreases very rapidly around the DBS lead tip. It therefore makes little sense to average it over the traditional 10 grams or 1 gram volume of brain tissue. Ideally, temperature should be used but this is not an easy safety metric to compute in this class of patients. A major difficulty lies in the fact that ultra-high resolution (0.1 mm uniform grid) needs to be used to resolve the temperature variation around the DBS electrodes. Such high resolution is not compatible with the use of uniform grids, at least not over the entire patient model which would lead to unreasonably long simulation times. Instead, finite element modeling of the temperature distribution on a tetrahedron discretization can be used since this method allows efficient modeling of both fine and large scale structures in the same computational domain. We are currently working on this issue, but in the meantime APAE provides an excellent, easy to compute alternative.

Exact knowledge of the E-fields was used in the pulse design process in this work. In practice, the APAE matrix could potentially be replaced by a “ B_1^+ artifact reduction matrix”, which can be derived from image data (57). Another possible strategy is to compute the APAE matrices for a number of generic body models and use those when designing pulses for actual DBS patients. This strategy is inspired by the virtual observation point approach now in use for control of local SAR in pTx (58). However, as we show in Fig. 8, this strategy does not work for DBS patients: Indeed, using the APAE matrices of other models than Janis in the pulse design process yielded sub-optimal APAE vs FA error tradeoffs, sometimes even worse than when using a simple unconstrained strategy (no APAE constraint).

A limitation of this study is that we did not model the internal helicoidal structure of the DBS implant. As shown in Ref. (35,59), this likely has a major impact on the predicted APAE. Ideally, the internal components of the DBS implant should be modeled as accurately as possible and include helicoidal internal wires, however this is exceedingly difficult even using state-of-the-art electromagnetic solvers such as Ansys Electronics. In our experience, the internal meshing algorithm of Ansys Electronics systematically fails when the pitch of the internal wire helix becomes too small – in Ref. (35) we were only able to model a 2 mm pitch helix, which is much greater than the ~0.3 mm pitch used in the actual devices. This is even more difficult when modeling realistic, non-straight implant lead geometries with extra-cranial loops (Fig. 1A). We do not expect this to have a significant impact on the relative ranking of coils however, as all coils should be affected similarly by this simplification. Other limitations of this study are those traditionally associated with electromagnetic simulation of RF coils. Simulations are necessarily simplified representations of actual coils. For instance, we did not simulate the cables connecting the individual transmit channels to the RFPAs nor did we model lossy lumped elements (all lumped elements modeled here are idealized components) and distributed matching networks. To simplify our simulations, we modeled ideal coupling in this work, which is also a simplification of the reality since in practice it may be difficult to perfectly decouple the multi-row arrays with 16 channels (2/8-body and 2/8-head). Additional validation work on phantom is needed to assess the magnitude of these deviations, which we plan to do in the future. Despite these limitations, the simulation study is valuable as it provides a

theoretical basis for choosing the best pTx coils for studying the mechanisms of action of DBS.

CONCLUSION:

In this work, we characterized the expected reduction in power absorbed around DBS electrodes (APAE) when using parallel transmission (pTx) as opposed to the more traditional birdcage coil (BC) at 3 Tesla. Our results show that pTx coils with 16 channels allow reduction of the APAE at constant flip-angle uniformity by more than 169X for all slice orientations. Importantly, this result was statistically significant, which we assessed by modeling N=5 realistic DBS patient models developed previously by our group (42). Using our estimation of the APAE variability across the N=5 realistic DBS patient models, we found the following statistically significant coil ranking relationships: (i) Using 2 rows of 8 channels is better than using a single row of 8 Tx channels and (ii) head pTx coils perform better than body pTx coils. Note that these APAE reductions obtained with pTx (compared to the BC coil) came at no cost in term of global SAR and flip-angle uniformity. Despite these good performance by pTx coils, we found that the APAE variance across patient models was large. For example, the APAE induced by the BC coil driven in quadrature had a simulated variability of ~88% (standard deviation divided by mean) across our N=5 body models. In contrast, the flip-angle uniformity variability was only ~8%. These results show that pTx is a promising technology for reduction of the RF heating safety risk in DBS patients at 3 Tesla, but that patient-specific modeling may be needed in order to deal with the large inter-patient APAE variability.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

A: Five DBS patient models simulated in this work. All patients are implanted bilaterally, i.e. they have a DBS lead implanted on the left and right hemispheres. Janis and Kurt have a single implanted pulse generator (IPG) on the right side, whereas Michael has his IPG implanted on the left side. Freddie and Angus have two IPGs driving the right and left leads independently. In addition, Angus has two abandoned leads that are not connected to IPGs. Patient names are fictitious to protect confidentiality. **B:** Coil models simulated in this work. The birdcage coil (BC) has two decoupled ports shown in red and can be driven in either quadrature mode (both ports driven with the same amplitude but with 90° phase differential) or as a pTx coil (ports driven with different amplitudes and phases). The notation X/Y signifies "X rows/Y channels per row". The diameter (D) and height (H) of each coil are indicated in inches.

A. Janis, transverse slice @ isocenter, no global SAR constraint

B. Janis, transverse slice @ isocenter, with global SAR constraint

Fig. 2.

A. L-curves showing the tradeoff between flip-angle (FA) error and both (i) the absorbed power around the electrodes (APAE) and (ii) global SAR. Global SAR is not constrained in these designs. L-curves for all coil models are shown on the same graph. The body model simulated is Janis. APAE is computed in a small volume surrounding the DBS electrodes, whereas global SAR is averaged over the entire head. **B:** Same as A, however with explicit constraint for the global SAR (for all coils, global SAR was constrained to be smaller or equal the BC-quad level).

Fig. 3.

L-curves showing the tradeoffs between APAE and flip-angle error for all patient and coil models. The target design is a uniform 10° flip-angle distribution for a transverse slice at isocenter + 4 cm. Note that the X and Y scales of these graphs are different for different patient models (the Y scale especially, i.e. the induced APAE, varies significantly across patients).

Fig. 4.

Flip-angle maps and absorbed power for all coils/patient models for design of a uniform 10° transverse slice located at +4 cm from isocenter in the Z (=B0) direction. **A:** FA maps at constant APAE (the constant APAE level is that of the BC-quad coil). The numbers below each map indicate the average flip-angle error across the slice in percent. Note that the extension cables are masked on these maps, as these create severe flip-angle artifacts that are difficult to correct (the central DBS lead artifacts are not masked however since these are in the middle of the brain and need to be corrected for). **B:** Maximum intensity projection maps of the absorbed power at constant flip-angle error (the constant FA uniformity level is that of the BC-quad coil). The numbers below each map indicate the peak absorbed power in mW.

Fig. 5.

APAE reduction factors at constant FA error, computed with respect to BC-quad, for the design of a transverse slice located at isocenter + 4 cm. The numbers in brackets indicate the range of APAE reduction factors across patient models for a given coil. The bold underlined numbers above the ALL PATIENT bar indicate the average APAE reduction factor across patient models for a given coil.

Fig. 6.

Flip-angle and APAE improvement capabilities of pTx coils across patient models for design of a transverse slice located at +4 cm away from the isocenter in the Z (=B0) direction. **A:** FA error at constant APAE across patient models (the constant APAE level is that of the BC-quad coil for each patient model). **B:** APAE at constant FA error across patient models (the constant FA uniformity level is that of the BC-quad coil for each patient).

A. Flip-angle error at constant APAE (transverse slice @ iso+4 cm)

B. APAE at constant flip-angle error (transverse slice @ iso+4 cm)

Fig. 7.

APAE vs FA error L-curves for the design of a transverse slice at isocenter + 4 cm in the body model Janis using the different pTx coils. The black L-curves (optimal) were obtained using the Janis APAE matrix in the pulse design, whereas the others were obtained using the APAE matrices of the other models (note however that for all L-curves, the APAE that is reported along the Y-axis correspond to the APAE computed using the E-fields of Janis. In other words, for the Kurt, Michael, Freddie and Angus there is a mismatch between the APAE matrix used in the pulse design and the APAE reported in these graphs). The black dash L-curves were obtained by using the APAE of Kurt, Michael, Freddie and Angus simultaneously in the pulse design (worst case of these four body models). Global SAR was constrained to the value of the BC-quad coil using the Janis global SAR matrix. The blue arrows point to the first pulse of the L-curve, which always corresponds to the unconstrained case (no APAE constraint).

Fig. 8.

APAE vs FA error L-curves for the design of a transverse slice at isocenter + 4 cm in the body model Janis using the different pTx coils. The black L-curves (optimal) were obtained using the Janis APAE matrix in the pulse design, whereas the others were obtained using the APAE matrices of the other models (note however that for all L-curves, the APAE that is reported along the Y-axis correspond to the APAE computed using the E-fields of Janis. In other words, for the Kurt, Michael, Freddie and Angus there is a mismatch between the APAE matrix used in the pulse design and the APAE reported in these graphs). The black dash L-curves were obtained by using the APAE of Kurt, Michael, Freddie and Angus simultaneously in the pulse design (worst case of these four body models). Global SAR was constrained to the value of the BC-quad coil using the Janis global SAR matrix. The blue arrows point to the first pulse of the L-curve, which always corresponds to the unconstrained case (no APAE constraint).

TABLE 1.

STATISTICAL SIGNIFICANCE OF PAIRWISE APAE COIL DIFFERENCES AT CONSTANT FLIP-ANGLE UNIFORMITY FOR DESIGN OF A TRANSVERSE SLICE OFF ISOCENTER (Z=4 CM). P-VALUES (POWER) ARE EXPRESSED IN PERCENT.

	BC-pTx	1/8-body	1/8-head	2/8-body	2/8-head
BC-CP	24.1 (18.8)	7.1 (46.0)	5.0 (55.9)	4.8 (56.9)	4.8 (56.7)
BC-pTx		61.7 (7.1)	6.5 (48.6)	5.6 (52.7)	5.6 (52.8)
1/8-body			2.8 (71.5)	3.1 (69.1)	3.1 (68.7)
1/8-head				5.9 (51.3)	6.2 (49.9)
2/8-body					84.5 (5.3)