

RESEARCH ARTICLE

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RF shimming in the cervical spinal cord at 7 T

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Abstract

Purpose: Advancing the development of 7 T MRI for spinal cord imaging is crucial for the enhanced diagnosis and monitoring of various neurodegenerative diseases and traumas. However, a significant challenge at this field strength is the transmit field inhomogeneity. Such inhomogeneity is particularly problematic for imaging the small, deep anatomical structures of the cervical spinal cord, as it can cause uneven signal intensity and elevate the local specific absorption ratio, compromising image quality. This multisite study explores several RF shimming techniques in the cervical spinal cord.

Methods: Data were collected from 5 participants between two 7 T sites with a custom 8Tx/20Rx parallel transmission coil. We explored two radiofrequency (RF) shimming approaches from an MRI vendor and four from an open-source toolbox, showcasing their ability to enhance transmit field and signal homogeneity along the cervical spinal cord.

Results: The circularly polarized (CP), coefficient of variation (CoV), and specific absorption rate (SAR) efficiency shim modes showed the highest B_1^+ efficiency, and the vendor-based “patient” and “volume” modes showed the lowest B_1^+ efficiency. The coefficient of variation method produced the highest CSF/spinal cord contrast on T_2^* -weighted scans (ratio of 1.27 ± 0.03), and the lowest variation of that contrast along the superior–inferior axis.

Conclusion: The study’s findings highlight the potential of RF shimming to advance 7 T MRI’s clinical utility for central nervous system imaging by enabling more homogenous and efficient spinal cord imaging. Additionally, the research incorporates a reproducible Jupyter Notebook, enhancing the study’s transparency and facilitating peer verification.

KEYWORDS

B_1^+ , radio-frequency, spinal cord, shimming, 7T

For affiliations refer to page 10

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1 | INTRODUCTION

MRI of the spinal cord (SC) is an important diagnostic tool for several pathologies, such as multiple sclerosis,¹ and traumatic injuries.² Accurate diagnosis requires images with both high in-plane resolution and high signal-to-noise ratio (SNR), the acquisition of which can be challenging at common field strengths of 1.5 T and 3 T. Using ultrahigh field (UHF) MRI systems with a field strength of 7 T or greater, these limitations can be overcome and image quality can be improved.³ The benefits of UHF SC MRI have been demonstrated with the visualization of small anatomical structures such as nerve roots, denticulate ligaments, or blood vessels,^{4,5} resulting in higher detection rates of lesions.^{6,7} These potential clinical benefits bring UHF MRI one step closer to a clinical routine use for central nervous system imaging.^{3,8,9}

However, in UHF imaging, care must be taken to address the issue of radiofrequency (RF) transmit, or B_1^+ , field inhomogeneity. At lower field strengths, the characteristic RF wavelength within human tissues is greater than that of the anatomy of interest, and the B_1^+ field can be considered homogeneous, or at least slowly varying. By contrast, at 7 T the wavelength in water is about 11 cm, compared with 26 cm at 3 T.¹⁰ As a result, the B_1^+ field in UHF imaging is inhomogeneous, which results both in inhomogeneous signal intensity and increased local specific absorption rate (SAR).^{11–14} This limits the utility of UHF MRI, particularly for small anatomies located deep within the body, such as the cervical SC.

Parallel transmission (pTx) can be used to overcome this limitation of UHF imaging. In pTx coils, multiple transmit (Tx) elements (channels) are arranged, typically in a circular pattern, as opposed to the single element used at lower fields. These individual elements can be used in circularly polarized (CP) mode, where the same RF waveform is played out on each element, with the same amplitude, and phase offsets such that their B_1^+ fields constructively add at the center of the object. Alternatively, the B_1^+ field can be adjusted, or shimmed, by changing the magnitude and/or phase of the RF waveform played out for each element (static RF shimming) or changing the waveform as well (dynamic RF shimming). In the past decade, several coils have been developed that are capable of pTx operation,^{15,16} and RF shimming has been demonstrated successfully for brain imaging using bespoke coils.^{16–20} However, despite the proliferation of bespoke^{4,15,21–25} and commercially available⁵ coils for the cervical SC, to date, the application of RF shimming in the cervical spine at UHF has only been preliminarily investigated.^{26,27}

In this paper, we demonstrate the feasibility and utility of participant-specific RF shimming for the cervical SC using a bespoke 8-Tx/20-receive (Rx) pTx-capable cervical

spine coil. We evaluate two RF shimming approaches provided by an MRI vendor as well as four approaches implemented in the open-source *Shimming Toolbox*.²⁷ We characterize B_1^+ efficiency and cerebrospinal fluid (CSF) versus SC contrast on gradient echo (GRE) scans across the entire c-spine and upper thoracic region.

2 | METHODS

2.1 | Acquisition protocol

2.1.1 | Participants and facilities

Five participants were scanned for this study at two sites. Two participants (30 year-old male, 31-year-old male) were scanned at the Montreal Neurological Institute (MNI, Montreal, Quebec, Canada), and 3 participants (20-year-old female, 29-year-old female, 25-year-old male) were scanned at the Massachusetts General Hospital's Athinoula A. Martinos Center for Biomedical Imaging (MGH, Charlestown, Massachusetts, USA). All data were acquired using custom 8-Tx/20-Rx cervical SC arrays²⁵ and 7T Magnetom Terra scanners (Siemens Healthineers, Forchheim, Germany) equipped with 2-kW/channel amplifiers. Informed consent was given before the scanning session (study approved by the Comité d'éthique de la recherche du Regroupement Neuroimagerie Québec for MNI and the Mass General Brigham institutional review board for MGH).

2.1.2 | Sequences

Three scans were first acquired using CP mode to map the combined B_1^+ field and enable the calculation of RF shim weights using the *Shimming Toolbox*: (1) The magnitude and phase of each Tx channel was mapped using a vendor-supplied TurboFLASH sequence (termed “tfl_rfmap” by the vendor) with a presaturation pulse, excitation flip angle (FA)=10°, TR=10 370 ms, TE=2.42 ms, 28 sagittal slices of 3-mm thickness, 20% slice gap, 2×2 mm² in-plane resolution, FOV=240×388×84 mm³, covering the SC from at least the top of C1 vertebra to the bottom of T3 vertebra. (2) A separate TurboFLASH scan (termed “tfl_b1map”) was acquired to map the combined RF field, using the same orientation, matrix size, resolution, and presaturation pulse as the previous sequence. (3) A 2D GRE scan with TR=600 ms, TE=2.86 ms, FA=15°, 70 axial slices of 3-mm thickness, 0.6×0.6 mm² in-plane resolution, FOV=230×225.4×210 mm³ was acquired to image cervical anatomy, perform SC segmentation, and compare

SC versus CSF contrast along the cervical spine across RF shimming methods. For all scans, B_0 shim coefficients were calculated on a rectangular volume around the SC, spanning between the intervertebral disks C2/C3 down to T2/T3, and kept constant throughout the session. RF shim weights were derived from the *tfl_rfmap* and GRE scans as described in Section 2.1.3.

Following RF shim calculation, the *tfl_b1map* scan and GRE scan were re-acquired for six shim conditions. For “Vendor: Volume Specific” (referred to as “volume” in the manuscript), the B_0 shim volume was defined as the region of interest (ROI) for RF shimming, and the shim weights were calculated by the scanner using a vendor-supplied approach. In “Vendor: Patient Specific” (referred to as “patient”), shim weights were likewise calculated by the scanner, but the ROI was defined as the whole FOV. Both vendor-supplied approaches aimed to minimize the coefficient of variation (CoV) within their respective ROIs. For each participant, the reference voltage was kept constant during all scans. For sub-04, the reference voltage of GRE scans was clipped at 400 V to minimize the occurrence of PALI errors.

2.1.3 | RF shim weight calculation

RF shim weights were calculated from the CP acquisitions using the following procedure. Image data were transferred to a separate laptop. The SC was segmented from the CP mode GRE scan using the *Spinal Cord Toolbox* (SCT).²⁸ The center of mass of that segmentation was then used to create a cylindrical mask of 28-mm diameter, within which RF shimming coefficients were estimated. The choice of the mask size was driven empirically, based on preliminary experiments looking at (i) the robustness of RF shimming coefficient estimation with respect to the number of voxels inside a mask, and (ii) the amount of subject motion within an approximate 1-h session. The mask was subsequently cropped between intervertebral disks C2/C3 and T2/T3 and then warped to the space of the *tfl_rfmap* scan. Four RF shimming algorithms were implemented in the *Shimming Toolbox*. The output of each algorithm is a complex set of Tx channel weights (w) that minimize the following expressions:

To optimize with phase-only shimming (referred to as “phase”), the coefficient of variation (CV) of the combined B_1^+ field was minimized within the ROI by adjusting the phase of each Tx channel, without taking the magnitude into account, as follows:

$$w_{\text{phase-only}} = \arg \min_w \left\{ \frac{\sigma(|B_1^+ \text{phase-only}(w)|)}{\mu(|B_1^+ \text{phase-only}(w)|)} \right\} \quad (1)$$

where

$$B_1^+ \text{phase-only}(w) = \sum_n \frac{1}{\sqrt{N_{\text{Tx}}}} w_n B_{1n}^+ \quad (2)$$

and $w_n = e^{i\phi_n}$

where B_{1n}^+ is the measured B_1^+ map of Tx channel n , and $\sigma(|B_1^+ \text{phase-only}(w)|)$ and $\mu(|B_1^+ \text{phase-only}(w)|)$ are the SD and the mean of the shimmed $B_1^+ \text{phase-only}(w)$ map, respectively. Crucially, only the phase of the B_1^+ maps is adjusted (via $w_n = e^{i\phi_n}$), whereas the amplitude is left unmodified.

To optimize for CV reduction across the cord (referred to as “CoV”), the CV of the combined B_1^+ field was minimized by adjusting both the phase and magnitude of each Tx channel as follows:

$$w_{\text{CV red}} = \arg \min_w \left\{ \frac{\sigma(|B_1^+ \text{CV red}(w)|)}{\mu(|B_1^+ \text{CV red}(w)|)} \right\} \quad (3)$$

where

$$B_1^+ \text{CV red}(w) = \sum_n \frac{1}{\text{norm}(w)} w_n B_{1n}^+ \quad (4)$$

and $w_n = A_n e^{i\phi_n}$. Thus, $w_{\text{CV red}}$ weights are calculated to minimize the CV by taking both the phase (via $e^{i\phi_n}$) and the magnitude (via A_n) of each transmit channel into account.

To optimize toward a target 20 nT/V in the ROI (referred to as “target”), a target B_1^+ amplitude (“ t ”) of 20 nT/V was used when determining the shim weights. 20 nT/V was chosen based on the performance of the coil as described in Lopez-Rios et al.²⁵:

$$w_{\text{target}} = \arg \min_w \left\{ \sum_n^{N_{\text{voxels}}} (|B_1^+(w)| - t)^2 \right\}. \quad (5)$$

To optimize for SAR efficiency (referred to as “SAR-eff”), the local SAR, calculated from virtual observation points, was considered to increase the mean B_1^+ per the square root of the maximum 10g-averaged local SAR, as follows:

$$w_{\text{SAR eff}} = \arg \min_w \{ \text{SAR}_{\text{eff}}(w) \} \quad (6)$$

$$\text{SAR}_{\text{eff}}(w) = \frac{\mu(|B_1^+(w)|)}{\sqrt{\max(\text{SAR}_{\text{local}}(w))}}, \quad (7)$$

where $w_{\text{SAR eff}}$ and w_{target} optimized both the magnitude and phase of each transmit channel.

After calculation, the corresponding weights were transferred back to the scanner and applied to the corresponding Tx channels. Figure 1 illustrates the workflow for calculating RF shim weights. Total time for the RF shim

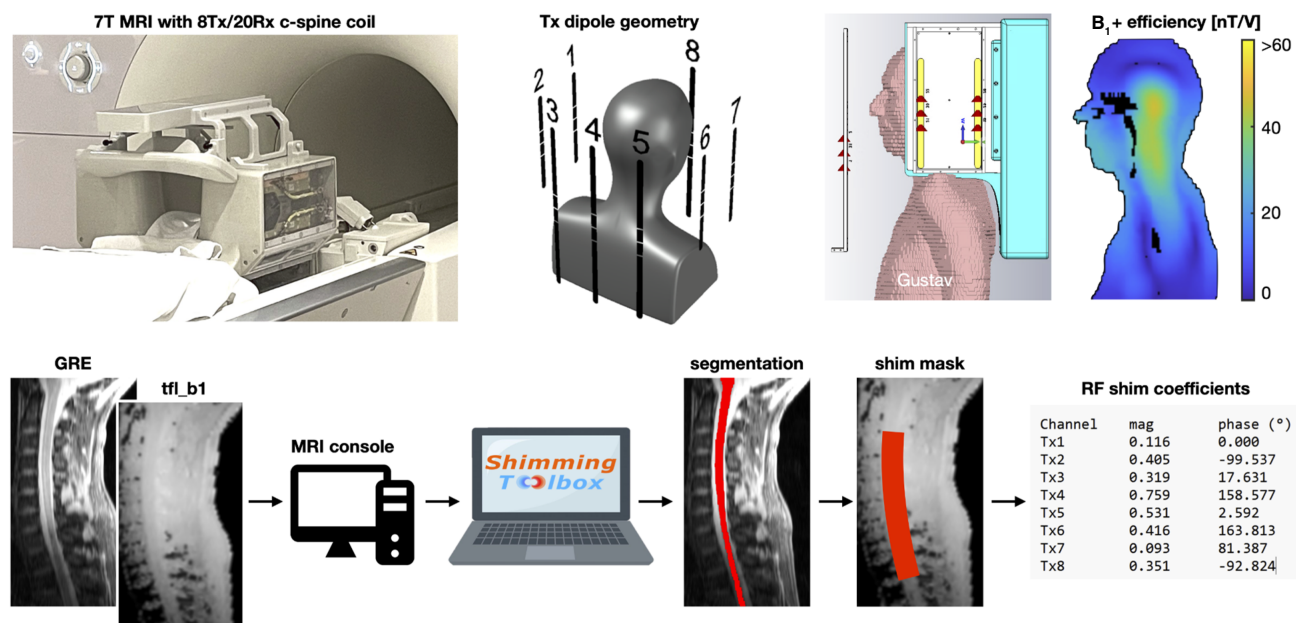


FIGURE 1 Overview of the RF shimming procedure. The top panel shows the RF coil used for the experiments, alongside the Tx coil geometry and the electromagnetic simulation results (performed on the Gustav model using *CST Studio Suite*) yielding the CP mode used for this coil. The bottom panel shows the RF shimming procedure (with approximate duration). First, gradient echo (GRE) and tfl_rfmap scans are acquired (4 min 30 s). Second, these images are transferred via ethernet socket from the MRI console onto a separate laptop running the *Shimming Toolbox* and *Spinal Cord Toolbox* (< 1 s). Third, the spinal cord is automatically segmented to produce a mask that is resampled into the space of the individual coil magnitude and phase images of the tfl_rfmap scan (~5 s). Fourth, the RF shim weights are calculated according to the defined constraints for each shim scenario (1 min total).

weight calculation, including data transfer, was approximately 1 min.

All processing was performed in *Python* using a reproducible Jupyter Notebook (see the Data Availability Statement). The processing pipeline was applied to each subject's data and took about 2 h to run on *Google Colab* (without paid license; 2 Xeon CPU 2.20GHz and 12GB RAM). The key processing steps are described subsequently.

2.1.4 | Anatomical T_2^* -weighted GRE

Automated segmentation of the SC was performed with SCT's *sct_deepseg_sc*²⁹ on the T_2^* -weighted images acquired with the "CoV" shim mode. Vertebral levels were labeled using existing manual disc labels. Automated disc labeling methods were not robust enough due to the thick axial slices that caused poor appearance of the intervertebral discs. A mask of the CSF was generated semi-automatically: a first pass using *sct_propseg* (with the flag "-CSF"),³⁰ followed by manual correction. The images from the other shim modes were registered to the "CoV" one using a deep learning algorithm.³¹ After reviewing all processes using SCT's quality control module,¹ the magnitude signal of the CSF and the spinal cord was

extracted slice-wise between C3 and T2 levels, smoothed (convolution with uniform filter of 50 pixels), and their ratio was computed to assess the consistency of the FA across different shim methods.

To create plots displaying participants with consistent vertebral levels along the x-axis, the x-tick positions were linearly interpolated. This interpolation was done to align with the evenly distributed distances of vertebral levels as defined by the PAM50 template.³²

2.1.5 | B_1^+ field mapping

Each tfl_b1map magnitude image (e.g., "CP," "CoV") was registered to the "CoV" GRE scan using the *SliceReg* algorithm.³³ The computed warping field was applied to bring the GRE's segmentation and vertebral levels into each respective tfl_b1map space. The tfl_b1map relative FA map was converted into normalized units (nT/V). The B_1^+ map values were extracted specifically within the segmented spinal cord. After reviewing all processes using SCT's quality control module, B_1^+ values were extracted slice-wise between C3 and T2 levels and smoothed (convolution with uniform filter of 20 pixels). For each RF shimming scenario, B_1^+ inhomogeneity was calculated as the CV of B_1^+ values between C3 and T2.

2.2 | Statistics

A one-way repeated-measures ANOVA was run to assess differences between shim methods (each being acquired on the same subjects, hence the repeated ANOVA). Post hoc analyses were performed with paired *t*-tests for each pair of shim methods and corrected for multiple comparisons with the Benjamini-Hochberg false discovery rate method.

3 | RESULTS

Figure 2 shows a comparison of B_1^+ efficiency maps across the tested RF shimming scenarios in 1 representative subject. Both the vendor and the custom RF shimming algorithms result in a markedly different B_1^+ efficiency along the ROI compared with CP mode. In particular, vendor algorithms evaluated in this subject appear to favor CV at the expense of B_1^+ efficiency.

Figure 3 shows the GRE magnitude images across the tested RF shimming scenarios in 1 representative subject. When interpreting these images in concomitance with the B_1^+ maps from Figure 2, we notice the consistency

between a higher B_1^+ efficiency and a higher magnitude signal.

Both the vendor and custom RF shimming algorithms result in a markedly different B_1^+ efficiency along the ROI compared with CP mode. In particular, vendor algorithms evaluated in this subject appear to favor CV at the expense of B_1^+ efficiency. As described in Table 1, across all subjects, “patient” mode resulted in the lowest B_1^+ homogeneity, whereas “volume” resulted in the highest homogeneity, and “CoV”, “phase”, “target”, and “SAREff” shim modes showed higher B_1^+ homogeneity than “CP” mode, with both vendor-supplied methods showing low B_1^+ efficiency.

The B_1^+ efficiency measured in the SC from C3 to T2 vertebral levels across participants is shown in Figure 4A. Overall, we observe the same trend across participants, with the “CP”, “CoV”, and “SAREff” shim modes showing the highest B_1^+ efficiency, and the “patient” and “volume” modes showing the lowest B_1^+ efficiency. Aggregate statistics are described in Table 1 (column “ B_1^+ efficiency”) and confirm the observed trends. Results of ANOVA and post hoc analyses are available as Table S1. An intriguing observation is that for sub-02, and to a lesser extent in sub-04, the “target” algorithm failed to achieve its goal

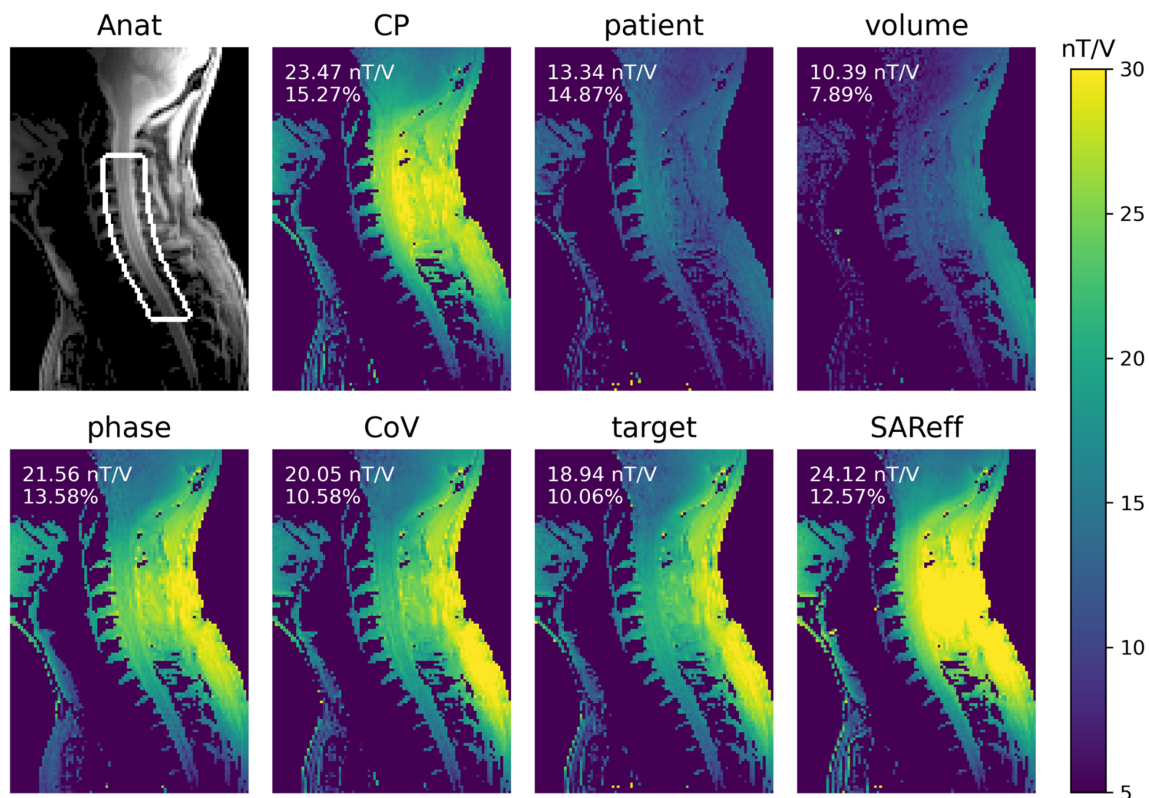


FIGURE 2 B_1^+ efficiency for one participant (sub-05) across all seven RF shimming conditions. The top left panel shows the tfl_b1map magnitude image with an overlay of the mask that was used to perform RF shimming. Text inserts show the mean (in nT/V) and CV (in %) of B_1^+ efficiency along the spinal cord between C3 and T2.

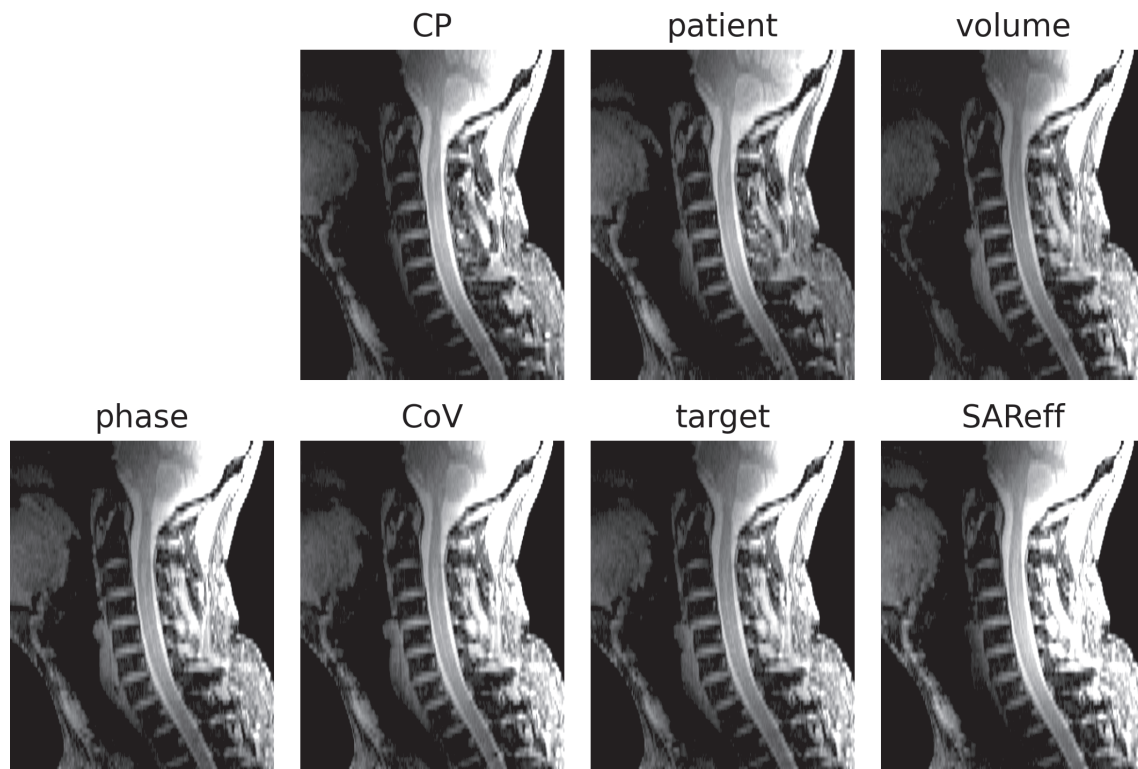


FIGURE 3 Gradient echo (GRE) magnitude signal intensity for one participant (sub-05) across RF shimming conditions. The colormap adjustment is the same across all images. The differences in B_1^+ efficiency along the cord between the various RF shimming conditions results in signal intensity differences. Visual contrast between CSF and spinal cord (SC) is affected, with “volume” showing poor visual contrast compared with, for example, the “CP” or “SAREff” modes.

TABLE 1 Summary statistics of B_1^+ homogeneity (lower is better), B_1^+ efficiency (higher is better), and CSF/cord contrast (higher is better) in the spinal cord between C3 and T2 vertebral levels (included).

Shim mode	B_1^+ homogeneity [%]	B_1^+ efficiency [nT/V]		CSF/Cord contrast	
	$100 \times (\text{SD}/\text{Mean})$	Mean	SD	Mean	SD
CP	17.69 ± 4.28	23.09 ± 4.17	3.90 ± 0.90	1.20 ± 0.04	0.03 ± 0.02
Patient	20.72 ± 9.76	12.62 ± 2.93	2.26 ± 0.32	1.23 ± 0.04	0.03 ± 0.02
Volume	9.08 ± 1.54	12.18 ± 2.84	1.40 ± 0.43	1.25 ± 0.05	0.02 ± 0.01
Phase	12.11 ± 1.15	20.22 ± 3.83	2.33 ± 0.40	1.24 ± 0.04	0.01 ± 0.01
CoV	11.32 ± 1.08	20.25 ± 6.78	2.18 ± 0.67	1.27 ± 0.03	0.01 ± 0.01
Target	11.17 ± 1.52	17.63 ± 3.16	1.92 ± 0.32	1.24 ± 0.02	0.02 ± 0.01
SAREff	16.02 ± 3.40	23.28 ± 5.53	3.36 ± 0.41	1.19 ± 0.05	0.02 ± 0.02

Note: The mean and SD, respectively, correspond to the average and standard deviation computed along the spinal cord of each participant. Each cell represents the mean $\pm$ standard deviation across participants. The best values are highlighted in bold.

of a 20 nT/V average. Conversely, the “SAREff” algorithm appeared to come closer to this mean value. Looking at the online output from *Shimming Toolbox*, for sub-02 and sub-04, the mean B_1^+ within the SC were, respectively, 19.3 nT and 19.0 nT/V, which is close to the target. However, the RFshim ROI was based on a noisy and aggressively masked *tfl_rfmap* sequence, whereas the evaluation ROI was based

on the *tfl_b1map*, which could explain the discrepancies between the target versus measured B_1^+ values.

Figure 4B shows the CSF-to-SC ratio measured from the GRE scan in the SC from C3 to T2 vertebral levels across participants. Overall, we observe the same trend across participants, with the “CoV”, “volume”, and “target” shim modes showing the highest CSF/SC contrast,

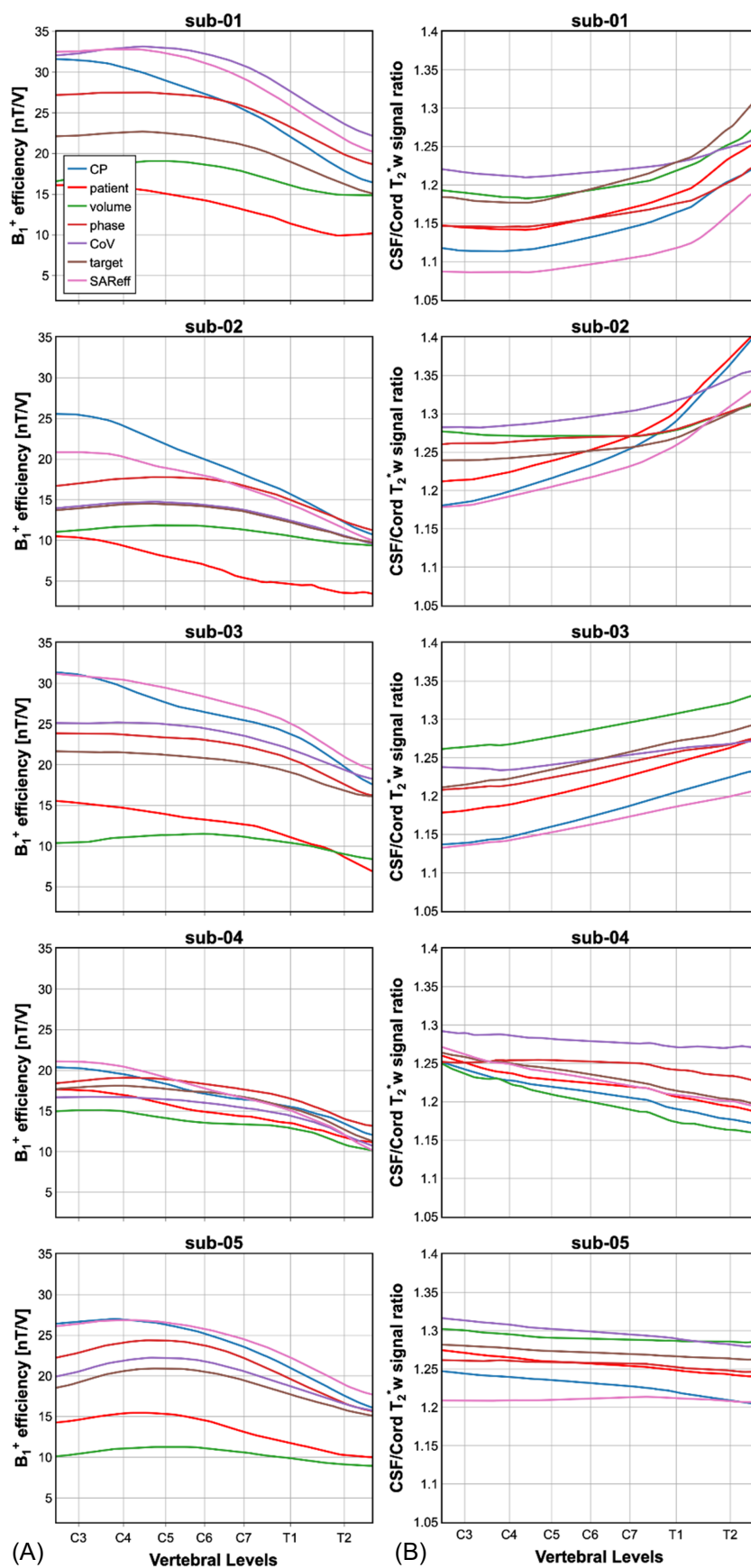


FIGURE 4 B_1^+ efficiency (A) and CSF/cord signal ratio from the gradient-echo scan (B) across subjects and across different RF shimming conditions. Data were measured in the spinal cord from C3 to T2 vertebral levels. To match the x-ticks across subjects, the C2-C3 and the T2-T3 intervertebral discs of each subject were aligned with that of the PAM50 template,³² and the curves were scaled along the x-axis to match this alignment.

and the “SAReff” and “CP” modes showing the lowest contrast. Aggregate statistics are described in Table 1 (column “CSF/cord contrast”) and confirm the observed trends. Results of ANOVA and post hoc analysis are available as Table S2.

4 | DISCUSSION AND CONCLUSIONS

To make full use of the benefits of 7 T MRI for SC imaging, the inherent drawback of B_1^+ field inhomogeneity needs to be overcome. In this work, we presented the feasibility and utility of using static RF shimming for SC imaging across two 7 T systems. Several vendor-based and automatic ROI-customized shim algorithms were evaluated.

4.1 | Evaluation metrics

Before delving into method comparisons, it is essential to first outline the evaluation metrics used. The first metric, Tx homogeneity, reflects the variability of the B_1^+ field within the target imaging region. Achieving uniform FAs across this ROI is crucial, notably to ensure a consistent contrast throughout the ROI.

Generally, enhancing B_1^+ homogeneity is possible with increased input power, although hardware limitations, such as the maximum capacities of the RF amplifier for peak and average power, cap this improvement. This leads to our second metric, Tx efficiency, which gauges the efficacy of the RF field produced in the specified region relative to the input power.

The third metric involves the CSF/cord contrast derived from a GRE magnitude image. This is a particularly pragmatic metric, because the contrast between CSF and the SC is crucial for subsequent image analyses and interpretations, such as the precise segmentation of the SC to measure indicators of conditions like cord atrophy. Ideally, if both tissues are subjected to the same FA, the contrast should remain constant along the superior–inferior (S-I) direction. Thus, examining variations in CSF/cord contrast along the S-I direction offers an indirect measure of B_1^+ homogeneity. Although the average CSF/cord contrast along the S-I axis is also significant, it is influenced by the T_1 relaxation properties of the tissues and the FA that optimizes contrast. Consequently, if the GRE scans were conducted with a higher or lower FA, the average contrast for each evaluated shimming method could have varied. Because the optimal FA for maximizing contrast was not established in this study, conclusions cannot be drawn about the shimming method that maximizes contrast. Instead, evaluations should concentrate on the

variations in contrast along the S-I axis, as this serves as an indicator of B_1^+ homogeneity.

4.2 | Comparison of RF shimming methods

Vendor-based approaches (“patient,” “volume”) were restricted by the possibility to optimize the ROI within which RF shim coefficients are calculated. In the “patient” approach, the optimization ROI is the full FOV, which leads to an inhomogeneous excitation within the cord (Table 1), whereas in the “volume” approach, a rectangular ROI can be specified and resulted in the best B_1^+ homogeneity across all methods. However, a rectangle is an imperfect fit for the anatomy of the spinal cord. The cervical SC is naturally curved and surrounded by other tissues and the trachea. Within this study, we found it challenging to draw a rectangle that included all of the SC but did not include any air or surrounding tissue, thereby biasing the shimming algorithm. The comparison of RF shim performance between a rectangular ROI and an ROI that tightly follows the curvature of the spine is illustrated on Figure S1. Another drawback of vendor solutions is the opaque nature of the shimming algorithm and its cost function. We observed that the “volume” approach tended to improve B_1^+ homogeneity at the cost of B_1^+ efficiency, whereas the “patient” approach performed poorly for both B_1^+ homogeneity and efficiency.

Custom RF shimming methods (“CoV”, “phase”, “target”, and “SAReff”) implemented in the vendor-agnostic *Shimming Toolbox* can overcome these limitations. The “CoV” method produced the highest CSF/SC contrast on T_2^* -weighted scans, and the lowest variation of that contrast along the S-I axis. Using the *Spinal Cord Toolbox*, the SC can be segmented to derive an ROI that tightly fits the cord area for an optimal calculation of RF shimming coefficients. Previous studies undertook a similar segmentation based approach for optimizing B_0 shimming in the SC.^{27,34} The optimization algorithms and their cost functions implemented in *Shimming Toolbox* are fully open source and transparent to the user and could be extended or modified at will.

We observed that changes in B_1^+ efficiency and homogeneity result in similar changes in visual image quality, with low-efficiency RF shim methods (“volume”) resulting in less apparent signal, whereas high-efficiency methods (“SAReff”, “CoV”) result in more apparent signal in GRE scans. B_1^+ homogeneity influenced the variation of CSF/SC contrast, thus the “CoV”, due to its high efficiency and homogeneity, the highest contrast showed the least variation.

Patient motion, particularly motion between the scans used to map the B_1^+ field and the acquisition of RF shimmed scans, was a potential source of errors in our study. Although patient motion was minimized using careful instruction and immobilization during our study, it was not actively mitigated. To minimize the potential impact, a large, 28-mm-diameter mask was chosen for RF shimming. Post-scan visual inspection of the GRE scans revealed that the SC was within this mask for all RF shimming scenarios and all participants.

The performance of the “CP” mode was similar to that of the patient-specific “SAReff” method, which is likely because the “CP” mode of both coils was calculated to maximize B_1^+ efficiency. However, what we did find surprising, is how good the “CP” mode was across all subjects (for both sites), despite it being fixed across subjects. We would have expected better performance for the “SAReff” model, as it provides personalized shim solutions for each individual. This suggests interesting follow-up research hypotheses: Are personalized shim solutions not as good in practice because of the difficulty in obtaining robust and reliable B_1^+ maps? Is it because of excessive noise and/or subject motion? Given how good performances are on “CP” mode across subjects, does that suggest that shim solutions based on universal pulses would be a promising solution for spinal cord applications at 7 T (see also the following discussion)?

4.3 | Differences between sites

We noticed some differences between sites. Notably, the CSF/cord contrast (Figure 4B) curves are fairly similar within sites (MGH, sub-01, sub-02, and sub-03; MNI, sub-04 and sub-05), but the trends are different between the two sites. More specifically, for MGH subjects, the contrast is increasing toward the lower vertebral levels, whereas the MNI subjects show the opposite trends. Despite the two sites being equipped with a coil that followed the same design and construction protocol, especially in terms of Tx dipole sizes and locations, the two coils are not exactly the same. Essentially, the MNI coil housing has two layers of dielectric materials (total thickness: 6.2 mm) between the Tx and Rx arrays, which were removed from the MGH design. However, the measured S-parameters and the simulated electromagnetic fields are very similar between both coils.

Other possible sources of discrepancies between the two sites include the subjects' body mass index (sub-04 and sub-05 are larger), positioning (sub-04 and sub-05 show exhibit more lordosis), and breathing (sub-04 and sub-05 exhibit more respiratory-related artifacts), which could

have affected the reliability of the B_1^+ maps, especially in the upper thoracic levels.

4.4 | Limitations and future studies

In this study, we used a configuration of eight transmit dipoles positioned in a single row around the neck area. Research suggests that a design incorporating multiple rows of transmit elements, similar to those used in earlier C-spine coil designs,¹⁵ has the potential to significantly enhance RF shimming due to the better complementarity of coil elements along the S-I axis.^{35,36} Expanding coverage areas, such as the head and neck, would largely benefit from a 16-channel Tx system, as shown by May et al.³⁷ However, such a system would incur higher costs.

Evaluation of the different shim approaches in terms of SAR was not part of our study, and SAR information was not collected at MGH. For comparison, we show the SAR of the GRE scans for our representative subject (sub-05) in Table S3. We observe that “volume” resulted in the lowest head-only 10-s averaged SAR, whereas “patient” resulted in the second-highest SAR, following “SAReff.” Further studies may investigate the effects of RF shimming on SAR, based on electromagnetic simulations and scanner side measurements.

The current workflow of the *Shimming Toolbox* would benefit from better interoperability with vendor-specific consoles, especially to reduce time required for data transfer. Nonetheless, the proprietary nature of vendor software poses a challenge to achieving transparent and cross-platform reproducibility. In this context, the development of vendor-neutral acquisition platforms^{38,39} offers a promising solution by facilitating a more open and interoperable environment for research and clinical applications.

Another consideration is the choice of B_1^+ mapping sequence. Here, we used the TurboFLASH sequence, which offers fast acquisition times at the cost of sensitivity to motion, susceptibility artifacts, and a limited dynamic range for B_1^+ field strength measurement, affecting its accuracy and reliability in certain scenarios. Alternative sequences can be explored and compared in future studies for SC applications, such as the dual excitation angle method (DREAM),⁴⁰ the actual flip-angle imaging (AFI).⁴¹

Slice-wise RF shimming would also be an interesting avenue for the SC, as demonstrated in the brain,²⁰ especially for sequences acquired axially with large slice thicknesses, as is often the case in quantitative MRI SC protocols.

Our study focused on tailoring the B_1^+ field to the individual participants to achieve specific B_1^+ constraints (homogeneity, target B_1^+ value, minimum SAR, etc.). A complementary approach can be found in Universal

Pulses,⁴² where instead of participant-specific tailoring, pulses are designed to create homogeneous excitation across a wide range of participants. Universal Pulses have been demonstrated for imaging the brain,⁴² heart,⁴³ and SC.^{44,45}

In conclusion, this study compared various RF shimming methods for SC imaging at 7 T, across two different sites. The “CP”, “CoV”, and “SAReff” shim modes showed the highest B_1^+ efficiency, and the vendor-based “patient” and “volume” modes showed the lowest B_1^+ efficiency. The “CoV” method produced the highest CSF/cord contrast on T_2^* -weighted scans. The study’s findings highlight the potential of RF shimming to advance 7T MRI’s clinical utility for central nervous system imaging by enabling more homogenous and efficient SC imaging. Future studies may explore the feasibility of applying static RF shimming to other UHF MRI that are adversely affected by signal inhomogeneity, such as functional MRI.⁴⁶

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CONFLICT OF INTEREST

Dr. Papp is an employee of Siemens Healthcare AB, Sweden, as of September 2023, but was a postdoc at Polytechnique Montreal at the time of the study and the initial preparation of the manuscript. His contributions were made outside of working hours and did not use any company resources for this work. Marcus Couch is an employee of Siemens Healthcare. His contribution was to help us implement some of the RF shimming techniques and to provide proprietary information. As of January 2024, Dr. Barry is employed by the National Institute of Biomedical Imaging and Bioengineering at the National Institutes of Health. This does not alter our adherence to Magnetic Resonance in Medicine policies on sharing data and materials. We have no other conflicts of interest to disclose.

DATA AVAILABILITY STATEMENT

The MRI data used in this study are available on OpenNeuro (<https://openneuro.org/datasets/ds004906/versions/2.4.0>). The data were organized following the BIDS convention for quantitative MRI.^{47,48} Briefly, the data-set structure consists of multiple subdirectories, each corresponding to a different subject (e.g., sub-01, sub-02). Within each subject’s directory, the data are further organized into anatomical (anat) and B_1^+ field mapping (fmap) scans. The anatomical scans include various T_1 -weighted and T_2^* -weighted images, whereas the field-mapping scans contain TB1 TurboFLASH and other types of images. Each scan type is saved in NIfTI format (.nii.gz). All figures from Section 3 can be reproduced directly from the data with the Jupyter notebook available at <https://github.com/shimming-toolbox/rf>

-shimming-7t/tree/r20240528, which has been published as a NeuroLibre reproducible preprint.⁴⁹

ENDNOTE

¹ https://spinalcordtoolbox.com/user_section/command-line.html#sct-qc.

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REFERENCES

- Kearney H, Miller DH, Ciccarelli O. Spinal cord MRI in multiple sclerosis—diagnostic, prognostic and clinical value. *Nat Rev Neurol*. 2015;11:327-338.
- David G, Mohammadi S, Martin AR, et al. Traumatic and non-traumatic spinal cord injury: pathological insights from neuroimaging. *Nat Rev Neurol*. 2019;15:718-731.
- Barry RL, Vannesjo SJ, By S, Gore JC, Smith SA. Spinal cord MRI at 7T. *Neuroimage*. 2018;168:437-451.
- Sigmund EE, Suero GA, Hu C, et al. High-resolution human cervical spinal cord imaging at 7 T. *NMR Biomed*. 2012;25:891-899.
- Massire A, Taso M, Besson P, Guye M, Ranjeva JP, Callot V. High-resolution multi-parametric quantitative magnetic resonance imaging of the human cervical spinal cord at 7T. *Neuroimage*. 2016;143:58-69.
- Ouellette R, Treaba CA, Granberg T, et al. 7 T imaging reveals a gradient in spinal cord lesion distribution in multiple sclerosis. *Brain*. 2020;143:2973-2987.
- Testud B, Fabiani N, Demortière S, et al. Contribution of the MP2RAGE 7T sequence in MS lesions of the cervical spinal cord. *AJNR Am J Neuroradiol*. 2023;44:1101-1107.
- Trattnig S, Springer E, Bogner W, et al. Key clinical benefits of neuroimaging at 7T. *Neuroimage*. 2018;168:477-489.
- Kreiter DJ, van den Hurk J, Wiggins CJ, Hupperts RMM, Gerlach OHH. Ultra-high field spinal cord MRI in multiple sclerosis: where are we standing? A literature review. *Mult Scler Relat Disord*. 2021;57:103436.
- Schick F. Whole-body MRI at high field: technical limits and clinical potential. *Eur Radiol*. 2005;15:946-959.
- Ibrahim TS, Lee R, Abduljalil AM, Baertlein BA, Robitaille PML. Dielectric resonances and B₁ field inhomogeneity in UHF MRI: computational analysis and experimental findings. *Magn Reson Imaging*. 2001;19:219-226. doi:10.1016/S0730-725X(01)00300-9
- Collins CM, Liu W, Schreiber W, Yang QX, Smith MB. Central brightening due to constructive interference with, without, and despite dielectric resonance. *J Magn Reson Imaging*. 2005;21:192-196.
- Röschmann P. Radiofrequency penetration and absorption in the human body: limitations to high-field whole-body nuclear magnetic resonance imaging. *Med Phys*. 1987;14:922-931.
- Yang QX, Wang J, Zhang X, et al. Analysis of wave behavior in lossy dielectric samples at high field. *Magn Reson Med*. 2002;47:982-989.
- Zhao W, Cohen-Adad J, Polimeni JR, et al. Nineteen-channel receive array and four-channel transmit array coil for cervical spinal cord imaging at 7T. *Magn Reson Med*. 2014;72:291-300.
- Pfaffenrot V, Brunheim S, Rietsch SHG, et al. An 8/15-channel Tx/Rx head neck RF coil combination with region-specific B₁⁺ shimming for whole-brain MRI focused on the cerebellum at 7T. *Magn Reson Med*. 2018;80:1252-1265.
- Van Den Bergen B, Van den Berg CAT, Bartels LW, Lagendijk JJW. 7 T body MRI: B1 shimming with simultaneous SAR reduction. *Phys Med Biol*. 2007;52:5429-5441.
- Metzger GJ, Snyder C, Akgun C, Vaughan T, Ugurbil K, Van de Moortele PF. Local B₁⁺ shimming for prostate imaging with transceiver arrays at 7T based on subject-dependent transmit phase measurements. *Magn Reson Med*. 2008;59:396-409.
- Orzada S, Maderwald S, Kraff O, et al. 16-channel Tx/Rx body coil for RF shimming with selected cp modes at 7T. *Proceedings of the 18th Annual Meeting of ISMRM*. Vol 50; 2010. Abstract 0050.
- Curtis AT, Gilbert KM, Klassen LM, Gati JS, Menon RS. Slice-by-slice B₁⁺ shimming at 7 T. *Magn Reson Med*. 2012;68:1109-1116.
- Kraff O, Bitz AK, Kruszona S, et al. An eight-channel phased array RF coil for spine MR imaging at 7 T. *Invest Radiol*. 2009;44:734-740.
- Wu B, Wang C, Krug R, et al. 7T human spine imaging arrays with adjustable inductive decoupling. *IEEE Trans Biomed Eng*. 2010;57:397-403.
- Rietsch SHG, Brunheim S, Orzada S, et al. Development and evaluation of a 16-channel receive-only RF coil to improve 7T ultra-high field body MRI with focus on the spine. *Magn Reson Med*. 2019;82:796-810.
- May MW, Hansen SLJD, Kutscha N, et al. A patient-friendly 16ch Tx/64ch Rx array for combined head and neck imaging at 7 Tesla. *Proceedings of the 21st Annual Meeting of ISMRM [Virtual]*. Vol 181; 2021. Abstract 0181.
- Lopez-Rios N, Gilbert KM, Papp D, et al. An 8-channel Tx dipole and 20-channel Rx loop coil array for MRI of the cervical spinal cord at 7 Tesla. *NMR Biomed*. 2023;36:e5002.
- Destrue A, Jomin P, Wichmann T, Guye M, Abdeddaim R, Callot V. B₁ shimming for cervical spine 7T parallel transmission MRI: preliminary in vivo imaging and preparation of virtual observation points. *Proceedings of the 30th Annual Meeting of ISMRM*. Vol 1463; 2022. Abstract 1463.
- D'Astous A, Cereza G, Papp D, et al. Shimming toolbox: an open-source software toolbox for B0 and B1 shimming in MRI. *Magn Reson Med*. 2022;89:1401-1417. doi:10.1002/mrm.29528

28. de Leener B, Lévy S, Dupont SM, et al. SCT: spinal cord toolbox, an open-source software for processing spinal cord MRI data. *Neuroimage*. 2017;145:24-43.
29. Gros C, de Leener B, Badji A, et al. Automatic segmentation of the spinal cord and intramedullary multiple sclerosis lesions with convolutional neural networks. *Neuroimage*. 2019;184:901-915.
30. De Leener B, Cohen-Adad J, Kadoury S. Automatic segmentation of the spinal cord and spinal canal coupled with vertebral labeling. *IEEE Trans Med Imaging*. 2015;34:1705-1718.
31. Beal E, Cohen-Adad J. Contrast-agnostic deep learning-based registration pipeline: validation in spinal cord multimodal MRI data. *Aperture Neuro*. 2023;2023:1-21.
32. De Leener B, Fonov VS, Collins DL, Callot V, Stikov N, Cohen-Adad J. PAM50: unbiased multimodal template of the brainstem and spinal cord aligned with the ICBM152 space. *Neuroimage*. 2018;165:170-179.
33. Cohen-Adad J, Levy S, Avants B. Slice-by-slice regularized registration for spinal cord MRI: SliceReg. In: *Proceedings of the 23th Annual Meeting of ISMRM, Toronto, Canada*. Vol 4428; 2015. Abstract 4428.
34. Kaptan M, Vannesjo SJ, Mildner T, et al. Automated slice-specific z-shimming for functional magnetic resonance imaging of the human spinal cord. *Hum Brain Mapp*. 2022;43:5389-5407.
35. Karadeniz N, Hajnal J, Ipek Ö. Comparison of the transmit performance and local SAR for multi-row parallel transmit coil arrays at 3 tesla human brain MRI. *Proceedings of the 31st Annual Meeting of ISMRM*. Vol 4405; 2023.
36. Nikulin AV, Bosch D, Solomakha GA, Glang F, Scheffler K, Avdievich NI. Double-row 16-element folded-end dipole transceiver array for 3D RF shimming of the whole human brain at 9.4 T. *NMR Biomed*. 2023;36:e4981.
37. May MW, Hansen SLJD, Mahmutovic M, et al. A patient-friendly 16-channel transmit/64-channel receive coil array for combined head-neck MRI at 7 tesla. *Magn Reson Med*. 2022;88:1419-1433. doi:10.1002/mrm.29288
38. Karakuzu A, Biswas L, Cohen-Adad J, Stikov N. Vendor-neutral sequences and fully transparent workflows improve inter-vendor reproducibility of quantitative MRI. *Magn Reson Med*. 2022;88:1212-1228.
39. Cordes C, Konstandin S, Porter D, Günther M. Portable and platform-independent MR pulse sequence programs. *Magn Reson Med*. 2020;83:1277-1290.
40. Nehrke K, Börner P. DREAM—a novel approach for robust, ultrafast, multislice B1 mapping. *Magn Reson Med*. 2012;68:1517-1526.
41. Yarnykh VL. Actual flip-angle imaging in the pulsed steady state: a method for rapid three-dimensional mapping of the transmitted radiofrequency field. *Magn Reson Med*. 2007;57:192-200.
42. Gras V, Vignaud A, Amadon A. Universal pulses: a new concept for calibration-free parallel transmission. *Magn Reson Med*. 2017;77:635-643. doi:10.1002/mrm.26148
43. Aigner CS, Dietrich S, Schaeter T, Schmitter S. Calibration-free pTx of the human heart at 7T via 3D universal pulses. *Magn Reson Med*. 2022;87:70-84.
44. Papp D, Boulant N, Massire A, Mauconduit F, Gras V, Cohen-Adad J. Universal pulses for the cervical spinal cord at 7T: a feasibility study. In: *Proceedings of the 31st ISMRM & ISMRT Annual Meeting & Exhibition*; 2023.
45. Aigner CS, Sánchez Alarcon MF, D'Astous A, Alonso-Ortiz E, Cohen-Adad J, Schmitter S. Calibration-free parallel transmission of the cervical, thoracic, and lumbar spinal cord at 7T. *Magn Reson Med*. 2024;92(4):1496-1510. Portico. doi:10.1002/mrm.30137
46. Barry RL, Smith SA, Dula AN, Gore JC. Resting state functional connectivity in the human spinal cord. *Elife*. 2014;3:e02812.
47. Gorgolewski KJ, Auer T, Calhoun VD, et al. The brain imaging data structure, a format for organizing and describing outputs of neuroimaging experiments. *Sci Data*. 2016;3:160044.
48. Karakuzu A, Appelhoff S, Auer T, et al. qMRI-BIDS: an extension to the brain imaging data structure for quantitative magnetic resonance imaging data. *Sci Data*. 2022;9:517.
49. Papp D, Gilbert KM, Cereza G, et al. Analysis code for the paper “RF shimming in the cervical spinal cord at 7T”. *NeuroLibre Reproducible Preprints*. 2024;25:1-7. doi:10.55458/neurolibre.00025

SUPPORTING INFORMATION

Additional supporting information may be found in the online version of the article at the publisher's website.

Table S1. Statistical results for B₁* mapping. Repeated ANOVA with post hoc paired *t*-test, with Benjamini-Hochberg false discovery rate (FDR) correction. Statistically significant differences ($p < 0.05$) are set in bold.

Table S2. Statistical results for CSF/spinal cord (SC) T2starw signal. Repeated ANOVA with post hoc paired *t*-test, with Benjamini-Hochberg false discovery rate (FDR) correction. Statistically significant differences ($p < 0.05$) are set in bold.

Table S3. Ten-second average head-only specific absorption rate (SAR) for gradient echo (GRE) scans acquired on sub-05 for all shim modes.

Figure S1. Effect of region of interest (ROI) size and shape for the performance of RF shimming. The left panel shows a rectangular ROI manually set on the vendor console, whereas the right panel shows an ROI that was automatically generated from the spinal cord segmentation and vertebral labeling. The B₁⁺ coefficient of variation (in %) is indicated for various RF shim methods.

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