

**RESEARCH ARTICLE**

**B_1** inhomogeneity correction of RARE MRI at low SNR: Quantitative in vivo \(^{19}\text{F}\) MRI of mouse neuroinflammation with a cryogenically-cooled transceive surface radiofrequency probe

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**Purpose:** Low SNR in fluorine-19 \(^{19}\text{F}\) MRI benefits from cryogenically-cooled transceive surface RF probes (CRPs), but strong \(B_1\) inhomogeneities hinder quantification. Rapid acquisition with refocused echoes (RARE) is an SNR-efficient method for MRI of neuroinflammation with perfluorinated compounds but lacks an analytical signal intensity equation to retrospectively correct \(B_1\) inhomogeneity. Here, a workflow was proposed and validated to correct and quantify \(^{19}\text{F}\)-MR signals from the inflamed mouse brain using a \(^{19}\text{F}\)-CRP.

**Methods:** In vivo \(^{19}\text{F}\)-MR images were acquired in a neuroinflammation mouse model with a quadrature \(^{19}\text{F}\)-CRP using an imaging setup including 3D-printed components to acquire co-localized anatomical and \(^{19}\text{F}\) images. **Model-based corrections** were validated on a uniform \(^{19}\text{F}\) phantom and in the neuroinflammatory model. Corrected \(^{19}\text{F}\)-MR images were benchmarked against reference images and overlaid on in vivo \(^{1}\text{H}\)-MR images. Computed concentration uncertainty maps using Monte Carlo simulations served as a measure of performance of the \(B_1\) corrections.

**Results:** Our study reports on the first quantitative in vivo \(^{19}\text{F}\)-MR images of an inflamed mouse brain using a \(^{19}\text{F}\)-CRP, including in vivo \(T_1\) calculations for \(^{19}\text{F}\)-nanoparticles during pathology and \(B_1\) corrections for \(^{19}\text{F}\)-signal quantification. **Model-based corrections** markedly improved \(^{19}\text{F}\)-signal quantification from errors > 50% to < 10% in a uniform phantom \((p < 0.001)\). Concentration uncertainty maps ex vivo and in vivo yielded uncertainties that were generally < 25%.
Fluorine-19 (19F) MRI shows promise in several biomedical applications. However, 19F-MRI suffers from low SNR due to the very limited availability of 19F nuclei in vivo. Consequently, 19F-MRI is constrained by long measurements. One SNR-boosting strategy has been to implement sensitivity-promoting surface RF coil technologies. Cryogenically cooled transceive surface RF probes (CRPs) have introduced a paradigm shift in preclinical imaging, providing substantial SNR gains compared with room-temperature RF coils. Further increases in SNR have been achieved with quadrature configurations, which provide a $\sqrt{2}$ SNR gain and better transversal B1 homogeneity compared with linear polarized RF coils.

Quadrature CRPs are typically single-tuned for X-nuclei, since dual-tunable capabilities would require electromagnetic decoupling between coil elements, degrading signal sensitivity. This adds extra post-processing challenges when locating quantified 19F signals in vivo. Moreover, the low-SNR and sparse nature of 19F prevents the MR system from performing reference power adjustments without an external 19F reference.

The greatest challenge of transceive surface RF probes like the 19F-CRP is their strong B1 inhomogeneities, which hamper T1 contrast and signal quantification, as the measured 19F signal depends on the number of 19F atoms per pixel, their distance from the probe surface, and relaxation times. Inhomogeneities in the excitation field ($B_1^+$) are typically corrected retrospectively using signal-intensity (SI) equations of corresponding RF pulse sequences. This is possible for gradient-echo or spin-echo techniques, but the SNR-efficient rapid acquisition with refocused echoes (RARE) technique lacks an exact SI equation.

We previously implemented three $B_1$ correction methods (model-based, hybrid, and sensitivity) for RARE 1H-MRI and transceive surface RF probes, considerably increasing image homogeneity and significantly reducing errors in signal quantification and T1 contrast.

The low SNR, signal sparsity, and lack of a priori location of the 19F signal constrain the reliability of signal quantification, even after $B_1$ correction. A procedure that evaluates the quality of the $B_1$ correction and quantification per image voxel is thus crucial. This is particularly relevant when monitoring and quantifying inflammation e.g., in the animal model of multiple sclerosis (MS), experimental autoimmune encephalomyelitis (EAE) using 19F-nanoparticles (NPs).

Here, we implemented and validated our $B_1$ correction approaches to correct 19F-MR images from a single-tuned quadrature 19F-CRP after estimating in vivo T1 of 19F-NPs in the EAE brain using a volume resonator. We performed Monte Carlo SNR simulations to estimate the associated concentration uncertainty. We also established a workflow using 3D-printed add-ons to facilitate in vivo localization of 19F-MR images from the 19F-CRP on anatomical images acquired from a 1H volume resonator. This workflow and correction method delivered the first quantitative in vivo 19F-MR images of an inflamed EAE mouse brain using a 19F-CRP. These results will be pivotal to drive future 19F research using transceive surface RF technologies to quantify inflammation or 19F-compounds in in vivo studies.

**METHODS**

**2.1 | Magnetic resonance hardware**

All experiments were carried out on a 9.4T small animal MR scanner (BioSpec 94/20; Bruker BioSpin, Ettlingen, Germany).

19F images were acquired using a 19F cryogenically-cooled transceive surface RF probe (19F-CRP CryoProbe; Bruker BioSpin) for mouse head imaging (inner diameter [ID] = 20 mm), composed of two elements operating in quadrature mode. Anatomical images were measured using a 72-mm (ID) linear volume resonator (Bruker BioSpin). T1 measurements of 19F-NPs in EAE brains and reference 19F images were acquired using a small-diameter (ID = 18.4 mm) mouse head 1H/19F volume resonator.
2.2 | Anatomical and $^{19}$F-MRI setup

Given the lacking $^1$H channel, an imaging setup including 3D-printed components was devised to acquire co-localized anatomical and $^{19}$F-CRP images.

2.2.1 | Animal bed modification

The standard animal bed uses a lever that elevates the bed, lifting the mouse head closer to the $^{19}$F-CRP. This feature hampers position reproducibility. To ensure spatial alignment of both $^{19}$F-CRP and anatomical images, a 3D-printed blocking component (Y-axis blocker) was designed to eliminate movement in the Y-axis (Figure 1A). Additionally, a new head holder was designed and 3D-printed to place the mouse head closer to the CRP surface (Figure 1A).

2.2.2 | $^1$H-MRI setup

The 72-mm-volume resonator was positioned around the center tube holding the $^{19}$F-CRP. Anatomical images were acquired after a CRP replica (dummy), inserted from the back of the scanner, was kept in place while the animal bed was inserted from the front.

![Image](image_url)

**FIGURE 1** Anatomical and fluorine-19 ($^{19}$F) imaging setup designed for a single-tuned cooled transceive surface RF probe (CRP). (A) Close-up view of the animal bed provided by the vendor with a custom-designed component that eliminates mobility in the Y-axis (y-axis blocker) and a new head holder to bring the animal’s head closer to the surface of the CRP. (B) Reference cap containing $^{19}$F-loaded nanoparticles (NPs) to perform $^{19}$F-CRP reference power adjustments and as reference for quantification. (C,D) $^1$H/$^{19}$F imaging setups. (E) For exemplary in vivo images, anatomical images and slice planning are performed using a 72-mm volume resonator and a CRP dummy. Afterward, reference power calibrations are carried out using the reference cap, and $^{19}$F images are acquired using the $^{19}$F-CRP.
2.2.3 | 19F-MRI setup

Both animal bed and dummy were removed and the 1H-volume resonator was retracted toward the back of the scanner. The 19F-CRP was mounted as instructed by the vendor.

A 19F-NP reference cap (section 2.3) was placed over the mouse head to perform 19F-CRP reference power adjustments and to acquire images for quantification (Figure 1B). Afterward, it was removed to acquire in vivo 19F images (Figure 1C–E).

2.3 | Sample and animal preparation

Table 1 summarizes all MR measurements, RF coils, and samples used.

Perfluoro-15-crown-5-ether (1200 mM PFCE; Fluorochem, Hadfield, United Kingdom; f ≈ 376.629 MHz) nanoparticles were prepared as described.27,28

To characterize 19F-CRP B1 fields, B1 maps and RARE images were used as follows:

- Low-T1 uniform phantom: 15-mL tube (ID = 14.6 mm, length = 120 mm, wall thickness = 0.8 mm; Fischer Scientific, Waltham, MA, USA) with 33.3% 2,2,2-trifluorethanol (Carl Roth & Co., Karlsruhe, Germany; f ≈ 376.633 MHz) in water with 0.08 mM of gadolinium (Magnevist 0.5 mmol/ml; Bayer Vital, Leverkusen, Germany) yielding T1 ≈ 300 ms.
- High-19F concentration reference cap (Figure 1B): homogeneous mixture of 60 mM NPs in 1 mL 0.75% agarose (dimensions 20 × 15 mm2; thickness ≈ 1.5 mm) sealed within PARAFILM (thickness = 0.14 mm; Sigma-Aldrich, St. Louis, MO, USA).

Both sets of maps were acquired separately to consider tube thickness (0.8 mm). This accounts for more than half the number of pixels of the reference cap.

Phantoms and mice were used to evaluate the performance of the B1 correction methods as follows:

- Test uniform phantom: 15-mL tube containing 0.2 mM of 2,2,2-trifluorethanol in water. To achieve T1 ≈ 1870 ms (in vivo PFCE-NPs T1; see section 3), 0.006 mM gadolinium was used.
- In vivo and ex vivo mice: EAE was induced in female SJL/J mice as described.26 Animals were weighed and scored (0–5) daily for disease signs. Intravenous injections of 19F-NPs (10 µmol PFCE in 200 µL) were administered daily from day 5 following EAE induction until the experiment end. Respiration and temperature were monitored during measurements. All animal experiments were approved by the Animal Welfare Department of the LAGeSo in Berlin and in accordance with international guidelines (86/609/EEC).

In vivo 19F-NPs T1 for model-based corrections was calculated in n = 3 EAE mice using a combination of ketamine-xylazine (initial dose 400 µL, followed by 100–200 µL injections administered intraperitoneally every 45 minutes until the end of the MR examination) to avoid confounding 19F signal. Ex vivo T1 of PFCE-NPs was computed on n = 3 ex vivo phantoms prepared as described subsequently.

In vivo 1H and 19F images were acquired on another n = 3 EAE mice from which n = 2 animals are shown. These were anesthetized with isoflurane (2% initial dose, 0.5%–1% maintenance). 1H- and 19F-MRI of an ex vivo phantom containing the central nervous system (CNS) of a EAE mouse perfused/fixed as described8 and embedded in a 15-mL tube filled with 4% paraformaldehyde (Santa Cruz Biotechnology, Dallas, TX, USA) were also performed.

- A reference cap (24 mM NP emulsion) was prepared as described previously for 19F-CRP adjustments and signal quantification of in vivo and ex vivo mice. A similar construction of smaller dimensions (10 × 5) mm3 was prepared to fit within the volume resonator.

2.4 | Magnetic resonance experiments

19F-CRP reference power calibrations were performed on a 1-mm slice parallel and close to the probe surface. All images were acquired as repetitions in axial and sagittal orientation. Noise scans (number of excitations [NEX] = 1 and reference power = 0 W) were acquired after each RARE image for SNR map computation.

2.4.1 | 19F-CRP B1 field characterization

The B1 fields of the 19F-CRP were characterized.23 Separate sets of maps were determined using the low-T1 uniform phantom and the high-19F concentration reference cap as follows:

- Flip angle (FA) mapping: FLASH measurements with TE/TR = 2.16/3000 ms, FOV = (25 × 25) mm2, matrix = 96 × 96, 5 slices (gap/thickness = 0.5/2 mm), 1 hour per orientation; FA = 60°/120°/240° (uniform phantom) and FA = 60°/120° (reference cap).
- B1 mapping: FLASH measurements with parameters as described previously and FA = 5° in both cases.
**TABLE 1** Detailed overview of measurements, RF coils, and samples used

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Abbreviation: CRP, cryogenically-cooled transceive surface RF probe; FA, flip angle; NPs, nanoparticles. ref., reference.

aIndicative values: Scan times may vary when using different scan parameters (e.g., spatial resolution, echo train length, TR).
For the sensitivity correction method, RARE images of the low-$T_1$ uniform phantom were acquired (TE/TR = 4.62/1000 ms, same geometry, echo train length (ETL) = 32, bandwidth = 50 kHz, centric encoding with flipback, 1 hour per orientation). All $^{19}$F-RARE images were measured using these scan parameters with varying acquisition times.

2.4.2 | $T_1$ relaxation times (reference, ex vivo, in vivo) of PFCE-NPs

Due to the inherent $^{19}$F characteristics (low SNR, signal sparsity, lack of an a priori known location), determining in vivo $T_1$ with $T_1$ mapping was unfeasible. We applied MRS techniques using the $^1$H/$^{19}$F volume resonator as follows:

- Non-localized spectroscopy (block pulse, 10 TRs [250–10 000 ms], number of acquisitions [NA] = 64, acquisition time [TA] = 35 minutes) to compute $T_1$ values of the two reference caps (24 mM, 60 mM).
- Localized spectroscopy (PRESS) to compute $T_1$ values in the brain after $^{19}$F-NP administration in ex vivo phantoms (n = 3, 12 TRs [250–15 000 ms], NA = 64, TA = 32 minutes) and in vivo mice (n = 3, 8 TRs [412.5–13 000 ms], NA = 128, TA = 1 hour 8 minutes). A default $B_0$ field map was measured before each experiment to optimize shim adjustment (MAPSHIM) computed on $^1$H using a 3D cuboid shape fitting the mouse brain.

2.4.3 | Uniform phantom MR measurements

A $^{19}$F-MR image of the test uniform phantom was acquired with the $^{19}$F-CRP (RARE: same parameters, 3 seconds, axial orientation) to assess $B_1$ correction performance in low SNR scenarios far from the probe surface. A reference $^{19}$F image (RARE: same parameters, 1 hour) and a $T_1$ map (RARE with variable TR [250–10 000 ms], ETL = 2, linear phase encoding, other parameters same as RARE scan, 24 minutes) were acquired with the $^1$H/$^{19}$F volume resonator for comparison.

2.4.4 | Ex vivo and in vivo MR measurements

Slice planning and anatomical images (FLASH: TE/TR = 3/120 ms, same FOV, matrix = 256 × 256, TA = 30/15 minutes per orientation ex vivo and in vivo, respectively) were acquired with the 72-mm volume resonator.

$^{19}$F-MR images were measured with the $^{19}$F-CRP with (RARE: same parameters, 15 minutes per orientation both ex vivo and in vivo) and without (RARE: same parameters, 6 hours/45 minutes per orientation ex vivo and in vivo, respectively) reference cap.

Reference images were acquired with the $^1$H/$^{19}$F volume resonator in ex vivo phantoms: reference cap ($^{19}$F RARE: same parameters, 30 minutes per orientation) and phantoms ($^{19}$F RARE: same parameters, 6 hours per orientation; $^1$H FLASH: same parameters, 1 hour per orientation).

2.5 | Data analysis

Data analysis was performed using MATLAB (The MathWorks, Natick, MA, USA).

2.5.1 | MRI data preprocessing

All data followed the same pre-processing workflow:

1. Complex averaging over smaller subsets of the total number of repetitions to mimic different scan times followed by a sum-of-squares (SoS) combination of the two channels ($^{19}$F-CRP):
   • Uniform phantom: one subset of a 3-second acquisition.
   • Ex vivo phantoms: four subsets corresponding to 15-minute and 1-3-6-hour acquisitions. Same with $^1$H/$^{19}$F volume resonator for comparison.
   • In vivo mice: three subsets corresponding to 15-30-45 minutes.
   • Reference caps: one subset corresponding to the total scan time.

2. Noise bias correction:
   • $^{19}$F-CRP: noncentral $\chi$ distribution using a lookup table for $n = 2$ channels.
   • Volume resonator: Rician distribution using a lookup table for $n = 1$ channels.

3. Thresholding (SNR cutoff = 3.5) and removal of isolated groups of < 3 connected pixels.

2.5.2 | $^{19}$F-CRP $B_1$ field characterization and RARE SI model computation

The $B_1$ maps were computed and denoised as detailed (10th-order and 8th-order polynomials for the low-$T_1$ uniform phantom and the high-$^{19}$F concentration reference cap, respectively).

The RARE SI model was calculated as a function of FA and $T_1$ relaxation value (SI = $f(FA,T_1)$) using extended phase graphs (EPGs). This algorithm provides a tool that depicts the magnetization response and allows computing echo intensities in multi-pulse MR sequences.
RARE scans with the same MR parameters as above were simulated for 20 equispaced $T_1$ values (150–2050 ms) and 32 excitation FAs (5°–160° in 5° steps). Finally, an 8th-degree polynomial was fitted to the simulated data for faster computation of results for arbitrary FAs and $T_1$ values, which did not introduce any oscillations or error within the desired parameter space ($R^2 = 1.0$, root-mean-square-error (RMSE) = $5.5 \times 10^{-4}$).

2.5.3 | $T_1$ relaxation times (reference, ex vivo, in vivo) of PFCE-NPs

PFCE-NPs typically show a single peak at $f \approx 376.629$ MHz. A Lorentzian line-broadening (factor = 70) and automatic phase correction (TopSpin 2.1) were applied. To compute $T_1$ values from MRS data, peak values were fitted as SI vs. TR datapoints on an exponential growth. Mean values and SDs were computed. $T_1$ values were used to correct $B_1^+$ using the model-based method.

2.5.4 | $B_1$ correction methods

The $B_1$ of $^{19}$F-CRP images was corrected using the sensitivity (uniform phantom) and model-based (reference caps, phantoms, and in vivo mice) methods. All post-processing was performed using software openly available on Github (pramosdelgado/B1correction-toolkit).

2.5.5 | $^{19}$F signal quantification

The 24-mM reference cap was used as reference to determine absolute $^{19}$F concentrations as follows:

$$c_{sample} = \frac{\overline{SI}_{sample} \times c_{ref}}{\overline{SI}_{ref}}$$

where $\overline{SI}_{sample}$ and $\overline{SI}_{ref}$ are the SIs for the sample and the reference, respectively, and $c_{sample}$ and $c_{ref}$ are the corresponding concentrations. To compute $\overline{SI}_{ref}$, a square-shaped region of interest (ROI; 3 × 3 pixels) was selected in a $B_1^+$-corrected homogeneous region, in the center of the reference cap.

2.6 | Monte Carlo SNR simulations to estimate the $^{19}$F concentration uncertainty

Given the sparse nature of $^{19}$F images and the spatially varying $B_1$ fields of the $^{19}$F-CRP, we computed concentration uncertainty maps after $B_1$ correction as follows (Figure 2):

**Step 1.** Monte Carlo SNR simulations (1000 iterations) were performed using measured ($T_1$ values) and synthetic data (SI computed using the simulated RARE SI model). Simulation parameters (Table 2) were defined to mimic realistic excitation FAs, $B_1^+$-values, and SNRs within the sample. Shorter parameter ranges were chosen for the reference cap after inspection of the central region of the FA, $B_1^+$, and SNR maps obtained (section 2.5.5). This was crucial to reduce matrix size and avoid memory problems.

**Step 2.** Noise levels for the prescribed SNR values were fixed for a 90° excitation and $B_1^+$ = 1 using a “reverse model-based correction” (inverse steps of the model-based correction).

**Step 3.** For each combination of reference and sample FA, $B_1^+$ and $T_1$ values, the CRP SI (for reference and sample) was calculated and separated into two channels. For each Monte Carlo iteration, complex Gaussian noise was added to both channels, and a SoS reconstruction was computed to simulate a noncentral $\chi$ distribution. A noise bias correction was performed as described, followed by a model-based correction. Finally, the concentration was estimated using equation (1). The mean SNR and mean and SD of the corrected SI throughout the 1000 iterations were determined for both reference and sample, along with the mean and SD of the concentration. Since the Monte Carlo samples conformed to a Gaussian distribution of mean $\approx 1$ (section 3), the corresponding uncertainties in corrected SI and concentration were defined as SD × 100 (%).

**Step 4.** To compute the uncertainty map of an acquired $^{19}$F image, measured data (FA, $B_1^+$, and SNR maps, $T_1$ value) were fed to the corresponding Monte Carlo uncertainty model. The uncertainties were interpolated pixel-wise using a simple linear regression after logarithmically transforming the SNR and uncertainty data and eliminating SNR values < 1.

2.7 | Correction method evaluation and validation

$B_1$ correction methods were validated using the following methods on the uniform phantom:

2.7.1 | Central profile plots of uniform phantoms

We quantified the improvement in image homogeneity by plotting normalized vertical SI profiles of original, corrected, and reference images against the distance from the CRP surface.
Parameter definition

- Number of Monte Carlo iterations
- Ground truth values for SI and concentration (sample, ref.)
- Sample: excitation FA, normalized $B_1^*$; $T_1$ relaxation times (ex vivo, in vivo) and SNR
- Reference cap: excitation FA, normalized $B_1^*$; $T_1$ relaxation times (in agarose) and SNR
- Lookup table for bias correction
- RARE SI model

Compute fixed noise levels

1. “Reverse” model-based correction to compute $S_{\text{CRP}}$:
   - Sample: FA = 90°, $B_1^*$ = 1, $T_1$ = 1869 ms
   - Ref.: FA = 90°, $B_1^*$ = 1, $T_1$ = 936 ms
2. Calculate noiseSigma for defined SNR values

SNR simulations

- For each FA, $B_1^*$ and $T_1$ relaxation time of ref. and sample:
  1. Compute pixel CRP SI ("reverse model-based correction")
  2. Separate in 2 channels
- For each Monte Carlo iteration:
  1. Add complex Gaussian noise and perform SoS reconstruction
  2. Perform noise bias correction
  3. Model-based $B_1^*$ correction
  4. Determine concentration: $c_{\text{sample}} = \frac{S_{I_{\text{sample}}}}{S_{I_{\text{ref}}}} \times c_{\text{ref}}$

3. Compute statistics:
   - Mean SNR
   - Mean and SD of corrected SI
   - Mean and SD of concentration (signal quantification)

Uncertainty map estimation

Error dataset for desired (FA, $B_1^*$, $T_1$)

Linear fit in log-log plot and uncertainty map calculation (find uncertainty for desired SNR)

Concentration was also estimated. Statistics including mean SNR, mean and SD of corrected signal intensity (SI), and mean and SD of the concentration were computed after each run. These simulations are then used to derive uncertainty maps for the measured data using the FA, $B_1^*$, $T_1$, and SNR measured at each pixel using a linear regression in a log-log plot (error vs. SNR). Abbreviations: Ref., reference; SoS, sum-of-squares.
2.7.2  Image homogeneity assessment

The percentage of integral uniformity (PIU)\(^{36}\) was computed for three internally tangential circular ROIs with increasing diameter placed on the central vertical line filling the region with signal.

2.7.3  Quantification performance

Ten ROIs were placed at pseudo-randomized positions (Figure 5B) on original, corrected, and reference images. Mean absolute percentage errors (MAPEs) were computed relative to the reference (volume resonator) images, as follows:

\[
MAPE = \left( \frac{\overline{SI}_{\text{reference}} - \overline{SI}_{\text{corrected}}}{\overline{SI}_{\text{reference}}} \right) \times 100 \%
\]

where \(\overline{SI}_{\text{reference}}\) and \(\overline{SI}_{\text{corrected}}\) are the mean SI in reference and corrected images.

A value was calculated for the original image and the three corrections summing over an increasing number of ROIs (top to bottom), with increasing distances from the CRP surface and decreasing SNR. Corrections were classified as excellent (MAPE \(\leq 10\%\), green), good (10\% < MAPE \(\leq 25\%\), orange), or unacceptable (MAPE > 25\%, red).

2.7.4  Statistics

Normality was assessed using the D’Agostino-Pearson test. Because none of the MAPEs on original or corrected data conformed to a Gaussian distribution, a Friedman non-parametric one-way repeated-measures ANOVA test was used followed by Dunn’s post-hoc test, in which all corrections were compared to original data (\(p\)-values < 0.001 were considered significant). The statistical analysis was performed using GraphPad Prism 5 (GraphPad Software, La Jolla, CA, USA).

3  RESULTS

3.1  \(^{19}\text{F-CRP B}_1\text{-field characterization and RARE SI model}\)

The sensitivity maps (Figure 3A,D) and the FA maps (relative to an excitation FA = 90°, Figure 3B,E) of the \(^{19}\text{F-CRP}\) revealed a strong decline with increasing distance from the RF probe surface, in both axial and sagittal orientations.
The $B_1^+$ inhomogeneity is clearly depicted in Figure 3C (axial) and Figure 3F (sagittal), which show the normalized central vertical profile lines. The maximum distance until which there is signal above the detection threshold (SNR > 3.5) is, in this case, approximately 14.6 mm from the CRP surface.

Figure 3G shows the 3D view of the RARE SI model simulated using EPG simulations. The SI was modeled as a function of FA and $T_1$. The SI demonstrates a lower SI with increasing $T_1$ (Figure 3H) and maximal SI for FA = 90° (Figure 3I). When using EPG simulations, the hybrid and sensitivity methods yielded the same results up to a constant factor (Supporting Information). Therefore, we only used the sensitivity and model-based correction moving forward.

3.2 | $T_1$ relaxation times (reference, ex vivo, in vivo) of PFCE-loaded NPs

Calculated $T_1$ values for PFCE-NPs in agarose (reference caps, $935.9 \pm 10.0$ ms) using non-localized MRS agreed with previously published values at 9.4 T. $T_1$ values of $^{19}$F-NPs in inflammatory lesions in the brain (PRESS) were $818.1 \pm 13.4$ ms (ex vivo) and $1868.7 \pm 43.9$ ms (in vivo). This indicated an effective reduction of 117.8 ms in $T_1$ for ex vivo compared to the reference caps, and an increase of nearly 1 second in $T_1$ in vivo measurements. Exemplary spectra are shown in Supporting Information Figure S1.

3.3 | Monte Carlo SNR simulations to estimate the $^{19}$F concentration uncertainty

Figure 4A–C shows the concentration uncertainty (uncertainty = SD $\times$ 100 [%]) for all FAs/$B_1^+$ and three SNR values fixed for FA = 90°, $B_1^+$ = 1, and $T_1 = 1869$ ms (in vivo). For the reference, representative values (FA = 60°, $B_1^+$ = 0.8) were used. The level of uncertainty increases with decreasing FAs and $B_1^+$. This trend is more pronounced for regions farther away from the RF probe surface. The contour lines represent SNR values. The green and red
isolines depict the border of the regions where uncertainty ≤ 10% and ≤25%, respectively. These borders occur at SNR ≈ 10.1 and SNR ≈ 4.25, respectively, independent of the FA/B₁ and SNR combination.

We studied the linear dependence of the SD of both corrected SI and concentration on SNR for exemplary data (FA = 90°, B₁ = 1.0, in vivo T₁) using the model-based method (Figure 4D, linear fit, dashed orange line). The corrected SI of the sample (blue dots) demonstrated a linear trend throughout the SNR range. The concentration SI decay away from the RF probe data (FA = 90°, 24-300, 1, 3, and 6 hours [NEX = 300], 1, 3, and 6 hours [NEX = 1200/3600/7200]) yielded improved PIUs (56.7% for model-based and 32.4% for sensitivity corrections). In general, PIU degrades with increasing distance from the RF probe, where acquired image artifacts prevail (Figure 5H).

### 3.4 Uniform phantom MR measurements

#### 3.4.1 Corrected images

B₁ correction performance was assessed in a low-SNR scenario at regions far from the probe surface using a low-concentration uniform phantom and a short acquisition. The SNR map is shown in Figure 5A. The original image shows a steep SI decay away from the RF probe surface, typical of transceive surface RF coils (Figure 5B). Compared with the reference image, B₁-corrected images (Figure 5C,D) yielded uniform SIs over the FOV (Figure 5E). A ghosting artifact due to fast RARE imaging is present in the uniform phantom image used for the sensitivity method, and in the test uniform phantom, producing an overshoot in the sensitivity-corrected image far from the probe surface.

#### 3.4.2 Central profile plots

Corrected SI profiles demonstrated close correspondence with the reference RF coil (green area) up to a distance of approximately 6–7 mm from the CRP surface for our specific scanning parameters, dimensions of the RF coil, and SNR (Figure 5F).

#### 3.4.3 Image homogeneity assessment

The calculated PIU in the reference image was 91.4% within the largest ROI (distance from CRP surface = 7.8 mm), indicating no substantial inhomogeneities across the image. In contrast, a PIU of 13.6% was computed for the original image within the same ROI. Corrections yielded improved PIUs (56.7% for model-based and 32.4% for sensitivity corrections). In general, PIU degrades with increasing distance from the RF probe, where acquired image artifacts prevail (Figure 5H).

#### 3.4.4 Quantification performance and statistics

According to our MAPE classification, only the model-based correction provided excellent results for SNRs between 38 and 7 (Figure 5G; ROIs = 1–7, distance = 2.1–6.3 mm). Uncorrected images showed high errors within this SNR range (84.7 ± 85.8%). Within this region (distance = 2.1–6.3 mm), the model-based correction performed best (7.7 ± 4.7%), followed by the sensitivity correction, which yielded good results (12.2 ± 8.2%). Both corrections provided equally good results (model-based 16.2 ± 16.5%, sensitivity 19.7 ± 16.6%) up to the eighth ROI (distance = 2.1–6.5 mm), in contrast to uncorrected images (89.9 ± 95.6%). When considering all ROIs (distance = 2.1–7.6 mm), only the model-based correction (19.7 ± 18.9%) yielded good results. In this case, the sensitivity correction provided unacceptable results (35.5 ± 33.3%), but was still lower than the MAPE of uncorrected images (105.8 ± 125.9°). Figure 5G also shows similarities between the proposed ranges using simulations (uncertainty ≤ 10% when SNR ≥ 10.1 and uncertainty ≤ 25% when SNR ≥ 4.25) and experimental results.

The model-based correction performed best overall, significantly reducing quantification errors compared with original mean errors (both B₁ correction methods p < 0.001; Figure 5I). Therefore, this method was used for further B₁ corrections.

### 3.5 Ex vivo MR measurements

Concentration maps of the ex vivo EAE phantom were computed for different measurement times (15 minutes [NEX = 300], 1, 3, and 6 hours [NEX = 1200/3600/7200]) using the 24-mM reference cap in images acquired with the reference...
volume resonator (Figure 6A) and original $^{19}$F-CRP images (Figure 6B). Qualitative comparison of the reference images after 3 hours and original CRP images after 15 minutes revealed distinct similarities, demonstrating the remarkable SNR capabilities of the CRP. However, the $^{19}$F signal at the lymph nodes, indicating accumulation of $^{19}$F-labeled inflammatory cells (white arrows) in reference images was absent in the $^{19}$F-CRP images, as the lymph nodes are located too far away from the CRP surface to be detected.

Assessment of the $^{19}$F concentration shown by original CRP images and corresponding model-based $B_1$-corrected images (Figure 6D) demonstrated that correction considerably improved the concentration estimation, compared with reference images (ground truth). The SNR maps from original CRP images showed the expected increase of SNR with scan time (Figure 6C), translating to fewer uncertainties in concentration (Figure 6E). Overall, the uncertainty maps indicated the reliability of the $B_1$-corrected concentration maps, with most pixels being green (uncertainty $\leq 10\%$) or orange ($10\% < uncertainty \leq 25\%$). Images corresponding to the axial orientation are shown in Supporting Information Figure S2.

### 3.6 | In vivo MR measurements

We studied the performance of the model-based correction in a typically time-constrained and low-SNR in vivo EAE $^{19}$F-MRI experiment.

The first animal shown (Figure 7) exhibited severe clinical symptoms (score = 2.5), whereas the second (Figure 8) presented moderate clinical symptoms (score = 1.5). Images were acquired in axial and sagittal orientations for 15, 30, and 45 minutes (NEX = 300/600/900). Images corresponding to the axial orientation are shown in Supporting Information Figures S3 and S4.

Concentration maps of uncorrected images of mouse 1 (Figure 7A) showed an overestimation of $^{19}$F
**FIGURE 5** Uniform phantom validation. (A) SNR map, (B) original, (C, D) corrected, and (E) reference images, respectively. The original image includes the placement of the 10 regions of interest (ROIs) selected for error calculations. (F) Normalized SI profiles perpendicular to the RF coil surface. (G) Mean absolute percentage error (MAPE) of original and corrected images for an increasing number of ROIs demonstrates a remarkable reduction in errors after B₁ correction compared to original images. The *model-based correction* provides quantitatively good results in regions far from the RF probe. (H) Percentage of integral uniformity (PIU) of corrected images shows a quantitative improvement in homogeneity in comparison with original images. (I) Statistical assessment of SI accuracy. Whiskers represent the 5th and 95th percentiles. Asterisks indicate statistical significance compared to uncorrected images.
concentrations in regions close to the RF probe surface, which correspond to meningeal inflammatory cell infiltration, common in EAE. White arrows indicate external signals (i.e., in ears and other adjacent tissues), which are not corrected when located outside of the FA/B₁ maps. The SNR maps (Figure 7B) correlate with the original concentration maps.

Following the model-based B₁ correction, concentration maps (Figure 7C) showed reduced ¹⁹F concentration in regions close to the RF probe and increased ¹⁹F concentration in regions with high SNR far from the CRP surface. The reliability of the correction is represented by the concentration uncertainty maps that mostly show values with 10 < uncertainty ≤ 25% (orange pixels) and ≤ 10% (green pixels) especially at higher SNR (Figure 7D).

Compared to mouse 1, mouse 2 presented with more ¹⁹F signal, even though its disease score was less severe. This is evident from the original concentration maps (Figure 8A) and corresponding SNR maps (Figure 8B). Mouse 2 exhibited meningeal inflammation, visible as a thin layer of ¹⁹F signal with an SNR ranging from 3.6 to 49.5 and ¹⁹F concentrations ranging from 0.1 to 1.7 mM, as well as inflammatory cell accumulation in deeper regions of the brain. After applying the model-based correction (Figure 8C), concentration maps showed an expected reduction in ¹⁹F concentration in the meninges and an increase in features far from the CRP surface. Corresponding concentration uncertainty maps (Figure 8D) demonstrate the reliability of the B₁ corrections, with most pixels being orange (10% < uncertainty ≤ 25%) and green (uncertainty ≤ 10%), especially at higher SNR.
DISCUSSION

The potential of $^{19}$F-MR has long been recognized. However, low in vivo $^{19}$F concentrations demand SNR-enhancing strategies. Transceive surface RF probes such as the $^{19}$F-CRP maximize SNR but their inhomogeneous $B_1$ field hampers quantification. To date, efforts in $B_1$ field correction for $^{19}$F-MRI have been scarce, and usually limited to less complex imaging techniques.

This study builds on our previous work on $B_1$ correction methods tailored for $^1$H transceive surface RF probes and SNR-efficient RARE imaging, to enable $^{19}$F signal quantification in low SNR time-constrained scenarios. Low-concentration uniform phantom images showed considerable increase in homogeneity after $B_1$ correction even in low-SNR regions distal from the coil. Ex vivo concentration maps using reference caps demonstrated substantial improvement in concentration estimation, compared with reference images. We established a method to determine concentration error after $B_1$ correction using Monte Carlo SNR simulations and an acquisition workflow to co-locate $^{19}$F-CRP images with anatomical images from an external volume resonator. Furthermore, first in vivo $^{19}$F-nanoparticle $T_1$ values were determined in EAE brains to compute model-based corrections. Successful implementation ultimately yielded the first quantitative in vivo $^{19}$F-MR images of inflamed EAE brains using a $^{19}$F-CRP.

Interestingly, differences in $T_1$ were observed for PFCE-NPs in reference caps, ex vivo, and in vivo. This is in agreement with previous studies showing significant changes in $^{19}$F-MRI images of inflamed EAE brains using a $^{19}$F-CRP.

By introducing EPG simulations, here we reduced the burden of our previous strategy of preparing and scanning several samples with different $T_1$ to compute the RARE SI model. This also improved the accuracy of the model by essentially eliminating possible imprecisions introduced by measurements, especially at low FAs where SIs corresponding to different $T_1$s are closer to each other. We found using EPG simulations that the hybrid and sensitivity methods yielded the same results, up to a constant factor. Imperfections originating from a measured model...
instead of EPG simulations disturb the symmetry underlying this degeneracy, leading to slight differences between the hybrid and sensitivity methods. This demonstrates that simulations have a clear advantage, which we expect would also be true for other MR sequences lacking closed-form SI equations.

The use of higher ETLs to further improve SNR through signal averaging produced ghosting artifacts in uniform phantoms (in test images, but also images used for sensitivity correction) in regions where $^{19}$F signal was lower. This effect has been widely recognized and produced an abnormal increase of signal with the sensitivity method in regions adjacent to the artifact, which could not be removed even when changing the phase-encoding direction. The model-based correction was affected to a lesser extent (test images still showed ghosting artifacts), since this correction uses $F_A$ and $B_1^-$ maps computed with FLASH images. This was observed when correcting the uniform phantom in which the model-based correction yielded MAPEs lower than 25% for all ROIs, and calculated PIUs were equally higher than those achieved with the sensitivity method.

Therefore, we conclude that the model-based correction method is more robust than the sensitivity method, which poses some constraints in MR scanning parameters.

Furthermore, the uniform phantom was prepared with $^{19}$F concentration (0.2 mM) and SNR (range 50 to 0) comparable to those achieved in EAE mice administered with PFCE-NPs (maximum $^{19}$F concentration 2 mM, SNR between 50 and 0 in all cases). Because in transceive surface RF probes the SNR is much higher when close to the RF probe, the $B_1$ correction approach and uncertainty propagation model were assessed in realistic scenarios and validated for low SNRs far away from the RF probe (Figure 5F–I).

Reference caps placed above the phantoms or mouse heads were developed to allow for reference power calibrations. Little extra time was needed to acquire separate reference images to compute $^{19}$F concentrations. Furthermore, individual $B_1$ maps were measured to correct more pixels in the reference caps, since the wall thickness of the 15-mL tube (0.8 mm) excluded more than half of the pixels of the reference. Corrections of the reference caps were nevertheless of poorer quality, with $B_1$ inhomogeneities at the sides.
This was expected due to the large gradient close to the probe surface. Also, reference power adjustments may not be reliable in the close slices, further demonstrating that FA calibration is non-trivial and could be improved.42,47

Reliable B1 correction is indispensable for robustly quantifying the 19F signal when using the 19F-CRP in studies using 19F-NPs to measure the inflammatory burden in EAE in vivo. In this study we presented two EAE animals with discrepancies between 19F signal and clinical score: the animal with lower clinical severity showed more 19F signal. This reflects the clinico-radiological paradox, well described in MS48 and EAE,49 whereby clinical status and radiological findings diverge, underscoring the urgent need to establish more quantitative MRI methods to assess disease severity objectively, such as that presented in the current study.

We performed Monte Carlo SNR simulations to estimate SI quantification uncertainties. Simulations were designed to include a wide SNR range (Table 2), taking into account the typically low SNR values for 19F (SNR = 0–10 in 0.5 steps) as well as higher SNRs (SNR up to 1500). We found that concentration uncertainty maps yielded a linear dependence of the uncertainty on SNR, with constant regions (≤ 10% with SNR ≥ 10.1 and ≤ 25% when SNR ≥ 4.25). This is consistent with the results previously demonstrated for 1H imaging, in which SNR was not limited. These SNR requirements are highly relevant for the experimental implementation of our approach and aim to guide other researchers to balance scan time with the uncertainty of the quantification of low-SNR 19F RARE-MRI applications.

To examine the accuracy of B1-corrected ex vivo concentration maps, these were compared to those obtained with a volume resonator. Despite the best efforts to select an identical anatomical position with both volume resonators, minor differences in 1H might cause slight changes in the visible 19F signal. Nevertheless, there was overall good agreement in 19F features and corresponding concentrations, confirmed by the computed uncertainty maps. In vivo error concentration maps showed positive results even when SNR values achieved were significantly lower than ex vivo, due to reduced scan times. Future studies using 3D-RARE combined with accelerated acquisition could help further improve concentration errors.50,51 Moreover, adiabatic pulses could be an interesting addition to 3D-RARE acquisitions to further improve B1+ field uniformity up to a certain region.52,53 A subsequent model-based B1 correction could be of value to increase the B1-corrected area.

To conclude, we demonstrated a workflow that allows 19F signal quantification using a model-based B1 correction method together with a single-tuned transceive surface RF probe and RARE. We also highlight several issues that should be considered when performing similar studies. This approach remarkably improved concentration errors from > 100% to < 25%. B1 correction methods will be critical to ensure that the detected 19F signal depends exclusively on 19F spin density and not on distance to the RF probe surface, while utilizing the SNR benefit provided by 19F-CRPs. These results are particularly promising for future clinical applications,54–57 in which the lower SNR achieved at clinical field strengths necessitates the use of transceive surface RF probes.

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Conflict of Interest
Andre Kuehne, Alonso Vázquez and Helmar Waiczies are employees of MRI.TOOLS, Berlin, Germany. Thoralf Niendorf is founder and CEO of MRI.TOOLS. Andreas Pohlmann is currently an employee of Siemens Healthineers. All other authors declare no conflict of interest.

Data Availability Statement
The code and data that support the findings of this study will be openly available in GitHub at https://github.com/pramosdelgado/B1correction_for_19F.

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References


FIGURE S1 Exemplary spectra used for $T_1$ calculation for (A) reference cap containing 24mM $^{19}$F-loaded NPs (non-localized spectroscopy), (B) ex vivo CNS of an EAE mouse with administered $^{19}$F-loaded NPs prior to perfusion (PRESS), and (C) in vivo mouse with active EAE and administered $^{19}$F-loaded NPs (PRESS). Measurements were performed using a $^1$H/$^{19}$F volume resonator. Selected TR = 10000 ms

FIGURE S2 Ex vivo phantom (score=2.0) in axial orientation for increasing scan times (15 minutes, 1 hour, 3 hours and 6 hours). Reference images (A) acquired with the $^1$H/$^{19}$F volume resonator show less $^{19}$F signal in the brain compared to $^{19}$F-CRP images (B). The steep gradient in $B_1$ field of the $^{19}$F-CRP prevents from detecting the prominent lymph node signals in contrast to the volume resonator. SNR maps for the CRP images are presented in (C). $B_1$-corrected images show concentration values closer to the reference obtained with the volume resonator (D). Uncertainty maps (E) reveal the reliability of the $B_1$-corrected concentration maps, with most pixels indicating green (uncertainty $\leq 10\%$) and orange ($10\% < uncertainty \leq 25\%$) values

FIGURE S3 In vivo EAE mouse 1 (score = 2.5) in axial orientation. Concentration maps of original images (A) show an initial overestimation of the $^{19}$F concentration in regions close to the RF probe surface (e.g. meninges) which partly correspond with regions with high SNR (B). After performing the model-based $B_1$ correction (C), $^{19}$F concentration maps are computed. Their reliability is presented by the uncertainty maps (D) which show green (uncertainty $\leq 10\%$) and orange ($10\% < uncertainty \leq 25\%$) values for most pixels

FIGURE S4 In vivo EAE mouse 2 (score = 1.5) in axial orientation. (A) Concentration maps of original images present signals in the meninges as well as in deeper regions of the brain, indicating increased inflammatory cell accumulation. (B) SNR maps show high SNR at pixels at the top of the mouse head and a reduced SNR in regions distant to the RF probe. After applying the model-based $B_1$ correction (C), concentration maps show an expected reduction in $^{19}$F concentration in the meninges and an increase in pixels far from the CRP surface. Corresponding uncertainty maps (D) demonstrate the reliability of the $B_1$-corrected concentration maps, with most pixels indicating green (uncertainty $\leq 10\%$) and orange ($10\% < uncertainty \leq 25\%$) values


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