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METHODS

A Numerical Co-Simulation Framework for Interspecies Dual-Subject Preclinical RF Coil Design

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ABSTRACT This study proposes a co-simulation framework for optimizing and validating design parameters prior to preclinical functional magnetic resonance imaging (fMRI) studies. The primary objective of this study was to establish a face-to-face fMRI platform for investigating social interactions between rats and/or mice. Conventional hardware setups for single-species imaging often result in suboptimal signal uniformity and sensitivity when applied to interspecies scenarios. To address this challenge, a two-stage co-simulation framework is proposed. The first stage uses electromagnetic simulations to optimize geometry-dependent radiofrequency (RF) coil designs, and the second simulates magnetic resonance imaging (MRI) signals to evaluate image performance and field homogeneity across different coil configurations and animal pairings. A range of RF coil geometries, such as the birdcage, loop, and fish-trap types, with different designs and dimensions, were modeled to maximize the $|B_1|$ -field strength and uniformity across the brain volumes of two rodents while maintaining compatibility with the available preclinical MRI system, which consists of a single-transmit channel. The simulation results indicated that, for the rat–rat combination, a medium-sized birdcage coil generated the strongest $|B_1|$ -field; for the mouse–mouse pair, the medium-sized fish-trap coil offered the best performance; and for the rat–mouse pair, a medium-sized loop coil achieved the most uniform $|B_1|$ -field and balanced image intensity across the brain regions of interest. This simulation-based optimization strategy provides a reliable and efficient framework for designing and evaluating RF coils before in vivo experiments. It can also define the design parameters optimized for dual-species fMRI to ensure consistent and balanced signal performance across rodent models in future applications.

INDEX TERMS MRI, preclinical, fMRI, electromagnetic simulations.

I. INTRODUCTION

Social interaction is a fundamental component of behavior across species. It supports survival through mating, parenting, cooperation, and social learning and shapes learning via reward, affiliation, and social hierarchy cues [1], [2], [3]. These functions arise from coordinated neural computations

spanning sensory processing, valuation, arousal regulation, and action selection [4], [5], [6]. Social interaction is a key area of research in translational neuroscience because social dysfunction is prominent in neurodevelopmental and psychiatric disorders [7], [8], [9], [10], [11]. Recently developed “second-person” frameworks propose that social exchange can be examined as a reciprocal, coupled dynamical process between individuals [12]. Human hyperscanning studies using functional magnetic resonance imaging (fMRI) have

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reported systematic inter-brain coupling during live interactions and introduced analytical approaches to quantify dyadic coordination [13], [14], [15]. However, conventional magnetic resonance imaging (MRI) settings impose structural constraints on the geometry of such interactions and experimental control. These constraints include limited face-to-face engagement, restricted interaction modalities, and limited flexibility in systematically manipulating interaction parameters. As a result, interaction-related computations may be difficult to distinguish from confounds associated with task structure, environmental constraints, and between-subject heterogeneity, leading to challenges in mechanistic interpretation. Preclinical rodent fMRI offers a complementary platform in which interaction conditions can be tightly controlled and replicated across cohorts while enabling integration with invasive measurements and causal perturbation approaches, which can be used to investigate mechanistic hypotheses linking circuit dynamics to whole-brain fMRI signatures.

fMRI provides a translational bridge by offering non-invasive whole-brain access to the hemodynamic correlates of neural activity [16]. The field of rodent fMRI has expanded with high-field small-bore systems and community efforts toward standardized acquisition and quality control, and high-field mouse studies have demonstrated robust whole-brain readouts under controlled stimuli and optogenetic perturbation paradigms [17], [18], [19], [20], [21], [22]. However, extending single-subject rodent fMRI to face-to-face dyadic imaging introduces constraints that are not addressed by conventional system setups and hardware assumptions. Preclinical scanners are optimized for one subject at the magnet isocenter, and dyadic positioning displaces one or both animals, increasing the susceptibility to distortion and artifacts and introducing asymmetric loading within the bore [23], [24].

In this setting, the performance of the radiofrequency (RF) coil becomes a critical bottleneck. A reliable fMRI readout requires efficient RF excitation and adequate sensitivity within the brain region of interest (ROI), and coil design directly influences the flip-angle fidelity, signal-to-noise ratio (SNR), spatial coverage, and interpretability of functional maps [25], [26]. Birdcage (BC) volume coils provide broad coverage and relatively uniform transmission, whereas loop-based surface coils offer higher local sensitivity but introduce stronger spatial bias and depth-dependent signal decay. These tradeoffs become more stringent when two brains are imaged simultaneously under bore-size and clearance constraints. This problem is further compounded by the single transmit channel common in preclinical platforms, which restricts compensation for asymmetric loading through parallel-transmit strategies. The challenge further increases in dual-species paradigms involving rat–rat, mouse–mouse, and rat–mouse pairings. Differences in body size, head geometry, and dielectric loading can produce unbalanced $|B_1|$ -field distributions and species-dependent sensitivities. Although multi-animal MRI systems and coil arrays have been

developed primarily to improve throughput, such as those used for simultaneous multi-mouse imaging using dedicated receiver elements or cryogenic probes, they are not optimized for face-to-face interaction geometries or for balancing the transmission and reception performances across species within one setup [27], [28]. Accordingly, dyadic face-to-face fMRI, particularly across species, requires coil selection principles and protocol constraints that have been explicitly evaluated under the intended geometry.

A co-simulation workflow can provide a scalable approach to efficiently address this design gap. Full-wave electromagnetic (EM) simulations, such as finite-difference time-domain and the finite element method, can predict the coil loading, tuning and matching behavior, and $|B_1|$ -field distributions in realistic geometries, enabling optimization prior to fabrication [29]. In addition, MRI simulators based on the Bloch and Bloch–Torrey equations can incorporate coil field maps, tissue properties, and sequence timing to predict signal evolution and spatial bias, supporting end-to-end evaluation of the expected image performance [30], [31]. Together, integrated EM and MRI simulations enable a systematic comparison of candidate coils in terms of the $|B_1|$ characteristics and sequence-relevant image sensitivity and uniformity.

This study proposes a simulation-driven protocol for the face-to-face fMRI of dual species (rats and mice) in a 9.4-T preclinical system with a single-transmit-channel constraint. Multiple RF coil geometries and sizes, including BC, loop-based surface coil, and modified BC “fish-trap” (FT) configurations, are evaluated to maximize the $|B_1|$ -field strength while maintaining adequate homogeneity across both brain ROIs. The EM results are coupled with MRI simulations to estimate the signal distribution and expected image quality across species pairings. This framework offers a reproducible pipeline for selecting hardware and protocol parameters prior to in vivo experiments, enabling reliable dyadic fMRI designs for future studies on the face-to-face interactions of rodents.

II. METHODS

The proposed workflow is illustrated in Fig. 1 and comprises three main stages. The first stage involves electromagnetic (EM) simulations, including the design of the three-dimensional (3D) RF coil geometries, positioning of the 3D animal models for each configuration, and assignment of the corresponding tissue electrical properties. The $|B_1|$ -field is extracted from these simulations to evaluate coil performance, as it is essential for both RF transmission and signal reception in MRI. In addition, complex vector fields, including the electric field (E), electric displacement field (D), current density (J), and magnetic flux density (B), are computed and extracted. The D, J, and E fields are used to generate virtual phantoms, whereas the E and B fields are employed to create coil profiles within the MRI simulation environment. These fields are computed using electromagnetic simulations and subsequently exported to MATLAB (MathWorks, Natick, MA, USA) for further processing.

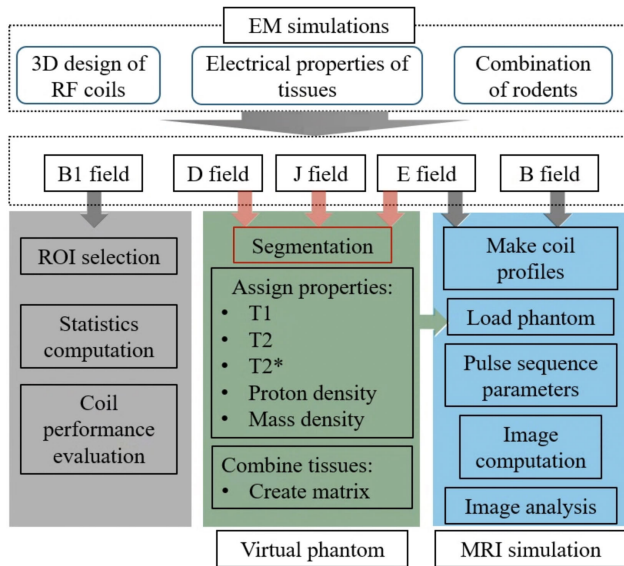


FIGURE 1. The proposed workflow consists of three main stages: EM simulations, virtual phantom creation, and MRI simulation.

The second stage focuses on the creation of virtual phantoms, which involves tissue segmentation and the assignment of relaxation parameters and tissue-specific MRI properties. Finally, in the third stage, MRI simulations are performed by integrating the coil profiles and virtual phantoms. The MRI simulator uses the E and B field distributions imported from MATLAB to generate the RF coil profiles. Subsequently, images are reconstructed based on the selected pulse sequence. At the end of the workflow, image analysis is performed to compare the performance of each coil under different loading conditions and animal configurations.

A. EM SIMULATIONS

The design of the RF coil plays a crucial role in determining MRI performance. Conventional coils used for preclinical imaging are typically optimized for a single animal, with the brain positioned at the center of the coil to achieve a strong and uniform $|B_1|$ -field [26]. By contrast, the proposed experimental setup requires two ROIs corresponding to the brains of two animals placed face-to-face. Therefore, the coil geometries must be re-optimized to provide a sufficient $|B_1|$ -field strength and uniformity across both ROIs simultaneously.

This study investigated multiple RF coil configurations with varying geometries and sizes to identify the optimal design for dual-animal fMRI of rats and mice. All coil models were designed and simulated using Sim4Life software (ZMT, Switzerland, <https://www.zmt.swiss>). Three combinations of rodent pairs were examined: rat–rat, mouse–mouse, and rat–mouse (Fig. 2). EM field simulations were conducted to compute the spatial distribution of the transmitted magnetic field ($|B_1|$) over the entire bodies of both animals, and a quantitative analysis of the field uniformity and intensity was

performed within the highlighted brain ROIs (Fig. 2). This study was conducted exclusively using 3D computational animal models and did not involve live animals or biological specimens. Therefore, approval from an institutional committee for animal care and use was not required. The anatomical models used in this study were obtained from a commercial provider and are specifically intended for use in EM simulations (IT’IS Foundation Virtual Population animal models).

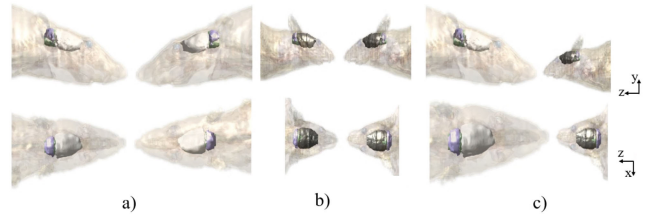


FIGURE 2. 3D models of the a) rat–rat, b) mouse–mouse, and c) rat–mouse combinations showing the selected brain tissues for ROI statistical analysis.

The inter-animal distance was fixed at 10 mm to maintain consistent coupling conditions. The rat model consisted of 77 tissues, representing a 567 g animal with a 260 mm body length (excluding the tail). The mouse model comprised 71 tissues, corresponding to a 44.7 g animal with a 98 mm body length (excluding the tail) [32], [33].

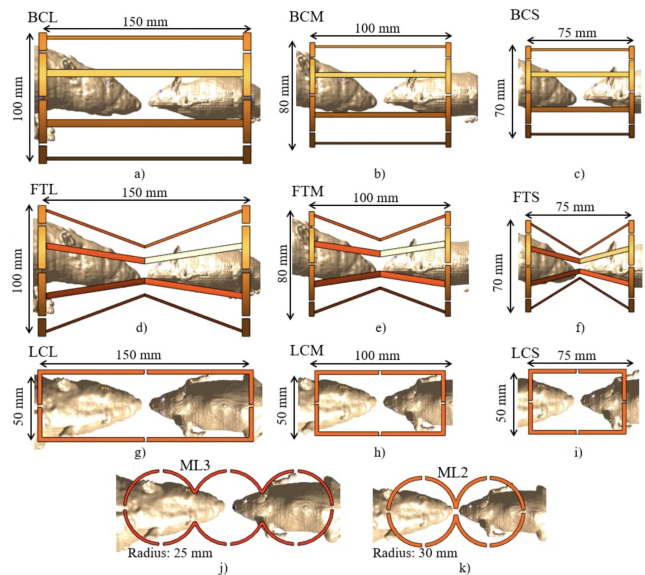


FIGURE 3. 3D models of the RF coils that were tested in this study. The birdcage coils: a) large (BCL), b) medium (BCM), and c) small (BCS); the proposed fish-trap coils: d) large (FTL), e) medium (FTM), and f) small (FTS); the loop coils: g) large (LCL), h) medium (LCM), and i) small (LCS); and the multi-turn coils: j) ML3 and k) ML2.

All RF coils were modeled and evaluated at 400 MHz, corresponding to the Larmor frequency in a 9.4-T MRI system. Each coil was tuned and matched to 50 Ω and excited using a single transmission channel, consistent with the hardware configuration of the scanner available at our institution.

The conductors were modeled as perfect electric conductors (PECs) to minimize the occurrence of simulation artifacts related to material losses.

Eleven coil types were analyzed, including high-pass BC coils of three sizes: large (BCL) with a 100 mm diameter and 150 mm length, medium (BCM) with an 80 mm diameter and 100 mm length, and small (BCS) with a 70 mm diameter and 75 mm length, as shown in Figs. 3a–c, respectively. Although conventional BC coils are widely used in small-animal MRI because of their strong and uniform central $|B_1|$ -fields, they may be suboptimal for the proposed setup, where the brain ROIs of the two rodents are positioned off-center. Thus, the central field uniformity of the BC coil is not effectively utilized. This study tested this hypothesis.

To address this limitation, a modified BC geometry was proposed, referred to as the FT coil (Figs. 3d–f) because of its resemblance to a traditional fishing trap. In this configuration, the rungs of the BC converge toward a narrower inner diameter at the center of the coil, bringing the conductors closer to each brain ROI, thereby enhancing the local $|B_1|$ -field intensity. Three FT variants were designed, similar to the BC designs: large (FTL) with a 100 mm outer diameter, 150 mm length, and 40 mm inner diameter; medium (FTM) with an 80 mm outer diameter, 100 mm length, and 40 mm inner diameter; and small (FTS) with a 70 mm outer diameter, 75 mm length, and 20 mm inner diameter. The closer proximity of the FT conductors to the animal heads was expected to yield a higher $|B_1|$ -field strength compared to the reference BC coils.

Because this study was conducted with a single transmission and reception channel, the evaluation of surface coil configurations was essential. These configurations are commonly used in such systems owing to their simplicity and high sensitivity near the surface. Three rectangular loop coils (LCs) were modeled: a large loop coil (LCL) with a length of 150 mm and a width of 50 mm, a medium loop coil (LCM) with a length of 100 mm and a width of 50 mm, and a small loop coil (LCS) with a length of 75 mm and a width of 50 mm.

In addition, a multi-turn coil (ML) configuration [34] was included. This design provides a broader coverage area while maintaining localized $|B_1|$ -field patterns. Two types of MLs were modeled: the first had three turns (ML3), with each loop having a radius of 25 mm to cover a total length of 150 mm and width of 50 mm, and the second had two turns (ML2), with each loop having a radius of 30 mm, spanning a total length of 120 mm.

Each coil configuration was tested for three experimental combinations: rat–rat, mouse–mouse, and rat–mouse (Fig. 2), with a fine spatial grid.

The $|B_1|$ -fields were computed and analyzed by calculating the mean value and coefficient of variation (CV), which is defined as the ratio between the standard deviation and mean $|B_1|$ magnitude. The ROI was confined to the rodent brain, restricting the statistical analysis to brain voxels and excluding non-target tissues.

In addition, the 3D complex vector fields E , D , J , and B were computed and extracted to create virtual phantoms and generate coil profiles for the MRI simulator, enabling simulation of the coil performance. These fields were cropped into a consistent spatial region with dimensions of $60 \times 60 \times 185$ mm across all cases and interpolated to match a matrix size of $512 \times 512 \times 512$. Subsequently, the processed field data were exported to MATLAB (R2023a, Natick, Massachusetts, The MathWorks Inc.) for further analysis and visualization.

B. VIRTUAL PHANTOM

For the MRI simulator, virtual phantoms were generated for each species combination (rat–rat, mouse–mouse, and rat–mouse). The simulator required the following voxel-wise input parameters: proton density (ρ), relaxation times (T_1 , T_2 , and T_2^*), and mass density (ρ_m). Moreover, the gyromagnetic ratio of protons (^1H) was considered in this study.

Tissue segmentation was performed using a lookup table comparing the conductivity and permittivity of each tissue [35], [36], [37], [38]. The conductivity (σ) was obtained from the ratio between the current density (J) and electric field (E):

$$\sigma = J/E \quad (1)$$

Similarly, the permittivity (ε) was computed as the ratio between the electric displacement field (D) and electric field (E):

$$\varepsilon = D/E \quad (2)$$

The voxel values of σ and ε were used in the EM simulations. The mass and proton density values obtained from the Sim4Life tissue database [38] were assigned using the tissue names in the lookup table. The relaxation time of each voxel was derived from the literature sources [39], [40], [41], [42], [43], [44], [45]. All tissue parameters (ρ , T_1 , T_2 , T_2^* , and ρ_m) are listed in Table 1. The data were organized in a cell structure to meet the input requirements of the MRI simulator.

C. MRI SIMULATOR

MRI image simulations were performed using MRiLab [30], a simulation platform capable of modeling RF coils, pulse sequences, gradient systems, main magnetic fields, and subject motion. In this study, three types of virtual phantoms (rat–rat, mouse–mouse, and rat–mouse) were evaluated to compare coil performance and signal behavior under identical conditions.

The MRI hardware configuration was defined with $B_0 = 9.4$ T, maximum gradient strength = 750 mT/m, and slew rate = 600 T/m/s. The selected pulse sequence was a gradient-recalled echo (GRE) sequence with the following parameters: flip angle = 8° , matrix size = 128×128 , field of view = 60×60 mm², bandwidth = 150 kHz, slice thickness = 2 mm, and TR/TE = 100/20 ms. The number of spins per voxel and the noise level were set to 10 and 1, respectively,

TABLE 1. Tissue properties for the MRiLab simulation.

Tissue	Mass density [kg/m ³]	Proton density [%]	T ₁ [s]	T ₂ [s]	T ₂ [*] [s]
Blood	1049.7	80.4	2200.0	45.0	25
Bones	1028.5	40.8	300.0	1.5	0.5
Brain (Whole)	1045.5	73.8	2000.0	40.0	25.0
Grey matter	1044.5	81.5	2100.0	45.0	28.0
White matter	1041.0	69.4	1500.0	30.0	20.0
Cartilage	1099.5	71.7	1600.0	25.0	10.0
Cerebellum	1045.0	78.5	1900.0	45.0	27.0
CSF	1007.0	99.3	4800.0	2000.0	200.0
Eye	1090.4	76.2	4000.0	1000.0	200.0
Fat	911.0	24.4	350.0	70.0	60.0
Midbrain	1045.5	72.3	1800.0	35.0	25.0
Muscle	1090.4	76.2	1800.0	25.0	15.0
Nerve	1075.0	78.0	1600.0	40.0	20.0
Skin	1109.0	65.1	1700.0	40.0	20.0
Tongue	1090.4	66.0	1500.0	30.0	15.0

to simulate realistic acquisition conditions. Image analysis was performed in the image domain to evaluate the SNR and visualize the qualitative contrast achieved for each coil type and species combination. The coil profiles were computed by adding the B and E fields from each of the simulated coils (Fig. 3), which were used as the transmission and reception coils. Images were then obtained for each coil and animal configuration.

III. RESULTS

The tuning and matching condition of each of the coils are plotted in Figure 4. For each animal configuration (rat–rat, mouse–mouse, and rat–mouse), the coils were independently tuned and matched at the operating frequency. The corresponding S_{11} responses were found to be very similar among the different loading conditions; therefore, for simplicity, only a representative tuning and matching profile is shown in Figure 4. As a result, the reported $|B_1|$ -field comparisons are not influenced by loading-induced mismatch effects and primarily reflect the intrinsic performance of the different coil geometries. The $|B_1|$ -field distributions for each coil type and animal configuration (rat–rat, mouse–mouse, and rat–mouse) were computed, and the mean $|B_1|$ -field within each brain ROI was evaluated. Fig. 5 shows $|B_1|$ -field maps for the rat–rat configuration with a 10 mm inter-animal distance, where the brain regions are highlighted for better visualization of the $|B_1|$ -field profile. Similarly, Figs. 6 and 7 show the field distributions for the mouse–mouse and rat–mouse combinations, respectively.

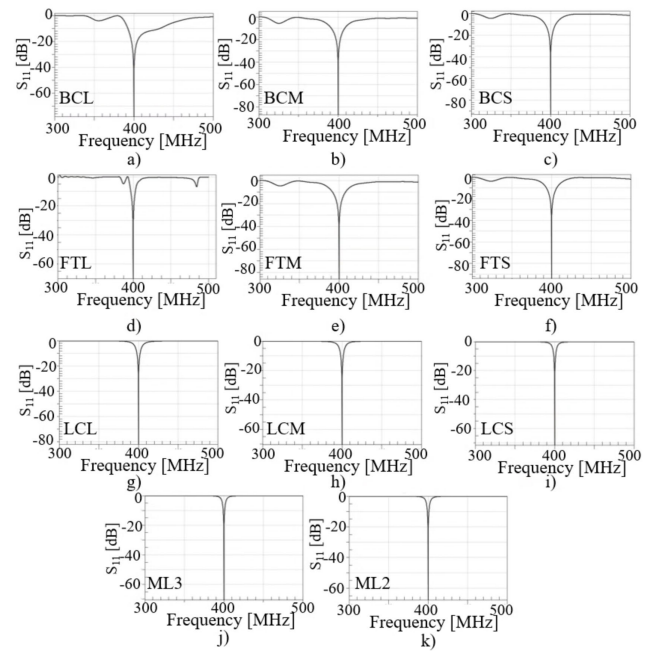


FIGURE 4. The tuning and matching given by the S_{11} profiles for the a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i), small (LCS) loop coils, and j) ML3 and k) ML2 coils.

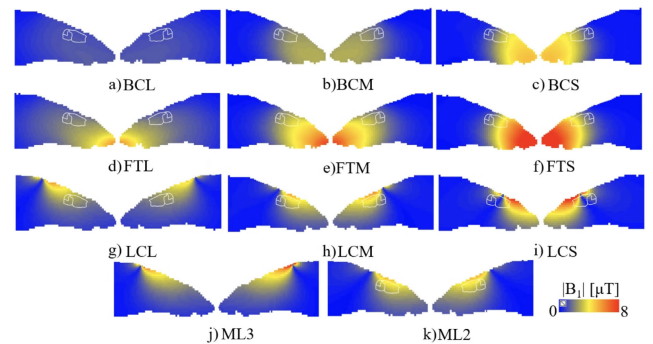


FIGURE 5. The $|B_1|$ fields computed for the rat–rat combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i), small (LCS) loop coils, and j) ML3 and k) ML2 coils.

The results indicated that the conventional BC coils produced a strong $|B_1|$ -field concentrated near the center of the coils. However, because the target ROIs were located in two separate regions (the two brains), the overall field uniformity across both regions was reduced, particularly for the BCS and BCM coils. Although the BCL coil provided improved field uniformity, it exhibited lower $|B_1|$ -field intensity. In the mouse–mouse configuration, the smaller head size and closer proximity of the brains to the center of the coil resulted in higher and more uniform $|B_1|$ -fields compared to those of the rat–rat setup. Among the tested coils, the FTS coil provided the most favorable $|B_1|$ -field profile.

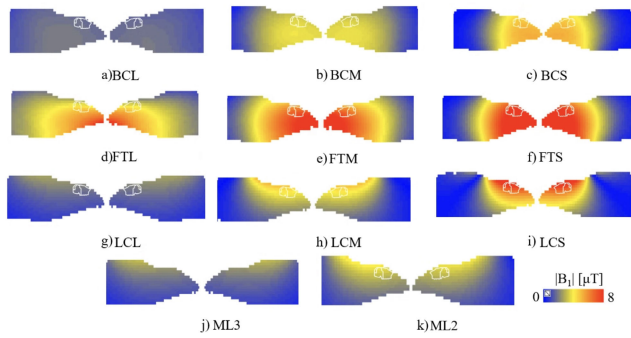


FIGURE 6. The $|B_1|$ fields computed for the mouse-mouse combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i), small (LCS) loop coils, and j) ML3 and k) ML2 coils.

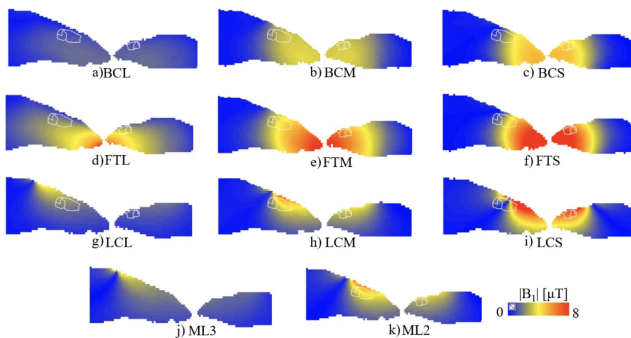


FIGURE 7. The $|B_1|$ fields computed for the rat-mouse combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i), small (LCS) loop coils, and j) ML3 and k) ML2 coils.

For the rat-mouse configuration, the $|B_1|$ -field distributions were asymmetric, as expected, because of the differences in animal size and electromagnetic loading. In most coil configurations, the mouse ROI exhibited a relatively uniform $|B_1|$ -field, whereas the rat ROI showed reduced field uniformity. Notably, the BCM coil achieved similar $|B_1|$ -field intensity and uniformity in both ROIs, suggesting a more balanced field distribution across the two animals. To quantitatively confirm these observations, a statistical analysis was performed for all coil types and configurations.

Fig. 8 shows a statistical comparison of coil performance. For the rat-rat configuration, the BCM coil produced the strongest mean $|B_1|$ -field. In the mouse-mouse setup, the FTM coil yielded the strongest $|B_1|$ intensity, which could be attributed to the proximity of their legs to both the coil center and mouse brain. In the rat-mouse configuration, the ML3 coil provided the strongest $|B_1|$ field in the rat brain, followed closely by LCM. For the mouse in the same setup, the FTS coil provided the strongest $|B_1|$ -field. However, because achieving similar $|B_1|$ levels in both species is desirable, LCM was concluded to offer the best balance, providing comparable field strengths in both the rat and mouse ROIs. Fig. 8g shows the maximum specific absorption rate average with

10 grams (SAR_{10g}) values computed for each coil and for the different rat-mouse combinations. The peak SAR_{10g} values were primarily observed in the paws and ears, particularly in regions close to the coils.

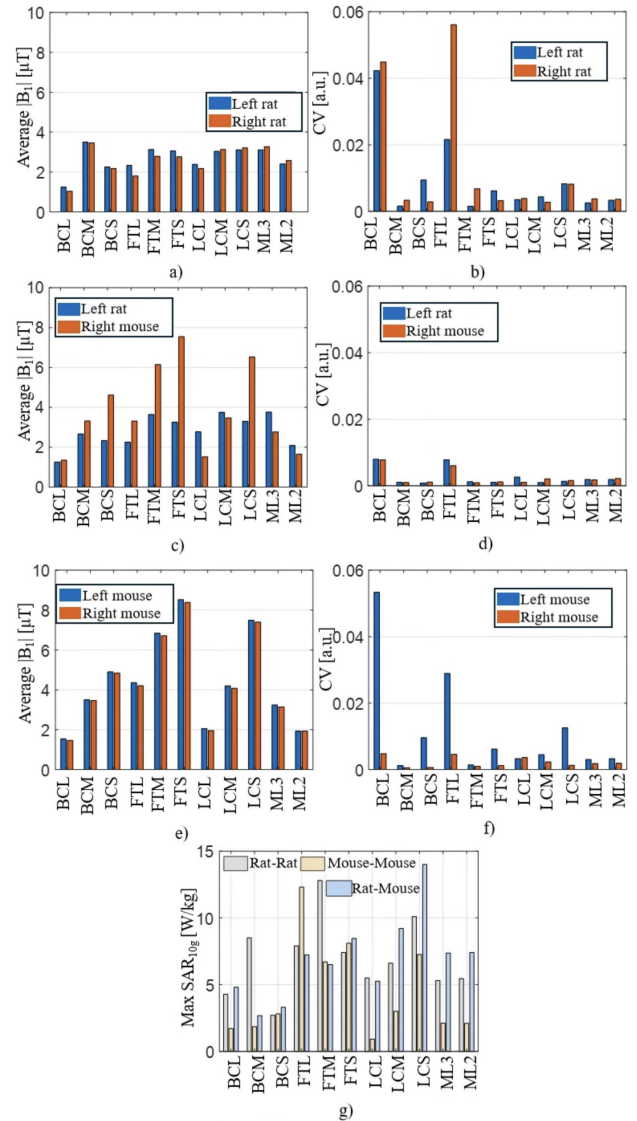


FIGURE 8. Statistics of the $|B_1|$ -field within the selected ROIs for the different animal configurations: (a) mean $|B_1|$ for rat-rat, (b) CV for rat-rat, (c) mean $|B_1|$ for mouse-mouse, (d) CV for mouse-mouse, (e) mean $|B_1|$ for rat-mouse, and (f) CV for rat-mouse. (g) Maximum SAR_{10g} values for each coil design, reported for the corresponding animal species and configuration.

In addition to $|B_1|$ -field computations, the MRiLab simulations provided valuable insights into selecting the optimal coils for imaging, enabling the evaluation of the effects of relaxation times, pulse sequence parameters, and, especially, the SNR within each animal's brain ROI.

Fig. 9 shows the axial MRI images of the rat brain simulated for each coil, with an ideal uniform reference coil included for comparison. The ML2 coil provided the highest image signal, whereas the BCM coil was predicted to have

the highest field intensity based on the simulated $|B_1|$ -field. Notably, all the proposed coils generated a higher signal intensity than the ideal uniform coil, which yielded a lower overall signal, although it was perfectly homogeneous. These findings suggest that for rat–rat studies, LCM, ML3, and ML2 provide the best imaging performance.

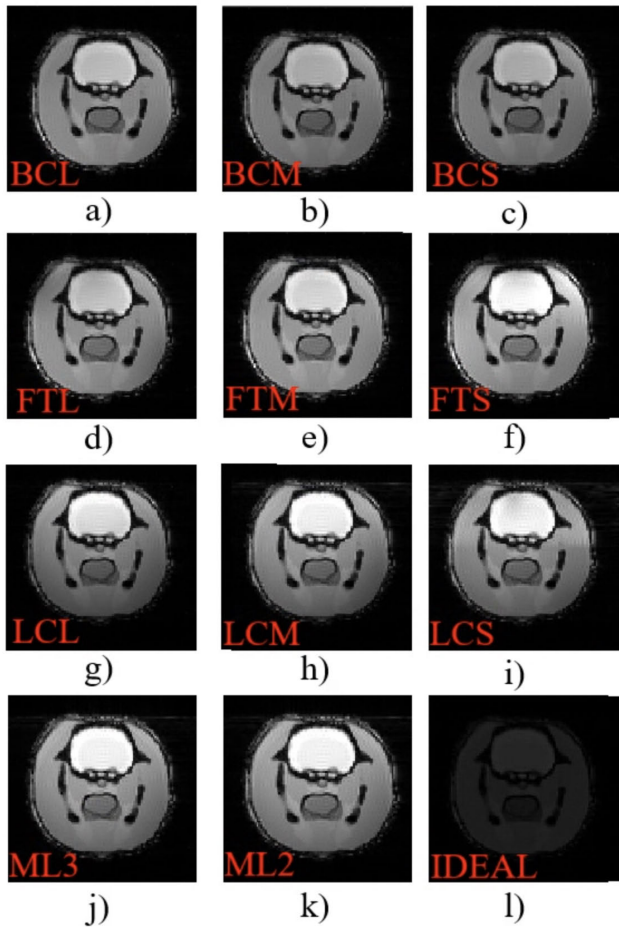


FIGURE 9. The simulated GRE images for the rat–rat combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, and the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i) small (LCS) loop coils, j) ML3 and k) ML2 coils, and l) an ideal uniform coil.

Similarly, the mouse–mouse GRE simulations (Fig. 10) showed that most coils achieved high and comparable image intensities within the brain ROIs, owing to the smaller anatomy. However, the volumetric coils, that is, BCM, BCS, FTL, FTM, and FTS, demonstrated superior field uniformity. Conversely, BCL and LCS produced the lowest image intensities, indicating reduced efficiency for mouse-only imaging. Thus, volumetric coil designs are preferable for mouse–mouse studies.

Finally, the main focus of this study, that is, the rat–mouse face-to-face configuration, is illustrated in Fig. 11, which compares the image intensity of each brain obtained using the different coils. The corresponding mean signal intensities in the two ROIs are plotted in Fig. 12, which shows that

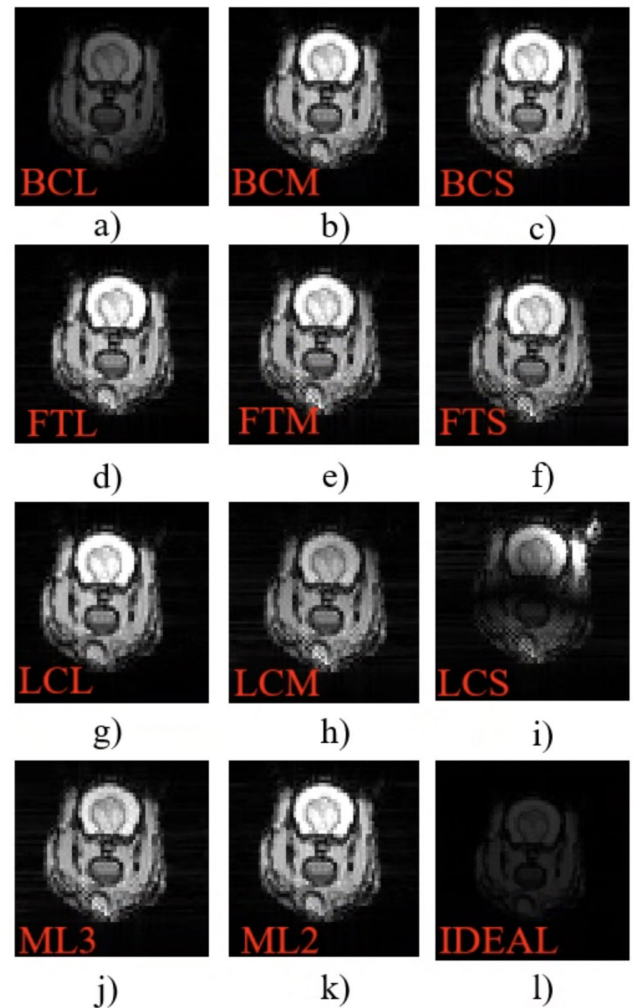


FIGURE 10. The simulated GRE images for the mouse–mouse combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i) small (LCS) loop coils, j) ML3 and k) ML2 coils, and l) an ideal uniform coil.

LCM produced similar image intensities in both species while maintaining high signal levels overall. Although the ideal uniform coil yielded a lower absolute intensity, it minimized interspecies differences, reinforcing the tradeoff between uniformity and signal strength.

IV. DISCUSSION

This study proposes an integrated simulation-based framework for RF coil design and selection tailored to face-to-face dual-animal MRI experiments, including interspecies pairings. By coupling EM modeling with MRI signal simulations, this approach enables quantitative comparisons of candidate geometries under pseudo-realistic loading and spatial constraints. The findings indicate that a simulation-guided pre-evaluation can effectively narrow the design space, predict the relative $|B_1|$ performance prior to physical fabrication

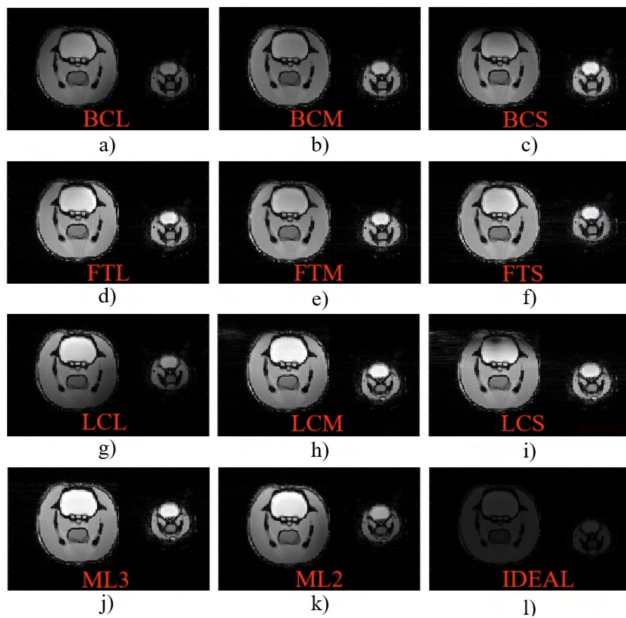


FIGURE 11. The simulated GRE images for the rat-mouse combination with a) large (BCL), b) medium (BCM), and c) small (BCS) birdcage coils, the proposed d) large (FTL), e) medium (FTM), and f) small (FTS) fish-trap coils, g) large (LCL), h) medium (LCM), and i) small (LCS) loop coils, j) ML3 and k) ML2 coils, and l) an ideal uniform coil.

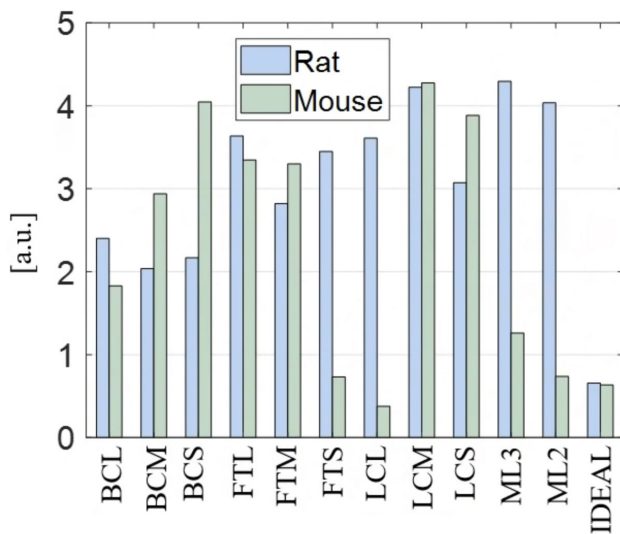


FIGURE 12. The mean signal intensity was computed on the brains of the rat and mouse for each of the coils based on the simulated MRI.

and in vivo testing, and provide a principal basis for selecting coil designs matched to specific dyadic arrangements.

A. GEOMETRY-BY-PAIRING DEPENDENCY OF $|B_1|$ PERFORMANCE

A key observation of this study was that the coil performance depended on the interaction between the geometry and anatomical scale of the paired animals rather than on the coil type alone. For rat-rat imaging, the BCM design achieved the best balance between the $|B_1|$ -field intensity and spatial uniformity, which is consistent with the volumetric efficiency

of BC structures over large volumes. For mouse-mouse imaging, the FTS coil yielded the highest $|B_1|$ -field strength while maintaining improved homogeneity within the brain ROI. This can be attributed to its convergent-rung geometry, which reduces the conductor-to-brain distance. For rat-mouse imaging, LCM provided the most balanced field distribution across both brains, despite pronounced differences in head size and loading asymmetry. Collectively, these findings demonstrate that no single coil geometry is uniformly optimal across all pairings, and coil selection should be conditioned on both animal combination and the intended face-to-face arrangement.

B. BALANCING SENSITIVITY AND COMPARABILITY IN DYADIC IMAGING

Two recurring tradeoffs emerged across the evaluated coil geometries. First, volumetric coils generally provided more spatially uniform $|B_1|$ -field distributions but exhibited reduced efficiency when the ROIs were located away from the region of maximum field homogeneity. In contrast, surface-type coils increased local $|B_1|$ -field intensity at the expense of greater spatial nonuniformity and reduced coverage. Second, interspecies imaging introduced a trade-off between within-brain sensitivity and cross-brain balance. Maximizing the $|B_1|$ -field in one brain often increases the imbalance between the two brains, which may complicate the interpretation of dyadic analyses requiring comparable activation amplitudes and spatial extents across animals. Conversely, coil designs that produced more similar $|B_1|$ -field amplitudes in both brains may facilitate more direct comparisons of functional activation patterns, even if neither brain achieved the maximum possible signal intensity. Therefore, the optimal coil design depends on the intended application. Coil geometries that promote balanced $|B_1|$ -field distributions between animals may be preferable for inter-animal comparative studies, whereas designs that maximize local sensitivity may be more suitable when signal detectability within a single species is the primary objective.

The values reported in this work are slightly higher than those reported in related work [46], which investigated the $|B_1|$ -field in four small mice arranged in a different configuration. However, the simulated $|B_1|$ -field patterns remain comparable, and the corresponding MRI images show similar signal contrast that the simulated images.

C. SNR-DRIVEN SENSITIVITY IN DYADIC fMRI

In addition to $|B_1|$ -field strength and uniformity, the SNR is a key determinant of fMRI sensitivity. This is particularly relevant for the high-resolution imaging of animal brains, where the SNR can translate directly into a temporal SNR (tSNR) [47]. The tSNR reflects the stability of the voxel-wise time series used for BOLD detection and connectivity analyses. Improvements in excitation fidelity (for example, flip-angle accuracy and excitation uniformity) and reception sensitivity can increase the signal amplitude

and reduce relative noise contributions, thereby increasing the tSNR and enhancing the statistical detection power for subtle BOLD effects. These gains are particularly important in dual-animal and off-isocenter arrangements, where reduced coil efficiency, asymmetric loading, and spatially varying sensitivity can degrade the SNR unevenly across brains, leading to biased comparisons. Thus, coil designs that improve the SNR while maintaining adequate cross-brain balance are expected to yield more reliable dyadic fMRI readouts and improve the reproducibility of interaction-related effects.

D. IMPLICATIONS OF THE FISH-TRAP AND LOOP DESIGNS

The FT geometry provides a structure-level strategy for improving the $|B_1|$ efficiency of dual-ROI arrangements. This design increases the local field strength in the paired brain region by converging rungs toward the center of the coil, partially compensating for the underutilized transmission uniformity of a conventional BC coil owing to off-center ROI placement. By contrast, medium-loop designs exhibit robust performance under asymmetric loading, suggesting their practical utility for rat–mouse pairings or experimental conditions with modest variability in head positioning. Together, these observations support the use of geometry-driven modifications to tailor the coil performance to dyadic constraints, rather than relying solely on a single conventional coil type scaled across all pairings.

E. SIMULATION-GUIDED COIL OPTIMIZATION

The proposed simulation workflow supports the virtual assessment of coil–subject interactions, including changes in loading-dependent performance and sensitivity to positioning, thereby reducing experimental trial-and-error. This is particularly relevant for preclinical systems with single-transmit-channel architectures, where geometric optimization of the transmission coil can improve the excitation efficiency and maintain balanced performance across off-center brain ROIs. Importantly, integrating EM simulation with MRI signal simulation provides an evaluation pathway that is more directly linked to the expected imaging outcomes than using $|B_1|$ metrics alone, enabling the interpretation of coil design choices in terms of predicted image uniformity and sequence-relevant sensitivity.

F. LIMITATIONS AND FUTURE DIRECTIONS

This study has several limitations that should be considered. First, conductors were modeled as ideal PECs, neglecting the resistive losses and frequency-dependent effects that reduce efficiency in real-world scenarios. Second, the tissue properties were simplified and assumed to be isotropic, which neglects the spatial heterogeneity that can influence high-field RF behavior. While these assumptions simplify the model, they may affect the absolute values of $|B_1|$ -field and SAR by neglecting conductor losses, fabrication-related imperfections, and tissue heterogeneity. As a result, the reported quantitative values should be interpreted as approximate rather than exact predictions of in vivo performance.

However, because the same modeling assumptions were applied consistently to all coil configurations, the framework remains appropriate for comparative assessment of relative performance trends across designs. Third, the MRI signal simulation was limited to a gradient-echo model. While this approach provides a simplified and controlled setting for comparing relative RF coil performance across different designs and subject combinations, it does not fully represent the acquisition conditions typically used in fMRI, where EPI-based sequences are more common. In particular, sequence-specific effects associated with EPI, such as distortion sensitivity, susceptibility-related artifacts, and differing signal behavior, are not captured in the current framework. Therefore, the results should not be interpreted as fully generalizable to all fMRI acquisition schemes. Rather, the present co-simulation environment is intended primarily for comparative evaluation of coil performance under a consistent imaging setup, which remains useful for relative hardware assessment. Fourth, this evaluation primarily emphasized the transmission field characteristics. Incorporating reception sensitivity, noise correlation, and realistic preamplifier/coil losses would strengthen the predictions of the SNR and performance differences among multi-element configurations. The framework is intended primarily for comparative evaluation of RF coil designs under a consistent simulation setting, so the reported values should be interpreted as relative performance estimates rather than exact predictors of in vivo behavior. Therefore, experimental validation is essential. The construction of the selected coils, bench characterization, and in vivo $|B_1|$ mapping in both species are required to verify the predictive accuracy of the simulations and calibrate the model parameters. Finally, although multi-channel transmission and reception arrays may further improve homogeneity and sensitivity, their benefits must be weighed against hardware complexity and the constraints of commonly available preclinical platforms. Overall, the presented simulation-guided framework provides a practical approach for selecting coil geometries that enables robust face-to-face dual-subject fMRI, including challenging interspecies arrangements.

V. CONCLUSION

This study demonstrated a co-simulation framework for optimizing RF coil geometries specifically tailored to dual-species face-to-face fMRI. The proposed approach provides quantitative design guidance, identifying BCM, FTS, and LCM as the optimal choices for the rat–rat, mouse–mouse, and rat–mouse configurations, respectively, and enables the systematic pre-evaluation of coil performance under experimental constraints. This framework offers a valid basis for coil selection and experimental preparation in dual-animal fMRI, particularly for face-to-face configurations, and can be directly applied to improve signal reliability and reproducibility in future preclinical studies.

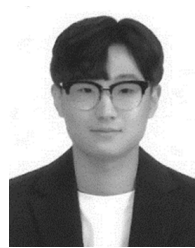
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