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Modeling Endovascular MRI Coil Coupling with Transmit RF Excitation

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Abstract

Objective—To model inductive coupling of endovascular coils with transmit RF excitation for selecting coils for MRI-guided interventions.

Methods—Independent and computationally efficient FEM models are developed for the endovascular coil, cable, transmit excitation and imaging domain. Electromagnetic and circuit solvers are coupled to simulate net B_1+ fields and induced currents and voltages. Our models are validated using the Bloch Siegert B_1+ mapping sequence for a series-tuned multimode coil, capable of tracking, wireless visualization and high resolution endovascular imaging.

Results—Validation shows good agreement at 24, 28 and 34 μ T background RF excitation within experimental limitations. Quantitative coil performance metrics agree with simulation. A parametric study demonstrates trade off in coil performance metrics when varying number of coil turns. Tracking, imaging and wireless marker multimode coil features and their integration is demonstrated in a pig study.

Conclusion—Developed models for the multimode coil were successfully validated. Modeling for geometric optimization and coil selection serves as a precursor to time-consuming and

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expensive experiments. Specific applications demonstrated include parametric optimization, coil selection for a cardiac intervention and an animal imaging experiment.

Significance—Our modular, adaptable and computationally efficient modeling approach enables rapid comparison, selection and optimization of inductively-coupled coils for MRI-guided interventions.

Index Terms

Endovascular MRI coils; Wirelessly-coupled coils; B_1+ field mapping; FEM simulation; MRI coil coupling

I. Introduction

Innovative endovascular coil designs are needed to address challenges of MRI-guided interventions under motion (cardiac, respiratory) and blood flow. Both passive techniques [1], [2] and active miniature coils [3]–[6] have been used for tracking and wireless visualization. Coils for high-resolution intravascular imaging are used in [7]–[9]. Novel designs like the meanderline [10], orthogonal solenoid [11] and internal birdcage [12] provided specific features like forward-looking capability and curvature tracking. The loopless design [13], [14] showed improvements in imaging coverage. Multifunctional coils [15]–[18] capable of some combination of tracking, wireless visualization, high-resolution intravascular imaging, transmit profile tailoring or curvature delineation, offered increased flexibility but with associated increase in complexity. Strategies such as current monitoring [19], phase dithering [6], [20] and reverse polarization [21]–[24] have been used to improve performance.

Choice of coil geometry must consider several interdependent factors, including excitation scheme, reconstruction, coil geometry, fabrication, inductive coil coupling and the influence of the cable in imaging performance and safety. Modeling can play an important role in coil optimization and selection by minimizing cost, time and effort in fabrication and testing.

While modeling receive coil performance is quite straightforward using Biot-Savart law analysis and reciprocity [8], modeling inductive coupling of resonant endovascular coils is more challenging. Modeling complexity results from the need to consider coil orientation, tuning and matching lumped elements, transmit RF excitation, effect of the connecting cable, and electrical properties of the imaged sample. This leads to several requirements: 1) Circuit and electromagnetic (EM) simulators need to be coupled. 2) RF excitation modeling needs to cover the large external coil volume, while simultaneously requiring fine meshing of the miniature endovascular coil geometrical features. 3) A long, lossy microcoaxial cable connecting to the receiver chain has to be modeled. These conflicting demands necessitate optimization for a computationally efficient simulation.

Here we develop generalized and comprehensive Finite-Element-Method (FEM) [25] full-wave simulation models of inductive coupling between an endovascular coil geometry and transmit RF excitation during MRI. The modeling strategy allows independent specification of system components, including RF excitation schemes, electrical and geometrical coil

features, cable models and tissue properties. Our model is validated using a 2D Bloch-Siegert imaging sequence for the multimode coil [18]. We show optimization of coil parameters against defined performance metrics and demonstrate utility in an MRI-guided cardiac intervention. Preliminary versions of this work have been reported [26]–[29].

II. Theory

For endovascular coils during MRI, the net RF excitation field is influenced by three components: 1) homogeneous RF excitation ($B_{1,tr}$) from a transmit coil; 2) cancellation of field inside the coil conductors ($B_{1,sc}$); and 3) field due to inductively-coupled RF currents ($B_{1,indcp}$)

$$B_{1,net} = B_{1,tr} + B_{1,sc} + B_{1,indcp} \quad (1)$$

$$B_{1,indcp} = f(I_{induced}) \quad (2)$$

$$V_{induced} = \left(-\frac{d\phi}{dt} \right), \quad \phi = \int B_{1,tr} dA \quad (3)$$

f is a function of coil geometry, $V_{induced}$ and $I_{induced}$ are the induced voltage and current in the coil, ϕ is the magnetic flux through the coil, and A is the area of the loop. For loopless coils [13], induced currents are small due to absence of low-impedance paths. For series-tuned resonant coils, inductively-coupled current is amplified by the quality factor (Q), and so $B_{1,indcp}$ field distortion is significant. We use full-wave FEM models to estimate the net RF magnetic field (B_1) during transmit. FEM is a good option for resonant high-Q coils with fine features as a frequency domain technique with tetrahedral volume meshing.

III. Methods

A. Fabricated Coils

A multimode coil (Fig 1a) was built on a 6 mm acrylic tube using 36 AWG wire. A 6-turn, series-tuned, tip-tracking solenoid was connected in series with a 3-turn, series-tuned, 3 cm long rectangular imaging loop. The coil was made resonant at 63.86 MHz using four 33 pF capacitors connected in series with the coil and matched using a quarter-wave π -network to a custom-made 50 Ω , 42 AWG microcoaxial cable (New England Wire Tech). Heat shrink tubing was used to insulate coil conductors. Coil impedance at each fabrication stage was recorded using the 8753C HP Network Analyzer. Coil and cable losses were measured under loaded conditions.

B. Endovascular Coil Models

FEM models were developed using COMSOL Multiphysics 4.1.2 (Fig 1b). Impedance at each stage was computed by S-parameter simulation with lumped capacitors for tuning. Wires were modeled using copper. Loaded coil (R_1) and cable losses (R_2) were empirically modeled using lumped elements based on values recorded during fabrication. Lumped element approximation for coil and cable losses significantly reduces model complexity. The rectangular imaging coil was modeled as a planar coil for simplicity. A finer mesh was used for coil elements and the surrounding region. A cylinder enclosing the coil modeled heat shrink tubing.

C. Modeling of Birdcage Coil Excitation

An idealized quadrature birdcage model was used to simulate the RF transmit excitation (Fig 2). This reduces overall computational time and resources. RF excitation was modeled as two sinusoidal currents on the surface of a large cylinder in time and space quadrature. Space quadrature was achieved by spatially shifting the two current densities by 90° in the x - y plane. Time quadrature was obtained by phasor representation of currents (Fig 2) in frequency domain simulations (4). x , y and z represent Cartesian coordinates on the cylinder ($\hat{z}$ is the unit vector). The amplitude A was calibrated empirically to match the excitation B_1+ field.

$$\mathbf{J}_{surface} = A \sin \left[\arctan \left(\frac{y}{x} \right) \right] + j A \sin \left[\arctan \left(\frac{y}{x} \right) - \frac{\pi}{2} \right] \hat{z} \quad (4)$$

This represents a good and computationally efficient approximation of the transmit RF excitation during MRI as it overcomes the need for fine meshing to model coil elements of a full birdcage model.

D. Modeling of Simulation Domain

The region outside the heat shrink tube was modeled as a homogenous, medium having the conductivity of a 0.9% NaCl and 0.3% CuSO₄ solution. The conductivity was calculated using the transmission line method [30]. A syringe, whose narrow end had been cut off and replaced by an extra plunger, was filled with the solution. Two copper electrodes were placed in contact with the solution. The resistance was measured using a vector impedance meter at different distances between the two electrodes to eliminate contact resistances. The conductivity was determined to be 1.82 S/m.

E. Bloch Siegert B_1+ Maps

A 2D SPGR Bloch-Siegert sequence [31], [32] was used to obtain B_1+ maps on a 1.5 T scanner (Signa HDx, GE Healthcare, Waukesha, WI). The endovascular coil was tightly fitted into 7 mm grooves in three 8 cm long, 9.5 mm diameter acrylic dowels. These were mounted on a leveled acrylic sheet (thickness = 1.32 cm) and placed in a rectangular tub of 31 cm $\times$ 18.3 cm $\times$ 13.5 cm. The tub was completely filled with a 0.9% NaCl, 0.3% CuSO₄ solution.

The birdcage body coil was used for transmit and a standard 8-channel cardiac phased array was used for receive. A 3-plane localizer was used to prescribe oblique slices. The imaging coil conductors were placed in-plane with the main slice by acquiring additional B_1+ maps for slices immediately above and below and iteratively adjusting the tilt in two orthogonal planes. The scan parameters were: FOV = 26 cm $\times$ 26 cm, TR = 33 ms, TE = 20.7 ms, NEX = 16, acquisition matrix = 512 $\times$ 512, slice thickness = 4 mm, excitation flip angle = 30° with an additional Fermi-shaped off-resonance pulse at $\omega = \pm 4000$ Hz and flip angle $\theta = 400^\circ$ inserted between the excitation pulse and readout. Images were optimized by trading off between SNR, resolution, partial volume effects and scan time. Final scan time was 19 minutes.

F. FEM Simulations

Models for various system components were incorporated into the final FEM simulation. The simulation model excitation (birdcage coil current density amplitude A) was calibrated empirically by averaging field values at a set of five points inside the phantom away from the endovascular coil, in the experimental B_1+ maps. The electromagnetic FEM solver and circuit solver were fully coupled to simultaneously solve for electric and magnetic fields and circuit voltages and currents. The magnitude of the B_1+ field, given by (5), was validated for several values of transmit gain (adjusted manually), of which results for three values are shown here.

$$|B_1| = \text{magnitude}(B_{1x} - jB_{1y}) \quad (5)$$

G. Pig study: Real-time cardiac MRI

We used our validated simulation models to select optimal multimode coil geometry for animal imaging studies. The aim was to track the tip of a 6 F catheter along the inferior vena cava of a sedated pig into the right ventricle (RV) of the heart, and subsequently to view the device and surrounding anatomy. Pig hearts and vasculature are similar in size and geometry to adult human hearts, and are therefore good test cases. All animal experiments were carried out in compliance with NIH guidelines for the use of laboratory animals.

Our simulations indicated that a series resonant 6-turn Helmholtz pair tracking coil in series with a 3-turn rectangular imaging loop on the 6 F catheter was the optimal configuration. The body coil was used for transmit and a modified 4-channel cardiac phased array for receive. Fast GRE imaging sequences were optimized iteratively to optimally position and capture images for the different coil functions. Real-time MRI-guidance system based on the RTHawk [33] and Vurtigo platform were used for tracking, imaging and visualization. Tip tracking was performed using a Hadamard encoding and interleaved with 3-plane 2D Cartesian imaging. After entering the RV, we obtained roadmap images at different RF transmit gains to obtain complementary information of the heart and the multimode coil as a wireless marker. Finally to demonstrate endovascular imaging functionality, we obtained a small FOV, high resolution, *ex-vivo* image of a cavity using the same endovascular coil as an MR receiver.

IV. Results

A. Endovascular Coil FEM Model

Table I confirms consistency of the FEM multimode coil model, within fabrication limitations and the planar coil approximation. This proves validity of the coupled EM and circuit model for the multimode coil itself.

B. Simulation Model Validation

Simulated and Bloch Siebert B_1+ magnitude maps in Fig 3, 4 and 5 show good agreement, given limitations (Table II). Distortion of the homogeneous transmit field profile ($B_{1,tr}$) due to inductive coupling ($B_{1,indcp}$) increases at higher background excitation. The tracking solenoid produces dipole-like field addition and subtraction at opposite ends (solid black arrow, Fig 3b). There is field cancellation inside the rectangular imaging loop and field amplification radially on either side. (dashed white arrow, Fig 3b). These effects are captured in both simulation and experiment. Limitations in experiment and simulation are summarized in Table II. Primarily, experimental images show significant signal averaging across a 4 mm slice with in-plane resolution of $0.5 \text{ mm} \times 0.5 \text{ mm}$ (wire diameter = 0.1 mm), whereas simulation shows limitations of numerical computation in regions of large field discontinuity close to the imaging coil.

Fig 6 compares B_1+ magnitude profiles from Fig. 3a and 3b: a) along the radial axis of the tracking solenoid; b) along the radial axis through the center of the imaging loop. There is good agreement, except for short T_2^* effects inside the acrylic tube and at a single pixel (at the interface of the coil conductors), significantly impacted by numerical computation limitations and signal averaging (indicated by finger, Fig 6b). Nevertheless, normalized performance metrics (that directly depend on this pixel value) can be predicted from simulation (Section IV-C) as the coil-medium discontinuity in material properties is a systematic error. Numerical errors may be reduced by finer meshes and tighter solver tolerance criteria at the cost of more computational time and resources.

Finally, we point out the appearance of the overflipping artifact in the Bloch Siebert B_1+ maps (zoomed inset). Practically, this phenomenon is useful in suppressing signal from near the coil to view the surrounding tissue more clearly.

C. Endovascular Coil Performance Metrics

We define two figures of merit (FoM) to characterize endovascular coil performance in (6) and (7). FoM1 defines the relative ease of locating the device tip. FoM2 is the radial coverage of the imaging coil. The experimental values of these FoMs are indicated in Fig 6.

$$FoM1 = \frac{\text{Tracking coil peak magnitude}}{\text{Imaging coil peak magnitude}} \quad (6)$$

$$FoM2 = \text{Imaging Coverage } (B_1 \text{ distortion} > 5\%) \quad (7)$$

The two FoMs are plotted against the number of turns of the tracking coil (N) in Fig 7. Coil behavior can be classified into three distinct regions. In region I ($N < 6$), the imaging coil dominates overall performance and so $FoM1 \approx 1$. We note the agreement between simulation and measurement of the two FoMs at $N = 6$. In region II, both tracking and imaging coil influence overall performance. There is a sharp decrease in $FoM1$, driven by increased resistance with additional tracking coil turns. $FoM2$ continues to rise while the imaging coil is dominant and drops thereafter. In region III, the tracking coil dominates performance. As N increases, tracking coil inductive coupling increases (which corresponds to high localized MR flip angles when imaging), resulting in $FoM1$ increasing consistently, going above unity at $N = 18$.

D. Analysis of Model Approximations

We analyzed our model error bound due to approximations in Table III. The overall error is dominated by the ideal birdcage coil approximation. Further model refinement is best focused on better modeling the transmit RF excitation.

E. Endovascular Coils for Real-time Interventional MRI

Fig 8 shows the multimode coil in tip-tracking mode, in the inferior vena cava (white arrows). The catheter was inserted through the femoral artery and navigated to the position shown. Fig 8a (zoomed), 8b and 8c show tracking (red dot) of the catheter tip in the inferior vena cava. Fig 8d and 8e show the tip-tracking coil signal peaks in different planes, corresponding to Fig 8a. We were able to successfully tip-track and guide the catheter through the inferior vena cava into the right ventricle (RV) of the pig heart.

Fig 9 demonstrates wireless marker capability inside the RV. The multimode coil signal is clearly visible on the dynamically updated roadmap image obtained with the external coil and enables visualization of a segment of the catheter distal end, displaying its orientation. Using two different transmit gain settings, we were able to obtain different image contrast. Fig 9a (lower transmit gain) shows the coil position, while Fig 9b (higher transmit gain) shows the surrounding tissue in more detail.

Fig 10b shows the multimode coil in the imaging mode. Comparison between Fig 10a (external phased array coil image) and Fig 10b demonstrates the SNR improvement obtained while imaging a cavity in an *ex-vivo* pig heart with the multimode coil. The larger SNR and higher achievable imaging resolution using the multimode coil (Fig 10b) demonstrates the value of an endovascular imaging coil, as opposed to using only an external coil. We also note in Fig 10b that the imaging FOV is limited to a small region, demonstrating why roadmap images using an external coil (which can be enhanced by tracking and wireless marker functions) is also required, in combination with endovascular coil imaging.

In an MRI-guided procedure, the multimode coil can be used in active tracking and inductively-coupled wireless marker modes to reach a target and obtain overall contextual

visualization. Subsequently, the high resolution and SNR features of the imaging mode are more useful to look at finer details along a vessel or cavity wall or treated tissue. The demonstrated multi-functional capability is facilitated by a combination of imaging and tracking coils into a design that also couples inductively with the transmit RF excitation. Therefore, FEM simulation models have an important role to play in coil optimization to obtain features needed for real-time MRI-guided interventions.

V. Discussion

Modular FEM models to simulate inductively coupled B_1+ field distortion were successfully validated using the Bloch Siebert B_1+ mapping sequence. A parametric study showed the impact of geometrical design and available performance tradeoffs. Simulation accurately predicted defined performance metrics. Finally, we selected an optimized coil to perform experiments in a pig and successfully demonstrated its performance in real-time, interventional, cardiac MRI. Simulation-based coil selection and optimization can help reduce fabrication and experimental costs, time and efforts.

Simulation is also much faster than experimental B_1+ mapping [34]. Our simulations required 7 minutes on a computer with two 2.33 GHz quad-core Xeon processors and 24 GB memory. The Bloch-Siebert B_1+ scans required 19 minutes without considering recurring setup time (about 1 hour for exact slice prescription). In our application, the greatest value was afforded by fast simulation to compare coil geometry. However, our framework allows inclusion of additional complexity at the cost of computational resources. We have characterized the uncertainty due to the major model simplifications and concluded that the greatest improvement in overall accuracy can be achieved by improving the transmit RF excitation model.

The modular strategy allows individual customization of each component. For instance, we can replace the homogeneous phantom model with an exact model for a particular organ [35]. Different RF excitation schemes like linear and quadrature excitation can be modeled, including, if necessary, a detailed model for the external coil. Fig 11 shows different multimode coil options that were considered for our pig study. Fig 11-III and 11-IV are modifications to the opposed solenoid design (Hillenbrand *et al.* [15], [16]) to add transmit field distortion and 3-point curvature tracking (using the two opposed solenoids and an independent larger peak-producing tracking coil) to its internal imaging and tip-tracking capability. Celik *et al.* [21]–[23] and Overall *et al.* [24] used reverse polarization for curvature detection and safety monitoring respectively. Simulation can be used to visualize the forward and reverse polarized B_1 fields in the presence of endovascular coils.

Fig 7 illustrates parametric design optimization simulations. To maximize radial coverage (FoM2), $N = 8$ turns of the solenoid tracking coil is optimal. For a dominant tracking peak (FoM1 > 1), $N = 18$ may be used. $N = 6$ represents a good compromise between FoM1 and FoM2. Other parameters that could be varied include the number of turns of imaging coil, catheter diameter, length of imaging coil, separation between tracking and imaging coil and distributed tuning elements. The modeling strategy also has a role to play in calibrating background excitation during wireless marker operation. B_1+ field values can be used in (8)

to compute flip angle maps and calibrated to detect overflipping. This can be used to obtain images with different contrast as shown in Fig 9.

$$\theta = \int_0^T \gamma B_{1,+}(t) dt \quad (8)$$

The intrinsic signal to noise ratio (ISNR) ϕ_s (9) is used to compare coil SNR performance, independent of imaging parameters [36]–[38]. In (9), ω is angular excitation frequency, M_0 is equilibrium magnetization per unit volume, k is Boltzman's constant, T is absolute temperature, R_{load} is sample loading, and $B_{1,+}$ is excitation field value. Previously, $B_{1,+}$ constrained (assumed constant) optimization of cylindrical wave function expansions [37] and direct current excitation simulation have been used to simulate ISNR [39]. Our fast, full-wave FEM simulations model transmit RF excitation, compute induced currents and use tissue models to rapidly simulate $B_{1,+}$ maps. Unconstrained ISNR can be computed in a post-processing step.

$$\phi_s = \frac{\sqrt{2}\omega M_0}{\sqrt{4kTR_{load}}} \times B_{1,+} \quad (9)$$

Our models do have some limitations. Simulation accuracy and sources of error are summarized in Tables II and III. While modeling gives consistent $B_{1,+}$ maps, it does not predict the overflipping effect. This feature may be possible if flip angle maps are derived using (8). RF heating studies require much more detailed modeling, especially for the cable. We have only discussed magnetic field interactions, whereas tangential electric fields are mainly responsible for RF heating (ISO/TS 10974, 2012, Clause 10). However, if a model for the cable is included, it should be fairly straightforward to visualize the tangential electric field since a full-wave EM solver is used. In its current form, the model may only be used for semi-quantitative assessments of configurations and orientations of maximal inductive coupling [40].

Despite these limitations, this work represents, to our knowledge, the first attempt to comprehensively model inductive coupling of endovascular MRI coils. The detailed, modular and individualized component modeling gives important insight into coil selection and optimization. This also allows trade-offs between accuracy and computational complexity for each individual component. Simplified and fast simulation models give rapid insight into endovascular coil design decisions, as a precursor to more time-consuming and expensive fabrication and imaging studies.

VI. Conclusion

Transmit RF inductively-coupled endovascular coils provide several desirable features like device tip-tracking and wireless visualization. Our modeling strategy addresses such functionality and is an important step in designing application-specific, optimized coils for MRI-guided interventions.

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Fig. 1.

(a) The multimode coil design with series-tuned solenoid tracking coil in series with a series-tuned rectangular loop imaging coil. The fabricated coil has a π -matching network connecting to the microcoaxial cable. (b) The FEM simulation model of the multimode coil approximates loaded coil and cable losses using lumped elements and so does not include a matching network. The transmit RF excitation (B_1+ green arrows) induces significant current (on the order of 0.5 mA) in the series resonant multimode coil (red arrows).

Fig. 2.

Idealized quadrature birdcage excitation: sinusoidal surface current density of a cylindrical domain. Two quadrature components are shown using red and blue arrows respectively. Their spatial and temporal relationship is shown in (b) and (c). The multimode coil is placed at the center of the cylindrical domain. A photo of the fabricated coil with dimensions is shown.

Fig. 3. $|B_{1+}|$ field map ($|B_{1+}| = 24 \mu\text{T}$) from a) Simulation model; and b) Bloch-Siegert sequence show good agreement, within modeling and imaging limitations (Table II). Experimental map is more spread out due to fabrication limitations and tilt. Notable features are: 1) Dipole-like field addition and subtraction at opposite ends of tracking solenoid (solid black arrows), 2) High fields that rapidly decrease radially away from the rectangular loop (dashed white arrows), 3) Field subtraction inside the rectangular loop. Location of the 1D profiles plotted in Fig 6 are shown by the dashed black lines.

Fig. 4. $|B_{1+}|$ field map ($|B_{1+}| = 28 \mu\text{T}$) from a) Simulation model; and b) Bloch-Siebert sequence. Note the higher background B_{1+} values in Fig. 4 compared to Fig. 3. This is due to larger induced current effects. Over-flipping, more clearly visible in Fig 5, just begins to appear in Fig 4b.

Fig. 5. $|B_1+|$ field map ($|B_{1,r}| = 34 \mu\text{T}$) from a) Simulation model; and b) Bloch-Siegert sequence. Note the larger induced current effects. This results in overflipping (zoomed inset) and manifests as lower (“apparent”) field values in the Bloch-Siegert maps just outside the rectangular loop imaging coil (white downward arrow). A small distance away, the artifact disappears and expected high field regions are seen again (black upward arrow). The asymmetry due to tilt, partly visible in Fig 3b and 4b, is more pronounced here due to the larger excitation field.

Fig. 6.

Radial 1D $|B_1+|$ profiles for a) Tracking profile; and b) Imaging profile (dashed lines, Fig 3) show agreement. Limitations are: 1) A single point (finger, Fig 6b), where a simulation discontinuity artifact, fabrication limitations and signal averaging influence field value; 2) No signal is seen within the acrylic due to short T_2^* . Two figures of merit (experimental) are shown. Ratio of Tracking Peak (TP) to Imaging Peak (IP) = $FoM1 \sim 1$, imaging coverage = $FoM2 \sim 0.9$ cm (both agree with Fig 7, $N = 6$).

Fig. 7.

Simulations indicate that FoM1 peaks at $N = 6$ (experimental agreement with Fig 6), dips to a minimum at $N = 9$ and subsequently rises consistently. In region I ($N = 4, 5, 6$), the imaging coil dominates inductively-coupled current, so $FoM1 \approx 1$. In regions II and III, the tracking coil has an impact on overall induced current. In region II ($N = 7, 8, 9$), increased tracking coil resistance causes a dip in FoM1. In region III, induced current in the tracking coil dominates and FoM1 rises above 1 at $N = 18$. FoM2 is highest while the imaging coil is dominant and declines rapidly once tracking coil dominates. Note the trade-offs between FoM1 and FoM2.

Fig. 8. Multimode coil in active tip tracking mode: (a), (b) and (c) show multimode coil in the inferior vena cava of a pig. The red dot, placed at region of maximum signal intensity, corresponds to tip tracking location. This clearly helps to track device movement into the RV. (d) and (e) display the tracking signal profile when viewed from 2 orthogonal imaging planes.

Fig. 9. Multimode coil in the inductively-coupled wireless marker mode in the pig heart: (a) and (b) show images at two different flip angles, indicating device positioning with respect to the cardiac anatomy.

Fig. 10.

Image of a cavity in an *ex-vivo* pig heart using the multimode coil placed inside the blood vessel: (a) Composite image from the external coil and multimode coil as a multi-channel receiver shows poor overall SNR. (b) Small FOV, high SNR image using the multimode coil as a single-channel internal receiver coil shows a clear image of the cavity with high SNR.

Fig. 11. Multimode coil geometries: I) Rectangular loop imaging + solenoid tracking; II) Rectangular loop imaging + Helmholtz pair tracking (selected for pig study); III) Opposed solenoid imaging + solenoid tracking; IV) Opposed solenoid imaging + Helmholtz pair tracking.

TABLE I

Fabricated and Simulated Coil Impedances

Fabrication Stage	Measured Coil Impedance (Ω)	Simulated Coil Impedance (Ω)
Tracking Coil	$6.1 + 173.0i$	$1.51 + 164.1i$
Tuned Tracking Coil (with added interconnect/coil loss)	$14.5 - 1.8i$	$14.5 - 2.06i$
Tuned Tracking Coil and Imaging Coil	$25.6 + 163.5i$	$23.7 + 172.5i$
Tuned Tracking Coil and Tuned Imaging Coil (with net added imaging coil / interconnect loss $R1 = 28 \Omega$)	$32.0 - 5.0i$	$32.3 - 6.2i$
Final Coil Termination	Match to 50Ω micro-coax	Lumped cable loss ($R2$) = 5.5Ω

Coil impedances measured at various stages of fabrication and compared to FEM coil model predictions.

TABLE II**Modeling and Experimental Limitations**

Sources of Simulation Error	Sources of Experimental Error
Numerical computation errors at regions of field discontinuity	Partial volume, resolution and averaging
Homogeneous transmit field approximation	SNR within reasonable scan time and slice thickness
Planar coil approximation	No signal from acrylic tube
Lumped element cable losses approximation	Phased array (receiver) coupling
Over-flipping artifact not captured	Reduced dynamic range (over-flipping seen as reduced field values)
Heat shrink tubing not conforming to coil dimensions	Fabrication limitations- in-plane positioning of coil conductors and tilt
	B_0 inhomogeneity (see [31])

TABLE III

Analysis of Model approximations

Approximation	Resulting model approximation error	Estimation of Error Quantity	Parametric Uncertainty	B ₁ Simulation Error Estimation Method	Error in B ₁ Simulation
Idealized Birdcage Coil	Transmit B ₁ inhomogeneity	Maximum transmit field variation in 2 x 4.5 cm slice (from BS B ₁ + maps)	14.9% (± 3.6 μ T)	Simulations at ± 3.6 μ T: Error = deviation of peak B ₁ value	11.5%
Lumped Element Approximation of Coil and Cable Losses	Additional mismatch, dissipative and sample loading losses	Maximum deviation in impedance bench measurements at 5 different positions	6.3% (± 2.1 Ω)	Simulations with lumped element losses R ₁ + R ₂ $\pm$ Max error: Error = deviation of peak B ₁ value	2.3%
Planar Coil Approximation	Variation in coil impedance	Analytically $-(\pi d - 2d)/(2L + 2\pi d)$	1.7%	$B_1 \propto \frac{I_{ind}}{Z_{coil}}$ & vprop;	1.7%
Net				Maximum error	15.5%