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INTERCOMPARISON OF PERFORMANCE OF RF COIL GEOMETRIES FOR HIGH FIELD MOUSE CARDIAC MRI

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Abstract

Multi-turn spiral surface coils are constructed in flat and cylindrical arrangements and used for high field (7.1 T) mouse cardiac MRI. Their electrical and imaging performances, based on experimental measurements, simulations, and MRI experiments in free space, and under phantom, and animal loading conditions, are compared with a commercially available birdcage coil. Results show that the four-turn cylindrical spiral coil exhibits improved relative SNR (rSNR) performance to the flat coil counterpart, and compares fairly well with a commercially available birdcage coil. Phantom experiments indicate a 50% improvement in the SNR for penetration depths 6.1 mm from the coil surface compared to the birdcage coil, and an increased penetration depth at the half-maximum field response of 8 mm in the 4-spiral cylindrical coil case, in contrast to 2.9 mm in the flat 4-turn spiral case. Quantitative comparison of the performance of the two spiral coil geometries in anterior, lateral, inferior, and septal regions of the murine heart yield maximum mean percentage rSNR increases of the order of 27–167% in vivo post-mortem (cylindrical compared to flat coil). The commercially available birdcage outperforms the cylindrical spiral coil in rSNR by a factor of 3–5 times. The comprehensive approach and methodology adopted to accurately design, simulate, implement, and test radiofrequency coils of any geometry and type, under any loading conditions, can be generalized for any application of high field mouse cardiac MRI.

Keywords

Magnetic Resonance Imaging; cardiac; mouse; radio frequency coils; B₁-field simulations; image processing; SNR; relaxation times

1. Introduction

The mouse has been emerging as one of the most prominent models for the study of cardiac function over the past few years. As an extension of such work, recent imaging attempts have been focusing on image-based cardiac phenotyping of normal and transgenic mice (Wiesmann 2001, Badea 2005). Despite the fact that a number of prior studies quantified

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Conflicts of Interest

There are no conflicts of interest to declare.

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There are no disclosures.

cardiac mechanical function with state-of-the-art imaging technology, including magnetic resonance imaging (MRI), some of such prior attempts (Wiesmann 2001, Ross 2002, Yang 2002, Zhou 2003, Gilson 2005, Bucholz 2008, Miraux 2009, Bucholz 2010) have not considered optimal radiofrequency (RF) designs, but have instead exploited conventional surface and quadrature transmission (TX)-reception (RX) coils. Other attempts have adopted new phased array technology for the development of instrumentation for multi-channel mouse cardiac ^1H MRI (Ruff 1998, Williams 2001, Cassidy 2004, Dawson 2004, Beuf 2006, Sosnovik 2007, Zhang 2008), where the combined use of transmit birdcage coils with receive phased arrays have proven particularly valuable for increasing image sensitivity over conventional volume coils (Gareis 2007, Ramirez 2010).

The reduced field-of-view (FOV) acquisitions for mouse imaging, the increased cost and technical complexity of multi-channel phased array systems, their inherent complex mutual coupling effects, in addition to insignificant additional benefits in signal-to-noise (SNR) performance of such arrays (relative to their prohibitive high implementation and receiver electronics cost), have led to an increasing use of birdcage coils (good filling factors, uniform B_1 excitation) and simple T/R surface coils for imaging the mouse (Doty 2007, Sosnovik 2007, Lanz 2010). While volume coil excitation and surface coil reception is preferable, nevertheless, volume coils impose limitations in high field mouse imaging where high workflows are often needed (Doty 2007). Also, modern high-field scanners are now routinely manufactured with a capability to support only a single RF channel. Such scanners also lack infrastructure that allows separate transmit/receive decoupling, establishing the choice of transceiver coils, obvious. Furthermore, it is often preferable to restrict the transmitted power only to the region of the coil's receive response region, as encountered in double-resonance polarization, or high-power decoupling schemes, favoring the preference for transmit/receive coils (Doty 2007).

Surprisingly, not much attention has been expended on the optimization of the size and geometry of transceiver surface coils employed in prior successful imaging attempts (Sosnovik 2007), with constructed coils having dimensions that range from 25–35 mm. Considering a distance of the infero-lateral cardiac wall from the mouse chest surface of approximately 10–14 mm, the ideal coil should be designed in such a way to evoke a B_1 -field response with tissue penetrance of such cardiac regions with adequate SNR. While the effort to quantitatively balance the tradeoff between sensitivity and uniformity is not easily attained, the foremost goal in this effort is achievement of adequate sensitivity for infero-lateral wall imaging, with adequate SNR, given the inherent non-uniformity of transceiver surface coils with depth. A multi-turn spiral transmit/receive surface coil is proposed in this work for use in a flat and cylindrical arrangement, for mouse cardiac imaging. While the concept of spiral coils in NMR, MRI, and other biomedical applications (cochlear implants, stimulators, infusion pumps, implantable devices) is not novel, and has been encountered over the last 5–10 years (Aslop 1998, Stork 2009), their use and application for mouse cardiac imaging is new. Important also is their use at high field strengths where the decreasing RF waveform penetration becomes a serious problem, ultimately leading to decreasing RF field homogeneity, and complicating the construction of excitation/reception coils (Aslop 1998).

With the recent growth of dedicated high-field animal magnets, including systems at the 4.7 (200 MHz), 7.1 (300.5 MHz), 9.4 (400 MHz) and 11.7 T (500 MHz) field strengths, full-wave electromagnetic modeling has become an essential aspect of the preliminary evaluation of the performance of RF coils for high field MRI applications (Ibrahim 2005). Specifically, finite-difference time domain methods (FDTD) have been widely employed to achieve electromagnetic modeling of RF coils, but their use and applications to dedicated cardiac coils is lacking.

Overall, this work establishes a methodology for accurate, quantitative comparisons of B_1 performance of any coil design and geometry, that includes the initial computer design, free space simulations, and simulations under realistic conditions of loading (both with phantoms and computer-generated animal biomodels) using closed form and analytical solutions of the Biot-Savart and Maxwell's equations, accounting for the morphology, and electromagnetic property characteristics of relevant anatomic structures. Based on such processes, the paper aims to: (a) introduce a transmit/receive surface coil design to improve B_1 reception using multiple turns in a spiral arrangement, (b) validate its B_1 -field response using computer simulations, and (c) compare its performance in flat and cylindrical designs (using simulations and actual MRI experiments in phantoms and mice), and with the conventional birdcage coil.

2. Methods

Theory

Computation of the B_1 -field in free space is based on a numerical solution of the Biot-Savart law, expressed as:

$$\vec{B} = \frac{\mu_0 I}{4\pi} \oint \frac{d\vec{\ell} \times \vec{s}}{s^3} \quad [\text{Eq. 1}]$$

where μ_0 represents the absolute permittivity of free space, I is the current in the current conducting elements, $d\vec{\ell}$ is the vector along the current conducting path, and $\vec{s}$ the vector that joins $d\vec{\ell}$ and an arbitrary point in the three-dimensional (3D) space with cartesian coordinates x, y, z . For Matlab (Matlab Inc., Natick, USA) simulations, the B_1 -field was numerically computed using the Biot-Savart law. For finite difference time domain (XFDTD, Remcom, USA) simulations, Maxwell's equations were solved and computed using finite element computations in the time-domain. For flat coils, the coil axis resides in the center of the coil spirals; for cylindrical designs, the coil axis lies on center of the plane that intersects transversely the semicircular conductors. For flat spiral coil geometries, the magnitude of the total reception B_1 -field was computed from the superposition of the numerical integration of the field contribution from each of the straight current conducting segments (calculated as the square root of the sum of squares of all contributory components B_{1x} , B_{1y} , and B_{1z} in three-dimensional space), in agreement with prior published work (Mispelter 2006). For cylindrical spiral geometries, the B_1 -field contribution of the cylindrical current conducting segments was accounted for, based on the numerical integration of the expanded form of the Biot-Savart equation [Eq. 1], adopted for the particular geometry, as described previously by Griffiths *et al.* (Griffiths 1999), according to:

$$B_x = \frac{\mu_0 I z R^{\varphi_2}}{4\pi} \int_{\varphi_1}^{\varphi_2} \frac{\cos\varphi}{[R^2 + x^2 + y^2 + z^2 - 2R(x\cos\varphi + y\sin\varphi)]^{3/2}} d\varphi \quad [\text{Eq. 2}]$$

$$B_y = \frac{\mu_0 I z R^{\varphi_2}}{4\pi} \int_{\varphi_1}^{\varphi_2} \frac{\sin\varphi}{[R^2 + x^2 + y^2 + z^2 - 2R(x\cos\varphi + y\sin\varphi)]^{3/2}} d\varphi \quad [\text{Eq. 3}]$$

$$B_z = \frac{\mu_0 I R^2 \varphi_2}{4\pi} \int_{\varphi_1}^{\varphi_2} \frac{1}{[R^2 + x^2 + y^2 + z^2 - 2R(x\cos\varphi + y\sin\varphi)]^{3/2}} d\varphi - \frac{yB_y + xB_x}{z} \quad [\text{Eq. 4}]$$

where B_{1x} , B_{1y} , and B_{1z} represent the three cartesian components of the B_1 -field in 3D space, R the radius of curvature and ϕ the angle subtended by the circular conducting elements of the cylindrical spiral coil geometry.

Autocad Designs

To optimize the design (and hence the performance) of spiral coils initial simulations of the free-space B_1 -field were conducted in Matlab for different coils having different number of turns, with infinitesimal widths (Figure 1). A rectangular design was adopted since it is associated with less technical complexity in its design, and ease of construction in practice, as compared to circular spiral coils. Subsequently, XFDTD was used to optimize the response of spiral coils considering different layout parameters. These included, for example, the width of the turns, the gap between the turns, and the basic geometry design (Figure 3). Four different cases of flat spiral coils were considered having 3, 4, 5, and 6 rectangular turns. In all designs, the outermost coil dimension was maintained the same and the track spacing and conductor width varied. The spiral windings turned into the center of the spiral loop in all cases. Based on Biot-Savart simulation results, the flat spiral coil with 4 turns yielded the best results in terms of the elicited maximum B_1 -field intensity and its depth of penetration. This spiral design was also adopted for the case of the cylindrical case, where the coil's field response is expected to improve further compared to the flat equivalent coil, due to the additive B_1 -field effects of the quadrature component.

Three radiofrequency (RF) coils were designed (Figure 1) using Autocad Mechanical Desktop software (Version 2007, Autodesk Inc., San Rafael, California, USA) for RF excitation and reception at 300.5 MHz, and included a single-turn coil with dimensions of $4.7 \times 4.4 \text{ cm}^2$; a three, four, five, and six-turn spiral coil (outer coil dimensions of $4.7 \times 4.4 \text{ cm}^2$) with the track widths and track spacings being 5, 4, 3.17, 2.63 mm and 2.8, 1.7, 1.35, and 1.12 mm, respectively. The excitation wavelength (λ_0) is approximately 1 m at 300.5 MHz, and exceeds significantly the spatial scale of imaging (approximately a few cm). The total coil conductor length is, however, as large as 20–40 cm, and comparable to λ_0 .

The cylindrical spiral coils had the same dimensions as the flat equivalents, and were wrapped on a cylindrical former with an internal diameter of 32 mm that matched that of a commercially available quadrature coil (M2M imaging corporation, Ohio, USA). Coil dimensions ($4.7 \text{ cm} \times 4.4 \text{ cm}$) were chosen based on the computational simulation process. Such dimensions are relatively larger than coil dimensions chosen in prior studies for mouse cardiac imaging (typically in the range of $6 \text{ mm} \times 8 \text{ mm} \times 10 \text{ mm}$), however, they accommodate a broad range of mice of various weights (20–35 g approximately), allowing a close fit of the coil around the thorax, and extending coverage to the antero-lateral side of the chest.

Maximum coil dimensions were empirically chosen. Designed schematics were saved in standard Autocad file formats (.sat, .cad) and were exported in the commercially available software package XFDTD (Version 7.0.2, Remcom Inc., State College, Pennsylvania, USA).

Coil Development and Implementation

Construction and preliminary testing was conducted in the Laboratory of Physiology and Biomedical Imaging at the U. Cyprus, and MRI testing was performed on gel phantoms and on two C57BL/6J mice (one post-mortem and one under anesthesia) at the Center for In Vivo Microscopy (CIVM) at Duke University, according to the National Institutes of Health and the Veterinary Services National ethical rules and directives. Square conductor designs are considered for practical simplicity and ease of construction using etching technology. The coils were designed and constructed with thin (0.02 mm thickness) copper sheets

(having 4 mm track widths), using standard photo-etching technology. For the technical construction, coil lithography/etching materials were also used that included professional photo-resist lamination of microwave substrates (Kepro Bench-Top Laminator Model BTL-212, Total Systems Concept Inc., Fenton MO, USA) customized CAD/CAM design software for translation of coil patterns to masks (Inventor Pro, Autodesk Mechanical Desktop), using high intensity photo-resist exposure (Gralab 600, Dimco-Gray Co, Centerville Ohio, USA), and hardware tools.

The coils were implemented as transmit/receive coils with resonant circuitries using high breakdown voltage (>1000 V) ceramic capacitors (ATC Inc., USA), and shielded RG58/59 coaxial cable (Belden Inc., USA). Electrical losses were minimized by distributing the total capacitance, through the introduction of 6–8 of copper-track breaks. Coil matching was achieved with the use of series and parallel input variable capacitors, as reported by Schenck (Schenck 1992). Variable matching and tuning capacitors were used (Oxley Inc., Cumbria, UK) to account for sample loading due to the coil assembly coupling after placement inside the scanner gradient insert. Coil tuning and matching was achieved with a state-of-the-art 1.5 GHz Network Analyzer (Model E5061A, Agilent Inc., USA). Interface of constructed probes with the scanner and measuring electronics (Network analyzer, digital oscilloscope, sweep generator) was achieved with standard BNC non-magnetic 50 Ω termination connectors. The coils were interfaced with a TR-switch/Preamplifier integrated circuit (Nova Medical, Wilmington, MA, USA) having a measured 0.4 dB Noise Figure (a rated maximum input power of 12 kW, and a standard scanner DC bias input voltage of -8 V). The peripheral TR-switch/preamplifier module is equipped with two channels, for each of the two quadrature channels (combined after the preamplifier using a phase shifter module) for the cases of birdcage volume excitation.

Electrical Performance

Electrical response was assessed with the measurements of unloaded and loaded parameters on phantoms and animals, and included values of the quality factors (Q_u , Q_l), reflection coefficient (ρ), single port reflection parameter (S_{11}), and composite impedance. These parameters were measured using the network analyzer by sweeping the frequency from low values to the Larmor frequency ($f_0=300.5$ MHz) at 7.1T. The free-space comparisons of the Biot-Savart to XFDTD modeling presents an attempt that exist in the literature to facilitate the validation of XFDTD to yield accurate quantitative measurements in high frequency RF simulations, despite prior claims and reports (Doty 2007). Additionally, the XFDTD platform (in comparison to the simple Biot-Savart computations in Matlab) allows simulations of lossy conducting media (adipose tissue, skeletal musculature, lungs, blood cavities, and cardiac tissue) to be carried out in real-life computational animal models and in phantoms, with specified electrical and magnetic properties, under both unloaded and loaded conditions using finite element mesh computations, as described below.

Simulations

Assessment of the coil performance also included B_1 -field simulations of the transverse and total field components in free space using the Maxwell's and Biot-Savart equations (Smythe 1950, Griffiths 1999) with Matlab. For XFDTD simulations, the coil conductors were defined as perfect electric conductors (PEC). A 1 V sinusoidal feed was applied, and the source and conductor impedances were set to 50 Ohms (pure resistive). Alternatively, a 1 A current source in parallel with a 50 Ohm source input impedance can be defined. Simulation tests from axial B-field comparison in the 4-turn spiral coil case showed that driving the XFDTD models in either mode of operation does not generate different field responses, yielding total percentage difference errors of the elicited magnitude B_1 field responses in the two cases (for axial, sagittal, and coronal orientations), in the range of $2.4 \cdot 10^{-5} \pm 5 \cdot 10^{-5} \%$. Excitation

was emulated using a 1.5 ms modulated gaussian pulse centered at the resonant frequency of 300.5 MHz. Simulations did not account for coil tuning or matching conditions. The distribution of the magnetic field was simulated at the mid-axial, the mid-sagittal, and at a coronal slice placed 5 mm above each of the three coils in the: (a) free-space case; (b) under loading conditions with a 50 ml gel cylindrical phantom (having an external diameter of 3.2 cm and a length of 11.7 cm) placed 1–1.5 mm away from the closest coil conductor; and (c) under loading conditions, with a rat-mesh model scaled to the dimensions of a typical 25–30 g mouse (with approximate superior-inferior, anterior-posterior, and lateral-medial-lateral dimensions of 8.9 cm, 4.5 cm, and 2.5 cm, respectively), and placed 1.1 (flat spiral coil) and 5 mm (cylindrical spiral coil) away from the closest coil conductor, to prevent an undesired merging of the computational rat-mesh model with the constructed RF coil electrical circuit models (that would invariably lead to simulation errors). To validate the free-space simulations results (solution of Maxwell's equations), the computed XFDTD results were compared with corresponding results from routines written in-house in Matlab (Version 7.1, Mathworks Inc., Natick, MA, USA), assuming infinitesimal wire segment widths. For the simulation, the Biot-Savart law was numerically solved for all the straight (or cylindrical) wire elements of each of the three coils. The reconstructed B_1 -field grid with Matlab was spatially matched to that of the XFDTD simulations and encompassed [133–168]×[84–142] axial grids, [102–133]×[84–190] sagittal grids, and [136–145]×[83–419] grids for coronal slices (with cell sizes of 0.25–1 mm³), for quantitative comparison purposes.

MRI Imaging

After successful implementation, the probe performances were tested on a 7.1 Tesla GE (General Electric, USA) small animal MRI scanner equipped with a GE-EXCITE console (EPIC 12.4) and 77 Gauss/cm gradients having a 5 to 95% rise time of 150 μ s, first on phantoms, and then on mice post-mortem. New coil configuration files were generated on the scanner with appropriate cable and reflection coefficient losses for testing. Testing included a series of phantom and cardiac image comparisons that allowed assessment of the best probe performance, in magnitude reconstructed images from the flat and cylindrical spiral coils.

Phantom Construction and Imaging

Saline solutions (154 mEq/l NaCl) were heated to boiling temperatures. Agarose powder (Signa-Aldrich, Missouri, USA) and copper sulfate (CuSO_4) were added at various % and molarity concentrations to attain desired T_1 and T_2 relaxation parameters, according to a pre-existing state-phase diagram, and indicative relaxations values published previously (Stanicz 2005). Upon dilution, the solution was poured into 50 ml plastic cylindrical vials and was allowed to solidify. The commercially available birdcage coil was used for all relaxation time imaging and measurements. A conventional Carr-Purcell Meiboom Gill (CPMG) pulse sequence with 16 echoes was employed for single slice T_2 imaging. TE was stepped in 16 equally spaced intervals of 7 ms duration, starting at an echo time of 7 ms. The acquisition parameters were: Field of View FOV=5 cm, repetition time TR=6 s, echo time, bandwidth BW=31.25 kHz, slice thickness ST=2 mm, and a 128×128 acquisition matrix. For T_1 imaging, a conventional saturation recovery (MEMPR) pulse sequence was employed with TE=7ms, FOV=5 cm, ST=2 mm, BW=31.25 kHz, number of averages NEX=1, and a 256×256 acquisition matrix. Sixteen different images were reconstructed at equal time intervals, every 375 ms.

For B_1 -field penetration comparative studies in phantoms (using the flat, cylindrical spiral coils and the birdcage coil), a conventional multi-slice spin echo sequence was used in axial, sagittal, and coronal orientations, with FOV=5 cm, ST=2 mm, BW=31.25 kHz, 128×128 matrix, TE=8 ms, number of slices=19, NEX=1, at fully relaxed conditions (TR=6 s).

Mouse Cardiac Imaging

Further to the successful implementation and testing of the transceiver probes, they were subsequently tested on two mice; one mouse under post-mortem conditions, (non-gated spoiled gradient echo study) and compared with the commercially available birdcage coil on a different mouse using a conventional spoiled gradient echo sequence with TE=3.7 ms, TR=100 ms, excitation angle=30°, NEX=2, FOV=5 cm, ST=2 mm, and BW=15.63 KHz. The first male C57BL/6J mouse (weight=35.8 g; age=32 weeks) was imaged using MRI post-mortem. The spiral coils were perfectly tuned and matched before each acquisition under loaded conditions (with the mouse in place, while positioned in the bore of the magnet) using a Hewlett-Packard Spectrum/Network Analyzer (HP4195A, Hewlett-Packard, CA, USA). The excitation pulse was calibrated with respect to a 90° pulse, based on the transmit power setting for optimum object projection intensity in pre-scan mode. Blood and tissues begin to deteriorate and undergo ultra-structural changes surprisingly quickly after euthanasia. To ensure proper comparison of our surface spiral coils and the birdcage, a separate imaging acquisition was completed using a second C57BL/6J mouse (age=32 weeks; weight= 34.5 g). To ensure that differences in estimated SNR values (for comparison of the coil performance as listed in Table 2) were not due to underlying physiological changes, a time-series image acquisition was obtained with the birdcage coil, with imaging onset during in vivo conditions (under 1.5% inhalational isoflurane), followed by euthanasia in situ (using excess overdose of isoflurane at approximately 5% to effect), and subsequent imaging post-mortem. Reconstructed images from the flat, cylindrical spiral surface coils, and the birdcage coil, were temporally and spatially matched (at the same intervals post-euthanasia, using the same prescriptions). In the case of the first mouse, the flat coil was scanned first, followed by the repositioning/wrapping of the same coil in the cylindrical frame (of the mouse cradle), approximately 15 minutes later. Coils were positioned in all cases on a specially designed mouse cradle (or a plexiglass clone equivalent), using cradle fiducial markers and positioning landmarks, to facilitate accurate placement of animals tested. The temporal evolution of image intensities in various myocardial regions post-mortem yielded non-significant variations, in the range of 3.1–7%, and within the expected error margin of such measurements.

MR Processing

Phantom images were saved in standard .raw or DICOM formats and were exported and read in Matlab. Custom-made Matlab routines were written that allowed further processing and visualization. Relaxation parameter estimation (T_1 , T_2) was achieved by region-of-interest (ROI) analysis. ROIs were selected in central phantom image regions and mean signal intensities were estimated at the various TR or TE frames followed by subsequent curve-fitting using standard mono-exponential rise or fall curves (Logger Pro, Version 3.5.0, Vernier Software and Technology, Beaverton, OR, USA).

Quantitative comparisons were based on image region-of-interest signal-to noise ratio (SNR) values (Henkelman 1986, Constantinides 1997), field homogeneity, and depth of penetration of transverse field components in stationary phantoms and mouse cardiac studies, described in detail below. SNR estimates were obtained as the ratio of mean signal estimates (in myocardial regions of interest) to the estimated standard deviation value from background ROI analysis (in artifact-free regions), corrected for magnitude-reconstructed biases according to prior reports in the literature (Henkelman 1986, ConstantinidesII 1997). Four rectangular regions (area of 0.34 mm²) were placed in anterior, lateral, inferior, and septal regions of the left ventricular (LV) wall. To allow quantitative B₁-field comparisons of XFDTD simulated and actual phantom MRI results, measured signal intensity values from MRI phantom mid-axial profiles were fitted using an inverse exponential function using Logger Pro. The fitted curve was used to extrapolate the coordinates of corresponding points

that matched the XFDTD mid-axial simulated phantom profile and a mean error estimate was computed.

Statistical Analyses

All reported values are stated as mean±standard deviation. Quantitative comparisons were based on image signal-to noise ratio (SNR) (Williams 2001, Cassidy 2004) values, effective field-of-view (FOV), field homogeneity, and depth of penetration of transverse field components in stationary phantom and mouse cardiac studies.

3. Results

B_1 -field and profile comparative simulation results from all coil geometries (single turn, three and four turn flat, and four-turn cylindrical spiral) in free space are shown in Figure 1. The maximum elicited free-space B_1 field magnitude and depth of field penetration for the three, four, five, and six spiral coils were 0.32, 0.40, 0.25, and 0.1 A/m, and 18, 19, 17, and 14 mm, respectively. Figure 2 presents diagrammatical representations of the designed coils and a custom made coil. Electrical performance characteristics yielded unloaded and loaded gains (S11 measurements) in the flat four-turn spiral coil of -14 and -17.3 dB, total impedances of $57.8+j21.1 \Omega$, and $55.0+j13.2 \Omega$; and corresponding values of -20.0 and -30.0 dB, $58.1+j10.4 \Omega$ and $49.5-j3.5 \Omega$ for the cylindrical four-turn spiral coil. The reflection coefficient was < 10% under all conditions. The unloaded to loaded quality factors (Q_u/Q_l) for the flat, cylindrical, and birdcage coils were 2.93, 1.58, and 1.49, respectively. Based on such measurements, the sample's noise component is comparable to the coil's noise component. Figure 3 presents simulation results as a result of the process followed for determining the optimal field response dependent on the number of coil turns, gap spacing, and conductor width. Percentage field maps were generated based on the normalization of the B_1 -field to the maximum value recorded within each map. Since the XFDTD computation avoids singularities, maximum field values were recorded on the surface of coil conducting elements. Amplitude normalization of such calculations allowed comparison of the differences in generated maps, based only on their spatial distributions. Mean percentage (%) differences were then computed on a point-by-point basis. Quantitative percentage B_1 -field differences of the 4-turn to the 5- and 6-turn coils ranged between $0.95\pm3.2\%$ and $1.3\pm3.8\%$, respectively. Figure 4, presents comparative free-space simulation results between Matlab and XFDTD at axial, sagittal, and coronal slices for the single turn, the flat, and cylindrical 4-turn spiral coils. The percentage difference of simulated B_1 -field distribution (Matlab, XFDTD) was quantified in a total central volume of $43\times41\times11 \text{ mm}^3$, at a depth of penetration from the coil-conductor surface of 1–13 mm. Percentage simulated field differences (means±SD) in axial, sagittal, and coronal orientations were 2.8 ± 3.1 , 7.7 ± 4.6 , and $6.5\pm5.5 \%$, respectively, for the flat spiral coil case. The cylindrical coil comparison yielded axial, sagittal, coronal percentage differences of 6.2 ± 5.3 , 5.7 ± 4.9 , and $7.8\pm7.1 \%$, respectively. Figure 5 shows prominent improvement in performance for the cylindrical spiral coil in comparison to the flat equivalent, both in terms of the increased field-of-penetration, but also in terms of the increased B_1 -field response in regions where the mouse heart is typically positioned in unloaded and loaded cases (with a 50 ml cylindrical phantom). Table 1 lists the estimated quantitative results for the various gel phantoms constructed [$T_{1,\text{measured}}=966\text{--}1361 \text{ ms}$, $T_{2,\text{measured}}=30\text{--}55 \text{ ms}$]. Typical axial XFDTD mouse simulation results (at the mid-ventricular level) are shown in Figure 6. Qualitative and quantitative comparisons of B_1 -performance (scaled for mouse results with the square root of the equivalent noise resistance from each coil measured, yielding relative SNR [rSNR] estimates), of the tested four-turn flat and cylindrical spiral coils under loaded conditions (gel phantom) using ^1H MRI are depicted in Figure 7A, indicating a 50.3% increase in SNR performance on the spiral cylindrical coil surface compared to the birdcage coil, and overall

increased performance for penetration depths up to 6.1 mm from surface (Figure 7B). The simulated flat spiral B_1 -field was scaled to match the maximum surface B_1 -field value (at the coil's center axis) obtained from the actual MRI experiment. The profiles depicted in Figure 7 are the actual profiles obtained from phantom imaging using the flat, cylindrical spiral coils, and the birdcage coil, on the same occasion and using the same prescription protocol and acquisition parameters. The maximum intensity of the profile obtained from simulated results of the flat 4-turn spiral coil was intensity matched to the maximum intensity of the constructed and tested 4-turn spiral coil. Also shown are actual MRI cardiac images from the flat and cylindrical spiral (post-mortem imaging), and a commercially available birdcage coil (*in vivo* imaging). Table 2 shows a quantitative comparison of the cylindrical and flat spiral coil arrangements in mean rSNR values estimated in anterior, septal, inferior, and lateral locations of the left ventricular myocardium from 0.34 mm^2 regions-of-interest in post-mortem mouse cardiac MRI. All rSNR values were corrected for the non-linear reconstruction biases associated with the background noise estimation (Henkelman 1986, Constantinides 1997). While phantom MRI using flat and cylindrical coils was performed under optimally placed coil positions, positioning the coil in optimal positions for cardiac imaging proved challenging. The offset in anterior-lateral coil position placement is reflected by the noted improved SNR performance of the flat coil in comparison to the cylindrical in those regions. Quantitative mid-axial phantom profile comparison between the flat and cylindrical cases yielded an 8 mm penetration depth at half-maximum field response for the cylindrical spiral and 2.9 mm for the surface coil. Additionally, simulated and actual mid-axial phantom profiles were in good agreement in the flat coil case as shown in Figure 7.

4. Discussion

This paper describes the design, implementation, testing, and use of a spiral radiofrequency coil for mouse cardiac MRI. While the concept of spiral RF coils is not new in MRI (Aslop 1998, Stork 2009), nevertheless, the introduction and use of such a spiral design for mouse high-field cardiac imaging is novel and new. Furthermore, its design accommodates comfortably the mouse thoracic areas, and can be easily shaped and accustomed to fit specially designed mouse cradles, thereby improving imaging efficiency and animal throughput for cardiac studies.

Traditionally, RF surface coils reduce the spatial coverage (and hence the field-of-view) compared to volume coils, while maintaining higher local SNR. The limitation for expansion of the spatial region of surface coil coverage (while maintaining or improving SNR), yielded to phased array designs (Roemer 1990, Noeske 2000, Gareis 2007), often combined with circularly polarized volume transmit coils (Ramirez 2010). This effort presents one of the first attempts to ameliorate such a surface coil limitation, in consideration of the small spatial scales, and the complexity and cost of phased arrays (Roemer 1990). Additionally, the use of such a surface coil is not prohibitive in terms of its concurrent use with dedicated and specially designed transmit (or receive coil arrays).

To allow quantification of the coils' response, electrical, in-vitro gel phantom, and postmortem *in vivo* tests were performed. Constructed gel phantoms yielded relaxation times in accordance to prior published values by Stanicz *et al.* (Stanicz 2005) for human myocardium at 1.5 and 3T $T_1(1.5\text{T})=1030\pm34 \text{ ms}$; $T_1(3\text{T})=1471\pm31 \text{ ms}$; $T_2(1.5\text{T})=40\pm6 \text{ ms}$; $T_2(3\text{T})=47\pm11 \text{ ms}$; and Schneider *et al.* (Schneider 2003) for mouse myocardium at 11.7T $T_1(11.7\text{T})=1100\pm270 \text{ ms}$; $T_2(11.7\text{T})=19\pm4 \text{ ms}$. Measured gel relaxation values are also in close-agreement to estimated values (based on the synthetic process). Significant deviations in measured and actual T_1 values were noted in one gel phantom, and are attributed possibly to the chemical synthesis process (mixture evaporation, agarose-water

mixture saturation and dilution error, inaccuracies in the prediction of the relaxation value from the state-relaxation curve, and relaxation value fitting-estimation errors).

Simulations formed a major part of this effort and establish a methodological platform for researchers who are conducting coil development for small animals at high field. In summary, simulations yielded accurate results in the free-space (compared with closed form or analytical solutions of the B_1 -field using Matlab). The import of redundant measurements (Biot-Savart Matlab and FDTD in Figures 1 and 4) provided evidence for the agreement in the results of the two sets in free-space. At low fields, the Biot-Savart generates results with reasonable accuracy to the solution of the Maxwell's equations. At high field strengths it is necessary to solve rigorously Maxwell's equations (Wang 2008). Results of this study also provide confidence to the argument against inaccurate computation of magnetic fields using Maxwell's equation in XFDTD, as reported earlier (Doty 2007). Such validation is established in the free-space case where Matlab is easily accustomed to accommodate such computation. Once this was achieved, subsequent results employed the powerful and unique computational features of the FDTD technique to design (or import) any coil geometry, account for sample (phantom and mouse) and coil loading, as well as quantitatively compute secondary effects of high-field penetration (skin depth effects, coil, and sample conductivity and permeability).

Given that the RF wave velocities of propagation are small compared to the speed of light (c), having an RF wavelength of 1 m approximately at 300.5 MHz. It is therefore believed that inductive, capacitive, and wave propagation effects are of such magnitudes that justify the validity of the quasi-static assumption.

Coil B_1 -field simulations favored the implementation and use of a 4-turn cylindrical spiral coil for mouse cardiac imaging. Accurate simulation results were also generated for the phantom, and the scaled-down mouse model cases. The placement of the computational mouse model on the designed coils is sub-optimal due to the inability to shape the model to fit the cylindrical former of the coil, in comparison to a real-life situation. This leads to non-optimal filling factors, and thus non-optimal B_1 -field penetration response for such simulations. Despite the versatility of the employed electromagnetic full-wave simulation package employed, stringent memory requirements are imposed by full-wave FDTD electromagnetic modeling calculations that increase exponentially as cell size is reduced, often becoming computationally intensive (Wang 2008). However, such FDTD calculations are relatively fast and range between 15–20 minutes on a desktop computer using an Intel processor [Intel(R) Core (TM) Duo CPU], depending on the loading conditions. As computational power and processor processing speed decreases, simulation times can be further reduced to only a few minutes.

The constructed coils were transceiver coils, and are associated with non-uniform excitation and reception field patterns. The effects of the non-uniform excitation of such coils are not easily discernible from their sensitivity response (reception field pattern). Typically, B_1 -mapping methods are employed as more meaningful ways to map and compare the coil sensitivity/homogeneity profile between different coil designs. Such methods, however, necessitate the use of direct imaging acquisition techniques and dedicated pulse sequences with RF transmit power cycling schemes to generate such maps at long TR or fully relaxed acquisitions ($TR \gg 3T_1$) (Hancu 1999, Ibrahim 2005). Such dedicated work fell, however, outside the scope of this concerted effort. Considering that all acquisitions for surface spiral coils were acquired at fully relaxed conditions, and that every effort was expended to ensure proper excitation angle calibration (based on volume or slice projection imaging in pre-scan mode in each of the two orthogonal projection directions), constructed B_1 -maps are considered accurate representations for coil B_1 -homogeneity response comparisons. For

penetration depth, noise also plays a major role, and B_1 -maps without the context of the coil's noise contribution would not be meaningful. Despite the fact that the flat and cylindrical spiral coils were the same physical coil placed on either a flat, or shaped on cylindrical support formers, B_1 -sensitivity images were scaled with the square root of the equivalent noise resistance from each coil under loaded conditions (phantom and mice), and relative SNR estimates (rSNR) were calculated that describe more accurately the sensitivity/penetration of such coils.

Imaging results showed improved performance of the cylindrical spiral coil in comparison to the flat counterpart, in agreement to prior published attempts for human imaging (Constantinides 1997). The cylindrical coil has increased field of penetration that allows visualization of the entire lateral and inferior myocardial walls with adequate rSNR. Its performance compares well and outperforms its flat equivalent in septal, inferior, anterior, and lateral myocardial areas (rSNR improvement between 27–167%), despite its non-optimal placement and positioning on the mouse (anteriorly and laterally), in this particular case. Its design and response, that clearly exploits the additive effect of the transverse B_1 -field component, also compares favorably with a commercially available birdcage coil within the region where the mouse heart resides. However, the birdcage coil exhibits the best performance associated with 3–5 times higher rSNR values over the entire left ventricular myocardial regions.

The fact that the constructed coils are transceiver surface resonators, certainly presents a conceptual and practical limitation for their wide-spread use, in the context of inhomogeneous excitation and their ability to allow quantitative cardiac mouse imaging. Direct use of such coils for high field mouse cardiac imaging is likely to result in increased loading effects between the RF and gradient coil inserts, imposing the need for use of specially constructed isolation shields. The inhomogeneous response of such coils, obviously translates to issues associated with artifacts, imperfect application of magnetization preparation pulses, or biased quantitative measures in cardiac functional and perfusion imaging. Furthermore, flip-angle sensitive techniques, such as fat saturation in cardiac MRI cannot be applied, due to the inherent variations in the flip angle spatial distribution, thereby limiting their potential applications.

Future work will focus on the combined use of the spiral coils and the commercially available birdcage coil, the concurrent use with transmit or receive phased arrays, as well as the design and use of superconducting cryocoils for optimal SNR performance in high-field, small animal cardiac imaging.

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Figure 1.

Theoretical simulation results of the free-space B₁-field distribution for a single-turn (top, left), three-turn flat (top, right); four-turn flat (middle, left), and four-turn cylindrical spiral coils (middle, right), based on the Biot-Savart law, using Matlab; (Bottom) Mid-axial B₁-field profile comparison for the various geometries.

Figure 2. Autocad coil designs and imported XFDTD cylindrical design; [Top] (Left to Right) single turn, three-turn flat, and three-turn cylindrical spiral coils; [Mid] XFDTD mouse model superimposed on the cylindrical three-turn spiral surface coil; [Bottom] constructed prototype four-turn spiral RF coil.

Figure 3.

Optimization of coil geometry and design parameters. Simulation results of the free-space B_1 -field distribution for 4-tun (left), a five-turn flat (right), and a six-turn flat (right) spiral coil (right), using XFDTD showing improved performance in the case of the four-turn spiral.

Figure 4. (Left column) Matlab, and (Middle column) XFDTD free-space simulated results of the B₁-field distribution of a four-turn flat spiral coil and a (Right column) four-turn cylindrical spiral coil at a (Top row) mid-axial, (middle row) mid-sagittal, and (Bottom row) coronal slice views. All color bars depict quantitative B₁-field variations from 0 to -40dB with respect to maximum computed field value.

Figure 5.

Simulated XFDTD B_1 -field results for four-turn flat (Left column) and cylindrical (Middle, Right columns) spiral coils under loaded conditions with a gel phantom at (Top row) axial, (Middle row) coronal, sagittal slice views. All color bars depict quantitative B_1 -field variations from 0 to -40 dB with respect to maximum computed field value.

Figure 6.

Mouse XFDTD simulation from the four-spiral cylindrical coil under loaded conditions with a mouse computational model. All color bars depict quantitative B_1 -field variations from 0 to -40dB with respect to maximum computed field value.

Figure 7. (Top; left to right) Flat, cylindrical spiral, and birdcage axial phantom response; (Middle) Corresponding cardiac MR images from each of the three coils in temporally matched post-mortem conditions; (Bottom) Quantitative mid-axial signal intensity profiles from each of the simulated and actual MRI phantom acquisitions shown on top.

Table 1

Estimated and measured T_1 and T_2 relaxation values for the constructed gel phantoms.

Gel	$T_{1,\text{estimated}}$ (ms)	$T_{1,\text{measured}}$ (ms)	$T_{2,\text{estimated}}$ (ms)	$T_{2,\text{measured}}$ (ms)
1	1000	966	38	30
2	1000	972	50	32
3	2000	1361	56	55
4	-	-	40	55

Table 2

Comparison of relative SNR (rSNR) values in the flat, cylindrical spiral coils, and the birdcage coil, in multiple cardiac regions of the murine heart from short axis MRI.

Region	rSNR _{flat} (Mean±SD)	rSNR _{cylindrical} (Mean±SD)	rSNR _{improvement} (flat vs cylindrical) (%)	rSNR _{birdcage} (Mean±SD)
<i>Anterior</i>	4.8±1.5	6.1±1.4	27	17.1±4.3
<i>Septal</i>	3.6±1.4	4.6±1.4	28	18.1±0.2
<i>Inferior</i>	1.5±0.3	2.9±0.3	93	13.5±1.3
<i>Lateral</i>	0.9±0.2	2.4±0.8	167	12.7±1.3