Optimized 64-channel array configurations for accelerated simultaneous multislice acquisitions in 3T cardiac MRI

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Purpose: Three 64-channel cardiac coils with different detector array configurations were designed and constructed to evaluate acceleration capabilities in simultaneous multislice (SMS) imaging for 3T cardiac MRI.

Methods: Three 64-channel coil array configurations obtained from a simulation-guided design approach were constructed and systematically evaluated regarding their encoding capabilities for accelerated SMS cardiac acquisitions at 3T. Array configuration $A_{\text{Uni-sized}}$ consists of uniformly distributed equally sized loops in an overlapped arrangement, $B_{\text{Gapped}}$ uses a gapped array design with symmetrically distributed equally sized loops, and $C_{\text{Dense}}$ has non-uniform loop density and size, where smaller elements were centered over the heart and larger elements were placed surrounding the target region. To isolate the anatomic variation from differences in the coil configurations, all three array coils were built with identical semi-adjustable housing segments. The arrays' performance was compared using bench-level measurements and imaging performance tests, including signal-to-noise ratio (SNR) maps, array element noise correlation, and SMS acceleration capabilities. Additionally, all cardiac array coils were evaluated on a healthy volunteer.

Results: The array configuration $C_{\text{Dense}}$ with the non-uniformly distributed loop density showed the best overall cardiac imaging performance in both SNR and SMS.
**INTRODUCTION**

MRI has become the tool of choice for many diagnostic tasks, including oncology, musculoskeletal conditions, central nervous system disorders, and cardiac diseases. Nonetheless, the slow acquisition speed still reduces cost effectiveness and general practicality. Recent advances in MRI include gradient and receiver coil design, supporting electronics, and new acquisition methods that take advantage of both improved hardware capability and new concepts of encoding and reconstruction. A major part of the improvements in acquisition speed has been achieved by employing acceleration techniques like parallel imaging (PI). Over the past 2 decades, clinically deployed PI technologies, such as SENSE and GRAPPA, have enabled typical acceleration factors of two to three times higher than conventional acquisitions through the use of multichannel receiver coil arrays. Using PI uniform undersampling of Fourier encoding provides acceleration, allowing structured image aliasing that can be resolved using spatial information from multiple receiver coils. These acceleration strategies always come with a reduction of the signal-to-noise ratio (SNR) since the time taken for data sampling is shortened. Additional noise amplification occurs through the requisite ill-conditioned image reconstruction algorithms.

Recently, an alternative approach has been developed, allowing the excitation and measurement of multiple slices simultaneously. Preliminary concepts for simultaneous multislice (SMS) acquisitions have been demonstrated as early as the eighties of the last century. However, following the widespread implementation of coil arrays, SMS imaging has adopted a new methodological direction to drastically reduce acquisition time. Unlike conventional PI, which necessarily shortens the data acquisition sampling, this simultaneous multislice (SMS) method provides acceleration by exciting spins in multiple slices at the same time using multiband (MB) radiofrequency (RF) pulses. These new MB acquisitions share the SNR benefits of 3D sampling that occur with 3D volume selection. However, in order to separate the simultaneous excited spins into individual MR slices, multichannel receiver coil arrays become a critical component in SMS applications. The combination of SMS imaging with advanced receiver coil encoding for slice separation have the potential to offer substantially accelerated data acquisition for a broad range of MRI applications. Any reduction in acquisition time can dramatically improve diagnostic image quality and practicality, which is particularly important in cardiac MRI. Breathing and motion of the heart restrict the time available for imaging and a trade-off must be found between SNR, spatial resolution, anatomic coverage, and temporal resolution. In addition, the rapid motion of the heart enforces complexity, where measurements must be split into multiple time frames, significantly prolonging the examination and reducing patient comfort.

The region of the heart consists of a much smaller volume compared to the human torso. This circumstance should be considered in an optimized array coil design that achieves highly accelerated cardiac imaging with combined in-plane PI and SMS techniques. As many state-of-the-art clinical MRI scanners provide 64 receiver channels, this offers an opportunity to translate the latest research in coil technology to cardiac MRI studies. Therefore, in this study, we analyzed whether coil arrays consisting of 64 channels could be optimized for SMS encoding performance in cardiac MRI. We implemented a simulation pipeline to determine the optimal configuration of the receiver elements in terms of element size, geometric coverage, and integration of the upper and lower arms. We constrained our simulation at a fixed element count, maintained sample noise dominance for possible small loop sizes, and included mechanical restrictions of body-shaped adaptable housing.

Based on the outcome of the simulation, we designed and constructed three 64-channel cardiac array coils and systematically evaluated their encoding capabilities for accelerated SMS cardiac MRI. We initially tested the constructed
coil arrays in in vivo cardiac diffusion imaging, where we validated the improved sensitivity and parallelism in arguably one of the most challenging and SNR-starved cardiac acquisitions.

2 METHODS

2.1 Array coil simulation

In order to find a suitable combination of 64 receiver elements for optimized SMS-encoding capabilities at the target heart region, the array coil must provide enough sensitivity variation along the slice direction. Therefore, with a first simulation approach, we determined how far the coil elements should be extended for the given target region. In particular, we wanted to determine the role of the elements distal to the heart in highly SMS-accelerated cardiac imaging. The simulation was carried out with the fast electromagnetic (EM) solver MARIE\textsuperscript{17-19} and included a conductive dielectric load ($\sigma = 0.71 \text{ S m}^{-1}, \varepsilon_r = 63.8$) using a meshed torso model. We constrained our simulation to 64 channels and a radially enclosed array coil topology to be placed on the upper torso and centered over the heart. Given the geometric constraints and the anatomical coverage, we extended the coil array coverage incrementally across the heart using four array layouts in an overlapped coil element topology (Supporting Information Figure S1, which is available online). Armed with the knowledge of the optimum array coverage, we simulated three 64-element array configurations as illustrated in Figure 1: (1) a tailing pattern with symmetrically distributed equally sized loops that overlap ($A_{\text{Uni-sized}}$); (2) a layout with symmetrically distributed equally sized loops with left-to-right gaps ($B_{\text{Gapped}}$); and (3) a configuration with a non-symmetrical loop distribution with higher loop density centered over the heart ($C_{\text{Dense}}$).

In the EM simulation, the loop conductor material was defined as copper ($\sigma = 5.8 \times 10^7 \text{ S m}^{-1}$). Excitation was emulated using a sinusoidal unit current at the Larmor frequency of 123.25 MHz (3 T). We used a simulation resolution of $(2 \times 2 \times 2) \text{ mm}^3$ for comparing the encoding capabilities between coil arrays. In order to match the simulation to the real phantom scans, we also simulated the three arrays $A_{\text{Uni-sized}}, B_{\text{Gapped}},$ and $C_{\text{Dense}}$ to $(4 \times 4 \times 4) \text{ mm}^3$ resolution. The complex reception profile of the coil was taken as the $B_1$ component of the simulated field. In addition, we calculated each array’s mutual noise resistance matrix from the inner product of the spatially varying electric fields of the coils over the sample volume.\textsuperscript{20,21} Simulations did not account for coil tuning or matching conditions and interelement coupling. The $B_1$ field image reconstruction and visualization were carried out in Matlab (R2020b, MathWorks Inc., Natick, MA, USA). Using the initial simulation outcome, we were able to parameterize the three coil configurations as described in Table 1.

Comparing the SMS-encoding power between the simulated coil arrays, we computed the inverse $g$-factor maps of emulated blipped-CAIPI acquisitions\textsuperscript{14,15} using the pseudo-multiple replica method\textsuperscript{7,22} with 1000-image pseudo-timeseries of the uncombined images. For a clinically realistic but ambitious scheme, we computed the accelerated noise

**FIGURE 1**  Anterior and posterior coil element configurations of the three 64-channel receiver arrays: ($A_{\text{Uni-sized}}$) symmetrically distributed equally sized loops in an overlapped arrangement, ($B_{\text{Gapped}}$) a gapped array with symmetrically distributed equally sized loops, and ($C_{\text{Dense}}$) a non-symmetrical loop density variation, where smaller elements were centered over the heart and larger elements were placed around the target region.
amplifications for the SMS multiband accelerations with $MB = 6$ and an in-plane acceleration factor of $R_{\text{SENSE}} = 2$. We evaluated 36 slices across the target heart region. Using a 4 mm slice thickness, we covered the field of view (FOV) of 144 mm in the z-direction, which provided interslice spacing of 24 mm for the six collapsed slices. A relative phase axis shift of FOV/3 was used. The same setting of the simulated slice prescription was then matched to the image acquisitions of an anthropomorphic torso phantom.

2.2 | Mechanical coil design

To isolate the benefit of different loop array configurations, it is critical to use the same coil former geometry for all three array coils. Therefore, we developed a generic mechanical coil design, which houses the electronic components of the three different coil arrays. The coil housing former was obtained from the average surface contours of aligned 3D MRI scans from 20 male upper bodies. All housing parts were modeled using a 3D computer-aided design (CAD) software (Rhino3D, Robert McNeel & Associates, Seattle, WA, USA) and printed in polycarbonate (PC) plastic using a rapid prototyping 3D printer (Fortus 450mc, Stratasys, Ltd., Eden Prairie, MN, USA). Based on the upper body model, we also designed a torso phantom for both simulation and imaging. The phantom was 3D-printed in polyamide (PA12) nylon plastic and filled with a dielectric water solution ($26.8 \text{H}_2\text{O}, 33.5 \text{g NiSO}_4 \times 6 \text{H}_2\text{O}, 134 \text{g NaCl}$). The conductive dielectric solution was measured to be $\sigma = 0.71 \text{ S m}^{-1}$ and $\varepsilon_r = 63.8$ using a dielectric probe kit (DAK-12, Schmid & Partner Engineering AG, Zurich, Switzerland). These dielectric parameters were matched to the average human muscle tissue at 123.25 MHz and showed nearly identical coil-to-sample loading conditions compared to real human torsos.

The housing shape was created to allow for a close-fitting surface of the natural morphology of the human upper body and comprised of two independent shell-like segments with a posterior and an anterior section (Figure 2). We included mechanical coil adjustability to account for different body shapes. The posterior section was designed so the subject could comfortably lie on the coil. The anterior section was placed on the subject’s chest and further divided into five segments, where a main body former links to four adjustable lateral wings (shoulder and thorax). The latter adapts to different body shapes. For large subjects, the shoulder wings fold out, while the lower wings remain between the chest and the arms. The coil adapts to small subjects, providing an advantageous close wrap of all anterior coil segments (all coil wings folded in).

2.3 | Array coil construction

The three array configurations obtained from the simulation pipeline were implemented in the 3D CAD model. The simulated loop positions were imported to the 3D-modeled coil former. The positions were then engraved on the former surface to facilitate accurate placement of the actual array coil elements according to the simulated geometry. Standoffs that
hold the coil-adjacent preamplifier boards and cable traps were also printed as part of the former.

Figure 3 shows a detailed circuit schematic of a representative coil element. A silver-plated copper wire (1.2 mm) was used to construct the loops. Depending on the loop diameter, we subdivided each element symmetrically with two or three gaps for small and large loop sizes, respectively. One gap was allocated to the coil’s output circuit board, which incorporated a capacitive voltage divider (Series 11, Knowles Capacitors, Norwich, UK) and a variable series capacitor (GFX2700NM, Sprague Goodman, Westbury, NY, USA) to impedance match the element’s output to an optimized noise match of 50 Ω. Additionally, the output circuit board used an active detuning circuit across one of the voltage divider capacitors. Active detuning during transmit was achieved using a PIN diode $D_1$ (MA4P4002B-402, Macom, Lowell, MA, USA) in series with a tunable inductor $L$ (150-02J08L, CoilCraft Inc., Cary, IL, USA), which together with capacitor $C_3$ resonated at the Larmor frequency. Thus, when the PIN diode $D_1$ was forward-biased (transmit mode), the resonant parallel $LC$ circuit inserted a high impedance in series with the coil loop, blocking current flow at the Larmor frequency during transmit.

For further enhancement of coil safety, a passive detuning circuit was implemented comprised of a cross diode (MADP-011048-TR3000, Macom, Lowell, MA, USA), the RF choke $L_{RFC2}$ (1812CS-272, CoilCraft Inc., Cary, IL, USA), and the capacitor $C_6$. As a final safety feature, we have implemented a series fuse $F$ (1999-6000-3150, current rating: 315 mA, Data Modul AG, Munich, Germany) for passive protection against large coil currents. For fine-tuning to the Larmor frequency again a variable capacitor (GFX2700NM, Sprague Goodman, Westbury, NY, USA) was used. Preamplifier decoupling was established to transform the preamplifier input impedance to a high series impedance within the loop. We used the impedance change of the series matching capacitor to transform the input impedance of the preamplifier to a parallel inductance across $C_2$ of the capacitive voltage divider. This parallel $LC$ circuit resonated at the Larmor frequency and introduced the needed high serial impedance into the coil loop. In this mode, the element is transformed to a voltage-source-driven pickup coil. Minimal current flows in the loop and inductive coupling to other coils is minimized, despite the presence of residual mutual inductance. The variable matching capacitor finely adjusted the impedance transformation of the preamplifier’s input.

FIGURE 3  Circuit schematic for a representative pair of coil elements and preamplifier chain. The coil elements contain four to five capacitors: $C_1$ to adjust tuning and $C_4$ to provide a low-noise figure-matched impedance. $C_2$ and $C_3$ have equal capacitance and provide a symmetric circuit design (the $C_7$ tuning capacitor is only incorporated into large loop sizes). A detuning trap is formed around $C_3$ using variable inductor $L$ and a PIN Diode $D_1$. If the active detuning trap fails, the coil circuit is passively detuned using a cross diode, an RF choke and the capacitor $C_6$. The bias-T and preamplifier are located on a preamp's daughter board.
2.4 Coil benchtop validation

Based on the design specifications of the three array coils, we constructed seven different loop diameters, ranging from 50 to 130 mm. The unloaded-to-loaded coil quality factor ratio ($Q_U/Q_L$) of one representative coil element of each loop size was assessed within the populated but detuned array assembly using the $S_{21}$ double-probe method. After populating all 64 receiver elements on every coil former, bench measurements verified the element tuning, active detuning, nearest-neighbor coupling, and preamplifier decoupling for each coil element. Prior to the actual bench measurements, the active detuning circuitry was fine-tuned at the Larmor frequency using an $S_{21}$ double-loop probe measurement by carefully adjusting the variable inductor $L$. Matching to 50 Ω for each loop element was achieved by adjusting the serial variable capacitor $C_4$. This impedance matching was monitored under an $S_{11}$ measurement using a coaxial cable directly connected to the preamplifier's input socket of the daughterboard, while the preamplifier was removed. If needed, the element was fine-tuned at the Larmor frequency with the variable capacitor $C_1$. Finally, we measured the preamplifier decoupling of a given loop with all other loops detuned. The preamplifier decoupling was measured as the change in the $S_{21}$, when the coil element’s output port was terminated with the low impedance preamplifier and the 50 Ω element.
termination, respectively. The nearest neighbor coupling was recorded using a direct $S_{32}$ measurement between pairs of elements. When necessary, the overlap was slightly altered to optimize the geometrical decoupling. After this procedure, a minor adjustment of tuning and matching was required.

Cable traps were adjusted under an $S_{12}$ measurement using a custom-made current probe on either side of the trap.\textsuperscript{23} The solenoid traps were pre-tuned slightly below the Larmor frequency using three high-power capacitors in series. Fine-tuning was achieved, by carefully incorporating a brass insert screw inside the center of the solenoid former, which reduces the inductance of the cable trap such that the trap circuitry resonates at the desired Larmor frequency. The smaller bazooka cable traps were fine-tuned by finding a suitable combination of the capacitors distribution directly on the shield.

### 2.5 MRI data acquisition and reconstruction

Imaging was carried out on a 3T clinical MRI scanner (MAGNETOM Prisma, Siemens Healthineers AG, Erlangen, Germany) with 80 mT/m maximum gradient amplitude and a maximum slew rate of 200 mT/m/ms. Timecourse stability of the constructed coils was evaluated with the 3D-printed torso phantom. A single-shot gradient echo EPI timeseries (repetition time (TR) = 1000 ms, echo time (TE) = 30 ms, flip angle (FA) = 90°, slice thickness (SL) = 5 mm, number of slices (nSL) = 16, matrix (M) = 128 × 64, FOV = (500 × 250) mm$^2$, bandwidth (BW) = 2298 Hz/pixel, number of measurements: 500) was used to measure the peak-to-peak variation in the signal intensity averaged over a 15-pixel square region of interest (ROI). Linear and quadratic trends were removed from the timeseries.\textsuperscript{25} Additional safety tests included transmit power absorption of the detuned array and measurements of potential component heating by gradient eddy currents and by the RF transmit fields.\textsuperscript{26,27} SNR and SMS g-factor measurements were obtained from anthropomorphic torso phantoms. To prospectively evaluate the performance of each coil with the target anatomy, cardiac diffusion tensor imaging (DTI) was performed using each constructed coil and the 38-channel standard coil in a healthy volunteer. Imaging was performed during breathhold with an ECG-gated diffusion-encoded stimulated-echo (STE) sequence, the volume was selected in the phase-encode axis using a slab selective RF pulse.\textsuperscript{28} The acquisition parameters were: TE = 34 ms, SL = 8 mm, M = 128 × 64, FOV = (360 × 200) mm$^2$, in-plane GRAPPA\textsuperscript{3} rate 2, $b$-values = 0 and 500 s mm$^{-2}$, 10 diffusion-encoding directions, and eight magnitude averages (repetitions). Twelve short-axis slices were acquired at the systolic sweet spot (160 ms after the R wave) to mitigate strain effects.\textsuperscript{29,30} SMS excitation was followed by a blipped-CAIPII readout, using a slice-phase-slice zonal excitation scheme.\textsuperscript{28} $MB = 2$ was used with a relative phase axis shift of FOV/3 and a separation between simultaneously excited slices of 48 mm. Variable-rate selective excitation (VERSE) was pre-tuned slightly below the Larmor frequency using three high-power capacitors in series. Fine-tuning was achieved, by carefully incorporating a brass insert screw inside the center of the solenoid former, which reduces the inductance of the cable trap such that the trap circuitry resonates at the desired Larmor frequency. The smaller bazooka cable traps were fine-tuned by finding a suitable combination of the capacitors distribution directly on the shield.

2.5.1 In vivo cardiac diffusion tensor imaging

To prospectively evaluate the performance of each coil with the target anatomy, cardiac diffusion tensor imaging (DTI) was performed using each constructed coil and the 38-channel standard coil in a healthy volunteer. Imaging was performed during breathhold with an ECG-gated diffusion-encoded stimulated-echo (STE) sequence, the volume was selected in the phase-encode axis using a slab selective RF pulse.\textsuperscript{28} The acquisition parameters were: TE = 34 ms, SL = 8 mm, M = 128 × 64, FOV = (360 × 200) mm$^2$, in-plane GRAPPA\textsuperscript{3} rate 2, $b$-values = 0 and 500 s mm$^{-2}$, 10 diffusion-encoding directions, and eight magnitude averages (repetitions). Twelve short-axis slices were acquired at the systolic sweet spot (160 ms after the R wave) to mitigate strain effects.\textsuperscript{29,30} SMS excitation was followed by a blipped-CAIPII readout, using a slice-phase-slice zonal excitation scheme.\textsuperscript{28} $MB = 2$ was used with a relative phase axis shift of FOV/3 and a separation between simultaneously excited slices of 48 mm. Variable-rate selective excitation (VERSE) was pre-tuned slightly below the Larmor frequency using three high-power capacitors in series. Fine-tuning was achieved, by carefully incorporating a brass insert screw inside the center of the solenoid former, which reduces the inductance of the cable trap such that the trap circuitry resonates at the desired Larmor frequency. The smaller bazooka cable traps were fine-tuned by finding a suitable combination of the capacitors distribution directly on the shield.

### 2.6 In vivo image reconstruction

Spatiotemporal registration (STR) was developed to reduce misregistration resulting from respiratory motion.\textsuperscript{28} STR is an automated registration approach based on matching radial intensity profiles over all repetitions. This approach recovers translations and rotations by minimizing the mean square distance between profiles of a $b = 0$ s mm$^{-2}$ reference image and all other diffusion-weighted images.

Subject motion during the STE mixing time causes significant signal cancellation in some images.\textsuperscript{32} These corrupted images must be rejected to maximize the resulting SNR. Following STR, we applied entropy-based retrospective image selection (ERIS) to accept or reject the diffusion-free and diffusion-weighted images.\textsuperscript{28} An entropy measure ($E_M$) based on Shannon’s definition of entropy was calculated from the distribution of signal intensity within the myocardium for each coil and across all repetitions.\textsuperscript{33} Images with $E_M$ within a prescribed range $E_R$ were accepted ($E_M \in E_R$), while those highly affected by artifact were rejected ($E_M \notin E_R$). The acceptance range $E_R$ was independently adjusted for $b = 0$ s mm$^{-2}$ and each diffusion direction.

The myofiber helix angle (HA) was calculated as previously described.\textsuperscript{34,35} Tractography was performed by
numerically integrating the primary eigenvector field into streamlines using an adaptive fifth-order Runge-Kutta approach. Regions of interest were chosen at anatomical locations spanning the ventricular myocardium: the anterior wall, lateral wall, septal wall, and posterior wall.

3 | RESULTS

Simulations suggest that the coil coverage should be extended by approximately 8 cm above and below the heart for highly accelerated cardiac imaging performances. The simulated arrays with shorter coverage show higher peak SNR in the anterior heart region; however, the arrays with larger coverage perform more favorably in the posterior heart region. In terms of encoding power in the allocated target region, the largest coverage (Array 4) shows most favorable g-factors at the critical posterior region of the heart (Supporting Information Figure S2). When taking into account both the baseline SNR and the local noise amplifications from the reconstructed SMS image, all simulated arrays perform almost identically on average in the target region (Supporting Information Figure S3). More critical, however, are the areas where intrinsically low-accelerated SNR can be obtained (the septal and posterior myocardial wall). In this case, the array with the largest coverage (Array 4) shows better SNR values for these critical areas (shorter lower box plot whiskers in Supporting Information Figure S3).

For the three constructed array coils, the \( \frac{Q_T}{Q_L} \)-ratios obtained from the seven incrementally sized coil elements ranged from 3.5 for the smallest to 8.5 the largest coil elements. Upon loading, a slight frequency drop of 0.3 MHz was observed, indicating that the E-field was sufficiently balanced by the distributed tuning capacitance across all loop sizes.

The bench metric results included geometric decoupling, active detuning, and preamplifier decoupling: coupling between nearest neighbor elements, which were critically overlapped, ranged from \(-12 \) to \(-21 \text{ dB} \) with an average of \(-15, -16, \) and \(-13 \text{ dB} \) for the array configurations \( A_{\text{Uni-sized}} \), \( B_{\text{Gapped}} \), and \( C_{\text{Dense}} \), respectively. In addition to these geometric decoupling levels, the elements received an additional reduction of 23 dB from the preamplifier decoupling. The active PIN diode detuning provided >40 dB isolation between the tuned and detuned states. Cable traps showed an RF current suppression of >35 dB at the Larmor frequency.

Figure 5 shows the noise correlation matrices for all three cardiac arrays obtained from noise-only phantom images. The noise correlation across all coils ranged from 0.8 to 44%. The average noise correlation for configurations \( A_{\text{Uni-sized}} \), \( B_{\text{Gapped}} \), and \( C_{\text{Dense}} \) were measured to be 5.2, 4.8, and 5.6%, respectively. The gapped array (\( B_{\text{Gapped}} \)) showed the lowest noise correlations, while the layout consisting of the coil density variation (\( C_{\text{Dense}} \)) showed the highest correlations.

SNR comparisons between the three constructed cardiac arrays and the commercially available 38-channel coil (\( D_{\text{Vendor}} \)) for the non-accelerated cov-RSS combined images are shown in Figure 6. The SNR in the representative transverse plane through the phantom’s heart region shows the typical sensitivity pattern of highly dense array coils. Coil arrays \( A_{\text{Uni-sized}} \), \( C_{\text{Dense}} \), and \( D_{\text{Vendor}} \) showed nearly identical SNR at the center of the slice. However, the gapped array (\( B_{\text{Gapped}} \)) showed 14% less central SNR when compared to the other array coils. We further evaluated the SNR within the 3D region of the phantom’s whole heart across multiple slices, including the specific regions of the mid-lateral wall, the mid-septum, and the apex. The coil array configuration \( C_{\text{Dense}} \) showed an average SNR improvement at the phantom’s heart region of 5% and 18%, when compared to

![Figure 5](image.png)

**FIGURE 5** Noise correlation matrices for all three constructed cardiac arrays obtained from noise-only phantom images. The average noise correlation for \( A_{\text{Uni-sized}} \), \( B_{\text{Gapped}} \), and \( C_{\text{Dense}} \) were measured to be 5.2, 4.8, and 5.6%, respectively.
configurations $A_{\text{Uni-sized}}$ and $B_{\text{Gapped}}$, respectively. In comparison to $D_{\text{Vendor}}$, the constructed coil array $C_{\text{Dense}}$ obtained an SNR increase of 26% in the heart region. Measurements of the specific regions of the mid-lateral wall, mid-septum, and apex showed SNR gains of 5, 2, and 8%, respectively, when layout $C_{\text{Dense}}$ was compared to $A_{\text{Uni-sized}}$. The corresponding SNR increases were 10, 6, and 12% when $C_{\text{Dense}}$ was compared to $B_{\text{Gapped}}$. 

**FIGURE 6** SNR comparison between the three constructed cardiac arrays ($A_{\text{Uni-sized}}$, $B_{\text{Gapped}}$, and $C_{\text{Dense}}$) and the commercially available 38-channel coil ($D_{\text{Vendor}}$) for non-accelerated cov-RSS combined images. The SNR maps were obtained from a representative transverse slice through the phantom’s heart target region.

**FIGURE 7** Comparison of inverse g-factor maps of the accelerated SMS acquisition ($MB = 6$ and $R_{\text{SENSE}} = 2$) for both the simulation and the phantom scan measurements. The measured and simulated SMS g-factors are qualitatively comparable. Bottom: Box plots of the measured g-factors in the heart ROI showing the most favorable encoding capabilities of $C_{\text{Dense}}$. 

Figure 7 shows the comparison of inverse g-factor maps of the accelerated SMS acquisitions for both the simulation and the phantom scan measurements. The g-factor maps were derived from the coil $B_0^*$ sensitivity profiles and noise covariance of the measured data. For the simulated data, the complex sensitivity profiles and the array’s mutual noise resistance matrix was used in order to compute the g-factor maps. We evaluated the arrays’ encoding capabilities for an acceleration scheme using a combined SMS and in-plane acceleration of $MB = 6$ and $R_{\text{SENSE}} = 2$. In this specific scenario, the SMS g-factors in the target heart region were measured at $1.74 \pm 0.31, 1.61 \pm 0.27, 1.32 \pm 0.16$, and $7.9 \pm 0.6$, for the configurations $A_{\text{Uni-sized}}, B_{\text{Gapped}}, C_{\text{Dense}}$, and $D_{\text{Vendor}}$, respectively. Thus, the array configuration $C_{\text{Dense}}$ shows the lowest overall SMS g-factor in the target region with 31 and 22% less noise amplification when compared to configuration $A_{\text{Uni-sized}}$ and $B_{\text{Gapped}}$, respectively. The 38-channel coil setup ($D_{\text{Vendor}}$) does not show satisfactory g-factor performance in the tested acceleration scenario, which generated roughly four-fold higher noise applications when compared to the constructed arrays. The simulated noise amplification showed on average approximately 12% lower SMS g-factors when compared to the measured data obtained from its corresponding coil configuration. Given the complexity of the constructed and simulated coils, this is a reasonably close qualitative match. Relatively, the simulation correctly predicted the differences in the encoding power obtained in the coil array configurations under realistic acceleration settings.

Supporting Information Figure S4 shows the inverse g-factor maps in transverse phantom planes for one-dimensional and two-dimensional in-plane accelerations. The three array coil configurations show nearly identical noise amplifications at the target heart region, with slightly favorable g-factors for array coil $C_{\text{Dense}}$.

Figure 8 shows the comparison of highly accelerated fully encoded diffusion cardiac images between the constructed array coils, obtained from the same subject. Mid-ventricle timecourse signal-to-noise ratio (tSNR) maps were extracted for all coil configurations. Configuration $C_{\text{Dense}}$ showed the highest tSNR relative to $A_{\text{Uni-sized}}, B_{\text{Gapped}}$, and $D_{\text{Vendor}}$. tSNR was also determined at the anterior, lateral, septal, and posterior wall. Notable tSNR improvement was seen with coil $C_{\text{Dense}}$ in comparison to configurations $A_{\text{Uni-sized}}, B_{\text{Gapped}}$, and $D_{\text{Vendor}}$. Figure 9 shows a tractogram depicting the myofiber helix angle for the entire heart, affirming the well-suited performance of coil array $C_{\text{Dense}}$.

4 | DISCUSSION

In this study we designed, constructed, and validated three 64-channel cardiac coils each using a different loop detector configuration. Recently introduced image acceleration techniques, such as SMS, place higher encoding demands on the array coil. To advance the current state-of-the-art, optimized coil configurations targeting highly accelerated cardiac MRI acquisitions were examined. The coil arrays’ performances were evaluated via (1) bench-level measurements such as $Q_L/Q_T$-ratios, tuned-detuned isolation, and neighboring coupling; (2) system-level validations, which included component heating, transmit field interactions, and stability measurements; and (3) phantom performance tests, which were carried out by pixel-wise SNR maps, SMS g-factor maps, and noise correlations. In addition, a simulation pipeline was developed to provide a tool for virtual array coil design with realistic coil-encoding results. A comparison between the simulation results and the performance measurements validated the accuracy of the simulation. Preliminary in vivo measurements were carried out on a healthy subject, and the resulting SNR was compared for all coils tested.

Although higher channel-count arrays in cardiac imaging are well-understood, determining an optimized coil coding structure for cardiac SMS acceleration techniques presents a new challenge. This anatomical target imposes new optimization constraints on array coil design for enhanced SMS encoding capabilities. Therefore, the primary rationale for the construction of the three 64-channel coils was to isolate encoding power using different coil layout configurations. This was accomplished by constructing identical coil formers, which also featured a certain range of adaptability to different body shapes.

In densely populated phased-array coils, the geometric decoupling of the adjacent coil elements is very laborious and time-consuming. Due to increased coupling resulting from the large number of neighboring coil elements, a potential drawback arises from a drop in $Q$-ratio, and thereby, SNR losses. To keep both the capacitive coupling of adjacent elements and the mutually induced eddy current losses low, a 1.2 mm diameter wire was chosen over flat circuit board material for the construction of the conductive loops. The wires allow coil conductors to cross with minimal cross-section area, which reduces the capacitive coupling. Synergistically, the sparse conductor crossover area minimizes copper shading and thus ensures reduced eddy current induced losses in the array’s coil elements. Manufacturing the coil elements from wire also increases the degree of freedom to finely adjust the critical overlap for the prototyped array coils. The $Q_L/Q_T$-ratio for the constructed loop elements from all array coils showed sample noise dominance. This ratio became less favorable for the smaller loop sizes (dia. 50 mm), which were used in the coil configuration $C_{\text{Dense}}$. An average frequency drop of 0.3 MHz upon sample loading indicated some remaining imbalances between the interaction of the coil’s electric
and magnetic fields. While this source of loss is small with a 0.3 MHz frequency shift, it could be further compensated using a higher count of distributed tuning capacitors to balance out the electrical field across a loop element.

However, additional discrete components and solder joints potentially increase the coil’s unloaded losses.

Another source for coupling arises from densely packed preamplifiers and their respective output cables. Therefore,
great care was taken to have short signal transmission lines. Especially in the anterior part of the coil, cable routing was difficult due to long cables between preamplifiers and connectors. Additional cable traps between preamplifier outputs and cable plugs had to be placed to prevent coupling and oscillations. The increased coupling was attributed to the population density of the different loop arrangement. The layout consisting of the coil density variation (C\text{Dense}) showed the highest noise correlations, which was attributed to both the shared voxels in the overlapped regions and the dense multi-branched cable routing. However, the gapped array (B\text{Gapped}) showed the lowest noise correlations, due to less overlapped sample regions and the ability for symmetric cable routing throughout the whole array, where the output coaxial cables can be easily aligned to the virtual ground of the loop elements. The negative impact of higher interelement coupling in layout C\text{Dense} can be compensated by the optimized image reconstruction method by exploiting both the coil sensitivity and the noise correlation information.

The coil configuration C\text{Dense} allowed for the highest sensitivity for both accelerated and non-accelerated imaging at the target heart region, when compared to the coil arrays A\text{Uni-sized} and B\text{Gapped}. The ability of C\text{Dense} to provide higher SNR in the heart is attributed to the higher number of smaller array elements, which are placed on the anterior surface close to the target heart region. In addition, the larger elements, which are located further away from the heart region, allow substantially better SNR contribution at the given target depth, instead of using equally sized smaller elements. The gapped array (B\text{Gapped}) produced lowest baseline SNR in the heart region of all three tested arrays. This is clearly attributed to the non-uniform coverage of the gapped loop array. However, in highly accelerated imaging scenarios, B\text{Gapped} showed better sensitivity overall when compared to the uniformly overlapped array (A\text{Uni-sized}).

Several cardiac coil studies have previously shown compactly covered arrays only for the thorax above and below the heart.\textsuperscript{40-43} Some coil designs were also curved around the left side of the torso.\textsuperscript{44} However, in the context of the recently introduced SMS acceleration techniques, these locally centered cardiac coil designs give suboptimal coverage of the target heart region. In particular, collapsed and off-centered shifted pixels with noisy contributions can be folded into the heart region during the SMS excitation process, thus negatively impacting the image reconstruction. Our simulated guided design approach showed that the coil elements should encompass the upper body radially but also be positioned within approximately 8 cm caudally and cranially of the heart. The latter contributed little to SNR in non-accelerated myocardial imaging. However, the performance of the distal elements suggests they play an important role in highly accelerated SMS cardiac imaging.

The g-factor maps from the combined SMS- and SENSE-accelerated imaging experiments at $MB = 6$ and $R = 2$ showed that the highest noise amplification occurred in the center of the phantom across all three constructed coil arrays. This is expected since the baseline SNR in the center of the phantom is lower compared to the periphery, and the variation in the sensitivity profiles decreases with phantom depth as well.

By enhancing the encoding power in the target heart region due to smaller coil elements, while simultaneously relaxing the encoding performance in the off-target regions using larger elements, the layout C\text{Dense} provides overall favorable encoding capabilities for highly accelerated SMS cardiac imaging. The stronger variation in the sensitivity profiles due to the smaller coil elements in the array configuration C\text{Dense} are highly advantageous during accelerated imaging. This is in agreement with the findings obtained from our full-wave simulations.

A single healthy subject was scanned with each of the constructed experimental coils, as well as with a commercial product coil using combined accelerations of $MB = 2$ and $R = 2$. In this lower range of acceleration, all coil arrays show similar performances, but the trend confirming a favorable coding performance of C\text{Dense} was already visible. The enhanced performance in the septal wall is the best evidence of this improved accelerated SNR performance. The depicted cardiomyocyte tracts compare favorably with the expected anatomy of a healthy subject.

To take cardiac diffusion imaging to a higher level of acceleration, the recently developed methodology of the generalized slice-dithered enhanced resolution (gSlider) holds great promise.\textsuperscript{45} This highly accelerated imaging technique provides significant improvements in both spatial and temporal resolution, and consequently, enhances the accuracy of diffusion-based microstructure indices.\textsuperscript{36,47} This framework has recently been introduced into cardiac diffusion MRI.\textsuperscript{48,49} In the future, we expect the coil design C\text{Dense} in combination with highly accelerated gSlider-SMS cardiac diffusion acquisitions to advance our understanding of human cardiac connectivity and to gain deeper insight into the in vivo heart microstructure at multiple scales.

5 | CONCLUSIONS

We developed, constructed, and validated three 64-channel receive-only coils for SMS-accelerated cardiac MRI at 3 T. SNR and SMS g-factor measurements in phantoms and a healthy subject revealed SNR improvement from the mid-ventricle to the apex and significantly lower SMS g-factor in highly accelerated imaging, when compared to a commercially available 38-channel coil. As SMS techniques in various cardiac imaging methods become more widely available, optimized
highly parallel cardiac arrays become a critical component. The constructed 64-channel array coil with the implemented non-uniform loop density configuration seems to be best-suited for highly accelerated cardiac MRI.

DATA AVAILABILITY STATEMENT
The data that support findings of this study are openly available in github at https://github.com/keyar-ray/cardi-acmri-coil.

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**SUPPORTING INFORMATION**

Additional Supporting Information may be found online in the Supporting Information section.

**FIGURE S1** Simulated coil array layouts to determine a well-suited array coverage for optimized SMS encoding capabilities for accelerated cardiac images. The simulated configuration Array 4 showed most favorable encoding performance in the heart target region, suggesting an important role of the distal elements below and above the heart in SMS accelerated cardiac MRI

**FIGURE S2** Inverse g-factor maps of a representative sagittal slice through the heart. The g-factor measurements were derived from axial SMS accelerations with multiband factors 4, 6, and 8; and additionally combined with in-plane accelerations of $R = 2$ and $R = 3$. The heart region was covered by 60 slices of 2 each. There were 15, 10, and 8 collapsed slices needed to reconstruct the full heart for $MB = 4$, $MB = 6$, and $MB = 8$, respectively. While all array layouts show similar encoding capabilities at the anterior heart region, the simulations suggest that Array 4 provides the best encoding power at the critical posterior region of the heart.

**FIGURE S3** Accelerated SNR for different SMS factors $MB$ and in-plane accelerations $R$ obtained from the whole target heart region (Note, $R = 1$ corresponds to no in-plane acceleration). Comparing the median and the 25th and 75th percentiles, all four simulated array coil arrangements show nearly identical accelerated SNR performances in the heart. The critical heart regions (posterior side) are represented by the lower end of the box plot's whiskers. In this case, the Array 4 with the largest upper body coverage shows the most favorable SNR values.

**FIGURE S4** Measured inverse g-factors maps of a representative transverse slice obtained from a phantom scan. In contrast to SMS-accelerated imaging, the differences in the encoding capabilities of the three constructed array coils are not as prominent for regular in-plane accelerations. The three constructed array coils show nearly identical noise amplifications within the target heart region, with slightly favorable g-factors for array coil C Dense

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