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A 3T Sodium and Proton Composite Array Breast Coil

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Abstract

Purpose—The objective of this study was to determine whether a sodium phased array would improve sodium breast MRI at 3T. The secondary objective was to create acceptable proton images with the sodium phased array in place.

Methods—A novel composite array for combined proton/sodium 3T breast MRI is compared to a coil with a single proton and sodium channel. The composite array consists of a 7-channel sodium receive array, a larger sodium transmit coil, and a 4-channel proton transceive array. The new composite array design utilizes smaller sodium receive loops than typically used in sodium imaging, uses novel decoupling methods between the receive loops and transmit loops, and uses a novel multi-channel proton transceive coil. The proton transceive coil reduces coupling between proton and sodium elements by intersecting the constituent loops to reduce their mutual inductance. The coil used for comparison consists of a concentric sodium and proton loop with passive decoupling traps.

Results—The composite array coil demonstrates a 2–5x improvement in SNR for sodium imaging and similar SNR for proton imaging when compared to a simple single-loop dual resonant design.

Conclusion—The improved SNR of the composite array gives breast sodium images of unprecedented quality in reasonable scan times.

Keywords

sodium; phased array; RF coil; decoupling; breast cancer; metabolic imaging

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INTRODUCTION

Cancer is responsible for a quarter of all deaths in the United States (1). Breast cancer is projected to cause 458,000 deaths worldwide with 1,383,000 new cases worldwide in 2012 (2). Breast cancer is also estimated to include 29% of all new cancer cases in women in the United States during 2012, resulting in 14% of cancer related deaths (1). Early detection and improved treatment have increased breast cancer survival rates in the United States over the past two decades (1). While proton (^1H) magnetic resonance imaging (MRI) is used for cancer detection due to its improved sensitivity when compared to mammography and ultrasound, ^1H -MRI suffers from intermediate specificity which can result in false positive studies leading to unnecessary interventions (3). Because sodium (^{23}Na) concentration is known to increase in malignant lesions when compared to surrounding healthy tissues (4,5), ^{23}Na -MRI may be able to improve specificity, potentially improving evaluation and assessment of breast lesions (6).

Sodium MRI shows promise in characterizing and assessing tumor viability (4,7), cartilage health (8–11), renal failure (12,13), tissue damage following stroke (14), and multiple sclerosis (15). However, in comparison with conventional ^1H -MRI, ^{23}Na -MRI is challenging due to relatively low ^{23}Na concentrations in biological tissues, rapid bi-exponential signal decay, and a low gyromagnetic ratio. Despite these challenges, recent improvements in coil and gradient hardware, the availability of whole-body scanners with high polarizing field strengths, and the development of more efficient pulse sequences have spurred renewed interest in ^{23}Na -MRI (7). These advances have enabled the acquisition of higher quality in vivo ^{23}Na -MRI images than previously possible, often within clinically reasonable scan times (16–18). While ^{23}Na -MRI has become more promising, there is still a need for improved image quality and signal-to-noise ratio (SNR) to make quantitative ^{23}Na -MRI feasible for many of the clinical applications under consideration.

Phased array coils can be used to improve the SNR of ^{23}Na -MRI. This is achieved through simultaneous data acquisition from multiple surface coils which have inherently increased signal sensitivity and limited noise volume by being placed in close proximity to the object or anatomy of interest (19). Specifically designed coil arrays also allow reductions in image acquisition time through the application of parallel imaging techniques (20–22). Phased array coil concepts have been extensively applied to ^1H -MRI coil design (23–26), routinely providing improved SNR and accelerated image acquisition (19) compared to that provided by volume coils (27) or other large coils of similar area (28). However, phased arrays have not been widely used in non-proton imaging, and typically require sophisticated custom hardware for implementation on commercial scanners. Despite these challenges, sites with the capability to support multi-channel non-proton receivers are becoming increasingly common. The first reported non-proton phased array was built for phosphorous imaging in 1992 (29) almost a decade before the first reported ^{23}Na array at 1.5T in 2000 (30). In the past few years, there has been a substantial increase in the number of ^{23}Na coil arrays developed for 3T (31–35), 4T (36), and 7T (37–39). Some of these array configurations are dual resonant, with the ability to image ^1H and ^{23}Na without repositioning the subject (33–37).

This paper presents a new dual resonant breast coil design consisting of a 7-channel ^{23}Na receive array, a larger ^{23}Na transmit coil, and a 4-channel ^1H transceive array. The new composite array design utilizes smaller ^{23}Na receive loops than those typically used in ^{23}Na imaging. Novel methods are also employed to decouple the receive loops from the transmit loops. A novel multi-channel ^1H transceive coil is superimposed on the ^{23}Na receive array, and decoupling between ^1H and ^{23}Na elements is achieved by intersecting the constituent loops to reduce the mutual inductance between the ^1H and ^{23}Na arrays. The performance of the new array design (both SNR and homogeneity) is compared to that of a coil used in prior studies consisting of a single ^{23}Na loop concentric with a single ^1H loop (40,41), both with decoupling trap circuits (42–44). Comparisons were performed both in a phantom and in vivo. The new design achieves excellent ^{23}Na -SNR over the sensitive volume while also providing good image quality for conventional ^1H imaging.

METHODS

The objective of this study was to determine whether a ^{23}Na phased array would be useful in improving ^{23}Na -MRI at 3T. The secondary objective was to create acceptable ^1H images with the ^{23}Na phased array in place.

The new dual-resonant $^{23}\text{Na}/^1\text{H}$ composite array design is first described, including the ^{23}Na receive array loops and circuitry, the ^{23}Na transmit coil, the ^1H transceive array elements, the associated transmit and transmit/receive (TR) switches, and the decoupling techniques employed. Subsequently, the trap coil consisting of a ^{23}Na loop concentric with a ^1H loop is used for comparison. Finally, the phantom and in vivo studies are detailed for assessment of coil performance. All experiments were done on a Siemens TIM Trio 3T MRI scanner (Siemens Healthcare AG, Erlangen, Germany).

$^{23}\text{Na}/^1\text{H}$ Dual Resonant Composite Array

The composite array design consists of a hemispherical fiberglass former with seven ^{23}Na receive loops, four ^1H transceive loops, and a single circular ^{23}Na transmit loop that surrounds the perimeter of the coil (Fig. 1A). A patient friendly support structure is used to position the coil and subject (Fig. 1B) (45).

^{23}Na Receive Array—Prior to construction of the ^{23}Na receive array, multiple 65mm diameter loops were tested with different gauge copper wires and capacitor positions. Wire thicker than 14AWG was not considered feasible for a receive array with small loops, due to the difficulty of using very thick wire for coil construction. The quality factors (Q) (46) of these loops were measured using two stationary decoupled magnetic field probes when the coil was unloaded and loaded (Table 1). The highest Q-ratios were measured using 14AWG and 16AWG wire with two capacitors per loop (Table 1). Because these values were similar, the receive array was constructed using 16AWG wire for its increased ease of use when overlapping the coils on the hemispherical former.

Seven 16AWG circular ^{23}Na receive loops were positioned on a hemispherical fiberglass former (Fig. 1A). Six of the loops are 65mm in diameter, and surround a single 75mm diameter loop placed at the top of the hemisphere (Fig. 2A). Each loop was positioned for

appropriate overlap decoupling. A loaded isolation (S_{21}) of -18dB was achieved between adjacent coils without preamplifier decoupling. The preamplifiers (Stark Contrast, Erlangen, Germany) had a gain of $\sim 35\text{dB}$ and noise figure less than 0.5dB . The change in any receive coil sensitivity (S_{21}) while loaded, measured with two decoupled magnetic probes with and without preamplifiers, was 15dB . Each loop incorporates a matching and a tuning capacitor (Series 1111P_P501X, Passive Plus, Huntington Station, NY; GYA36000, Sprague-Goodman, Westbury, NY), with combined active/passive ^{23}Na decoupling circuitry positioned at the tuning capacitor location (Fig. 2B). The capacitor values are $\sim 180\text{pF}$ and $\sim 1000\text{pF}$ for tune and match respectively. The change in loop sensitivity (S_{21}) between resonant and detuned states is greater than 45dB when loaded, when using two decoupled magnetic probes. The maximum and average off diagonal noise correlation coefficient was 0.49 and 0.29 , respectively.

Combined active/passive decoupling is achieved by placing a crossed diode pair (MA4P7464F-1072T, Macom, Lowell, MA, USA) in series with an inductor that is resonant with the tune capacitor when forward-biased (Fig. 2B). Active and passive decoupling were combined to simplify the coils by using only one decoupling circuit, while maintaining the ability to detune the coil with DC biasing and detune the coil if high power RF is used with improper current biasing. The loops remain resonant during receive when the diodes are not biased. Each loop is connected to an independent DC bias line that provides the $100\text{mA}/10\text{VDC}$ or -30VDC bias current. To avoid biasing the crossed diodes with the negative DC bias, a single diode was added into the bias line so that only positive DC current can be supplied to the loops (Fig. 3A). By eliminating the negative DC voltage at the loop, the crossed diodes are not activated during receive.

Passive ^1H traps are not used in the ^{23}Na receive loops because they would increase coil resistance, reducing sensitivity. Unfortunately, the match/tune capacitor values are too large to implement active ^1H decoupling across those capacitors so no active ^1H decoupling is implemented in the ^{23}Na loops. Larger coils or decreased wire thickness could be used to decrease the capacitor values (Table 1); however, this would likely result in reduced ^{23}Na receive sensitivity and the capacitor values would still be too large to implement effective active ^1H -decoupling.

Each receive loop is attached to a 60cm coaxial cable ($< 1/10$ of the NMR signal wavelength for ^{23}Na at 3T in the coaxial cable). The long coaxial cable allows the receive circuitry to be placed in a convenient location for patient positioning and comfort. A ^{23}Na trap on the coaxial cable shield is used to reduce common mode currents in the long cables, and a phase shifter circuit is used to obtain a 180-degree phase shift between the coil and preamplifier to achieve preamplifier decoupling (19).

^{23}Na Transmit Coil—The ^{23}Na transmit coil consists of five co-axial copper loops equally spaced on a 57mm tall, 178mm diameter acrylic tube (Fig. 2C,D). The loops are connected at their capacitors, making the transmit coil behave like a single-turn solenoid (Fig. 2C,D). The capacitors are distributed along the height of the coil to help evenly distribute the current on the cylinder of the coil. The coil has four tuning capacitor junctions. The transmit coil was tuned inside a large copper shield that simulated the radiofrequency

(RF) shield in the bore of the scanner. The capacitor values are ~500pF and ~660pF for tune and match capacitances respectively.

Decoupling is achieved by breaking the RF current path with serial diodes in two positions. At each of these two positions, four diodes are placed in parallel, equally spaced along the acrylic tube to distribute current along the height of the coil. The coil is designed so that it is resonant when forward-biased with +100mA and detuned when unbiased. When forward-biased, the coil remains tuned and the diodes remain on even when high RF transmit power is used (up to many kilowatts) (47–49). As shown in Figure 2D, a fifth diode was placed antiparallel to the original four diodes to protect them from large reverse bias voltages that can occur during transmit resulting in permanent diode breakdown if not properly forward-biased. Loaded isolation between the detuned transmit coil and resonant receive coils was measured to be –39dB. Loaded isolation between the tuned transmit coil and detuned receive coils was –42dB.

¹H Transceive Array—The inherent drawback of many ¹H and ²³Na coil configurations is that the ²³Na loops have high capacitor values when compared to the ¹H loops, creating low impedance loops at the ¹H frequency. The low impedance ²³Na loops have a shielding effect at the ¹H frequency, resulting in ¹H flux blockage (50), which is similar to the effect of a solid conducting copper loop. When a ²³Na and ¹H loop overlap, the ¹H loop will be affected by the presence of the ²³Na loop to a much greater degree than the ²³Na loop will be affected by the presence of the ¹H loop. When frequency shift, Q-ratio, and SNR are measured for two overlapping in-plane 65mm diameter ¹H and ²³Na loops, the ²³Na loop will have only minor changes (<1%) when a ¹H loop is present, regardless of the center-to-center coil distance. In comparison, the ¹H loop Q-ratio decreases with increasing overlap, and ¹H frequency greatly increases when the center-to-center distance is less than one radius. The shielding effects of the ²³Na loops on the ¹H RF can be reduced substantially by intersecting the ²³Na loops with the ¹H elements (50).

The composite array design employs four local ¹H loops arranged as a ladder network (51), with a minimum loaded isolation between any two loops of the coil measured at –9.5dB. The ¹H loops are positioned over the ²³Na receive array such that the wire elements of the ¹H loops bisect the ²³Na loops (Fig. 1A, 2A). The ¹H loops are mounted 1cm away from the fiberglass former to reduce coupling with the ²³Na loops, to fit over ²³Na loop circuits, and to improve ¹H homogeneity. The tune and match capacitors were adjustable (Series 1111P_P501X, Passive Plus, Huntington Station, NY in conjunction with Series NMA_HV, Voltronics Corp, Salisbury, MD) so that the coil could be tuned/matched without any additional circuitry such as a match inductor (19). The capacitors on the shared rungs were used to minimize coupling between adjacent loops. The capacitor values are ~30pF, ~14pF and ~160pF for the tune, decoupling and match capacitances, respectively.

The ¹H transceive loops contain a crossed diode pair that is forward-biased during ¹H transceive but unbiased during ²³Na transmission and reception (Fig. 2E). The crossed diode pair allows easy tuning/detuning of the ¹H loops for further decoupling between the ¹H and ²³Na loops. The minimum isolation between any ²³Na receive loop and ¹H loop when the coil was loaded was: –20dB at the ²³Na resonant frequency (32 MHz) when the ¹H

loops were tuned, -37dB when the ^1H loops were detuned; -41dB at the ^1H resonant frequency (123 MHz) when the ^1H loops were tuned, and -75dB when the ^1H loops were detuned. The isolation between ^{23}Na and ^1H loops was unchanged regardless of whether the ^{23}Na loops were tuned.

The ^1H loops each use quarterwave cables that, when combined with the quarterwave cable in the TR switches described below, form a halfwave phase shift between the loops and preamplifiers for optimum preamplifier decoupling.

Dual Resonant TR Switching—The scanner provides a single transmit port for both the ^1H and ^{23}Na RF transmit signal. The transmit RF is passively filtered with the use of ^1H and ^{23}Na traps (Fig. 3B) before going into the ^{23}Na transmit switch or ^1H power splitter (Fig. 3C). The filter attenuates the ^1H signal by -39dB on the ^{23}Na output, and the ^{23}Na signal by -47dB on the ^1H output.

After the filter, the ^{23}Na transmit signal passes through a large capacitor before arriving at the cylindrical ^{23}Na transmit coil. A DC bias line inserted between the large capacitor and transmit coil allows the transmit coil to be biased during ^{23}Na transmit with $+100\text{mA}$ and unbiased during ^{23}Na receive.

The ^1H transmit signal is split across four different ports by using 90° hybrid couplers (1J0280-3, Anaren, East Syracuse, NY, NY, USA) to initially divide the signal evenly in half, followed by two more hybrid couplers to divide the signal evenly between four ports (Fig. 3C). The isolation port of each coupler is terminated with 50Ω . Between the initial coupler and one of the secondary couplers, an extra cable length is added to create a 45° phase shift. The four outputs of the power splitter each have equal magnitude but different phase shifts of 0° , 45° , 90° , and 135° . Each output is then connected to a ^1H TR switch.

The ^1H TR switches are different from standard TR switches, in that they incorporate a reverse diode where only a standard forward diode would typically be used, offering improved protection against incorrect DC biasing (Fig. 3D). The TR switches are supplied with a forward current during transmit (DC1 in Fig. 3D) to activate the preamplifier protection circuitry, which consists of a quarterwave cable and a diode that is shorted during transmit. A second DC line (DC2 in Fig. 3D) was added between the TR switch and the loops, so that the ^1H loops can be turned on during ^1H transceive. Large DC blocking capacitors are used to ensure that the DC bias that controls the TR switch is independent from the DC bias that controls the loops.

$^{23}\text{Na}/^1\text{H}$ Dual Resonant Trap Coil

Multiple sodium breast MR studies to date have employed $^{23}\text{Na}/^1\text{H}$ dual resonant trap coils (40,41). To gauge the performance of the composite array design, the composite array is compared to a coil with a single ^1H loop concentric with a single ^{23}Na loop that is similar to coils used previously in dual resonant $^{23}\text{Na}/^1\text{H}$ breast MRI (Fig. 4). Both the ^1H and ^{23}Na loops are single-turn coils built using 10mm wide copper tape placed on a 65mm tall 133mm diameter acrylic tube. The ^{23}Na loop was positioned so that it would surround the center of the breast with a 10mm gap between the ^1H and ^{23}Na loops. The mutual inductance between

the ^1H and ^{23}Na loops is reduced through a single passive resonant trap in each coil (Fig. 4B) (42–44). The traps were tuned prior to insertion into the coil. The ^{23}Na loop has four tuning capacitor junctions with $\sim 390\text{pF}$ per junction and a match capacitance of $\sim 600\text{pF}$. The ^1H loop has three tuning capacitor junctions with $\sim 17\text{pF}$ per junction and a match capacitance of $\sim 42\text{pF}$. The ^1H trap on the ^{23}Na loop has a 41pF capacitor and the ^{23}Na trap on the ^1H loop has a $\sim 780\text{pF}$ capacitor. Loaded isolation measured between the ^1H and ^{23}Na loops was -30.5dB at ^1H frequency and -14.5dB at ^{23}Na frequency. The low isolation at ^{23}Na frequency and larger isolation at ^1H frequency does not completely indicate the effect of each coil on each other, since the presence of the ^{23}Na coil affects the ^1H -SNR far more than the ^1H coil affects the ^{23}Na -SNR (50). Switching for both ^1H and ^{23}Na channels was done using a dual-resonant TR switch (Stark Contrast, Erlangen, Germany).

The trap coil was compared to a similar dual-tuned coil without traps and to single-tuned coils without traps to consider the effects of $^1\text{H}/^{23}\text{Na}$ coil coupling at 3T. The ^{23}Na loop on the trap coil received 80% of the SNR of the single-tuned ^{23}Na coil. The SNR performance of the ^{23}Na loop on the dual-tuned coil that contained no traps had no detectable difference from that of the single-tuned ^{23}Na coil. The ^1H loop on the trap coil had similar SNR when compared to the single-tuned ^1H coil and received 1 to 4 times the SNR of the dual-tuned coil without traps. The primary advantage of the traps is to improve ^1H -SNR by reducing ^{23}Na shielding effects, despite the decreased ^{23}Na -SNR caused by the non-zero impedance of the trap (43). When comparing a single resonant ^1H loop to a ^1H loop on a dual-tuned coil without traps, the dual-tuned coil will have similar ^1H -SNR near the ^1H loop but will exhibit significant reductions in SNR on the opposite side of the ^{23}Na loop (often losing as much as 75% of the SNR).

Comparisons to the composite array were done with the trap coil design due to the trap coil's use in published ^{23}Na breast studies (5,40,41). The trap coil is placed over a hemispherical fiberglass former similar to the one used with the composite array. During experiments, the coil was placed in a support structure (Fig. 1B) (45) such that the subject could lie prone, head first on the scanner table, to reduce respiratory and other motion artifacts. The support structure consists of an acrylic ramp, a flat carbon fiber board that holds the coil, and a head rest. The entire setup is padded for subject comfort during scanning.

Phantom Studies

A fast-gradient spoiled sequence using the 3D cones k-space trajectory was used to image ^{23}Na in a 10cm diameter spherical $\text{NaCl}/\text{CuSO}_4$ phantom (18). The 3D cones sequence consists of spirals that follow a cone-like trajectory, using cones of many different shapes and sizes to fill k-space for a given resolution and field-of-view (FOV). The 3D cones sequence is used to minimize image blurring and signal loss caused by the short $T2^*$ of ^{23}Na as the trajectory achieves more efficient k-space coverage than radial acquisition trajectories and samples the signal before significant $T2^*$ decay. (52)

Phantom studies were conducted with the composite array fully assembled. A spherical phantom with concentrations of 12mM CuSO_4 and 150mM NaCl were used. The 3D cones scan parameters were: $\text{TR}/\text{TE} = 50/0.27\text{ms}$, flip angle = 70° , voxel size = $2.5 \times 2.5 \times 2.5\text{mm}$,

FOV = 22.5cm, cones = 143, shots = 1378, readout time = 9.0ms, averages = 75, with a total scan time of ~1.5 hours. A long scan time was chosen to produce images of very high SNR for the phantom study. A 2D GRE sequence was used to image ^1H with the following scan parameters: TR = 1000ms, TE = 3.28ms, flip angle = 45° , voxel size = $1.0 \times 1.0 \times 3.0\text{mm}$, FOV = $250 \times 125 \times 3\text{mm}$, averages = 4, total scan time of ~8.5 minutes. All scans were acquired in both the sagittal and axial planes and repeated with both the composite array and trap coil. The final SNR values of the composite array were calculated using the root sum-of-squares from images of the individual coil elements normalized by their noise (19). For statistical analysis of the phantom studies, the FOV was segmented into three regions: [1] a hemispherical region expected to contain the breast tissue of interest (referred to as the volume of interest, or VOI), [2] a region of background noise with no signal-yielding tissue, and [3] a signal-yielding region outside the VOI. Signal homogeneity and SNR were evaluated across the VOI.

While resolution is determined by the point-spread-function of a sequence and not determined by a coil, a resolution phantom was scanned to demonstrate the feasibility of this setup. The resolution phantom was 3D printed with varying line thicknesses between 1.25mm and 3.0mm (Fig. 5A,B). The 3D print was inserted into a hemispherical mold that was filled with water, 12mM CuSO_4 and 150mM NaCl. Coronal ^{23}Na images were obtained with 3D cones using the following scan parameters: TR/TE = 40/0.27ms, flip angle = 70° , voxel size = $1.25 \times 1.25 \times 4\text{mm}$, FOV = 22.4cm, cones = 80, shots = 1078, readout time = 8.2ms, averages = 20, with a total scan time of ~20 minutes. A 3D GRE sequence was used to obtain coronal ^1H images with the following scan parameters: TR = 15ms, TE = 5.75ms, flip angle = 25° , voxel size = $0.23 \times 0.23 \times 0.7\text{mm}$, FOV = $180 \times 123 \times 73\text{mm}$, averages = 1, total scan time of ~14 minutes.

Transmit flip angle (B1) maps of the ^1H transceive loops were obtained using the dual angle method (53) using a GRE sequence repeated in both the sagittal and axial planes, with scan parameters: TR = 1000ms, TE = 3.28ms, FOV = $250 \times 125 \times 3\text{mm}$, voxel size = $1.0 \times 1.0 \times 3.0\text{mm}$, flip angle = $45^\circ/90^\circ$, averages = 4, total scan time = 17 minutes.

B1 maps of the ^{23}Na transmit coil were obtained using the phase sensitive method (54,55). Sodium B1 mapping scan parameters were: TR = 100ms, TE = 15ms, FOV = $386 \times 145 \times 108\text{mm}$, resolution = $128 \times 48 \times 36$, voxel size = 3.0mm isotropic, averages = 30, readout bandwidth = 166 Hz/pixel, EPI factor = 3, total scan time = 58 minutes. The high number of averages used for the ^{23}Na phantom experiment was used to achieve high SNR for the comparisons.

Human Imaging Studies

Human imaging studies were conducted after informed consent and with approval of the local institutional review board (IRB). A fast-gradient spoiled sequence using the 3D cones k-space trajectory was used for ^{23}Na imaging in the breast of a normal volunteer (18), with the following scan parameters: TR/TE = 40/0.27ms, flip angle = 70° , voxel size = $1.25 \times 1.25 \times 4\text{mm}$, FOV = 22.4cm, cones = 80, shots = 1078, readout time = 8.2ms, averages = 20, with a total scan time of ~20 minutes. A standard ^1H GRE acquisition was performed to compare ^1H -SNR. The GRE scan parameters were: TR = 11ms, TE = 4.7ms, flip angle =

15°, voxel size = 0.90x0.90x1.2mm, FOV = 172x172x88mm, 1 average, with a total scan time of ~3 minutes. To generate water and fat images using 3-point Dixon, the same ^1H GRE acquisition was performed at TE = 5.75ms and TE = 6.8ms. All scans were repeated with both the composite array and trap coil. Images were acquired in the sagittal plane. The volunteer was moved when switching coils but not between ^{23}Na and ^1H scans. The final images were combined using root sum-of-squares of the individual coil element magnitude images with normalized noise floors (19). The FOV was large enough so that the noise floors could be calculated using slices that contained no signal.

RESULTS

Phantom Studies

^{23}Na -SNR Performance—Within the VOI of the phantom, the composite array had a mean ^{23}Na -SNR of 123 ± 43 and the trap coil had a mean ^{23}Na -SNR of 29 ± 8 . An image comparing the central sagittal slice shows an improvement in ^{23}Na -SNR by a factor of five near the expected location of the nipple, and a factor of three or more across most of the remaining breast volume (Fig. 6). A histogram created from the voxels within the VOI (Fig. 6D) shows that while the spread of ^{23}Na -SNR values is much larger using the composite array, the dramatic improvements in ^{23}Na -SNR are also clearly evident.

The separate 1.25mm thick lines are detectable on the ^{23}Na images using the resolution phantom (Fig. 5C).

^{23}Na Homogeneity—Flip angle maps for ^{23}Na excitation using the ^{23}Na transmit and receive loops are shown in Figure 8. Reasonable homogeneity is observed across the VOI, although some variation is observed, particularly in the center of the coil near the nipple and toward the edges of the breast.

^1H -SNR Performance—The composite array had a mean ^1H -SNR of 725 ± 494 and the trap coil had a mean ^1H -SNR of 530 ± 224 . The ^1H -SNR in the composite array relative to the trap coil improved by roughly a factor of two near the ^1H loops, although it decreased by 20% near the center of the breast phantom (Fig. 9A–C).

^1H Homogeneity—The composite array had a mean ^1H flip angle of $48 \pm 19^\circ$ and the trap coil had a mean ^1H flip angle of $44 \pm 12^\circ$ in the signal-yielding region in the central sagittal slice. The composite array obtains adequate homogeneity in the VOI for GRE images. There is some transmit B1 focusing near the center of the ^{23}Na loops and transmit B1 shielding where the ^{23}Na loops overlap (Fig. 9E).

Human Imaging Studies

^{23}Na -SNR Performance—Sodium SNR improvements similar to those seen in the phantom study were observed in vivo in a normal human volunteer using the composite array (Fig. 10A,C). The fibroglandular tissue in the central sagittal slices of the composite array had a mean ^{23}Na -SNR of 30 ± 18 and the trap coil had a mean ^{23}Na -SNR of 8 ± 3 . Improved ^{23}Na -SNR is evident with noticeably improved depiction of small anatomic

features within the breast (Fig. 11A). The composite array obtains excellent ^{23}Na -SNR over the entire VOI (Fig. 11A).

^1H -SNR Performance—Both the trap coil and the composite array obtain good ^1H images (Fig. 10B,D). The signal-yielding region of the breast in the central sagittal slices had a mean ^1H -SNR of 32 ± 10 using the composite array and a mean ^1H -SNR of 24 ± 7 using the trap coil. The in vivo ^1H -SNR in the composite array was double that of the trap coil on the edges of the breast and similar throughout most of the breast.

DISCUSSION

The composite array obtains a 2–5x increase in ^{23}Na -SNR, which is a substantial improvement over anything that has been obtained in the past by single channel coils used in many ^{23}Na breast studies (4,5,40,41). A 2–5x increase in ^{23}Na -SNR translates to a 4–25x decrease in scan time for a given resolution, which can make a dramatic impact on the use of ^{23}Na -MRI, improving the clinical feasibility of breast ^{23}Na -MRI. The in vivo sodium breast images show a level of detail and structure not previously achieved, demonstrating imaging at a $1.25 \times 1.25 \times 4\text{mm}$ nominal voxel size at 3T in a scan time of only 20 minutes.

The high ^{23}Na -SNR images of the breast were obtained by using a receive array of small receive loops that are well decoupled from a large, homogeneous transmit coil during both transmit and receive. Although the Q-ratios of the ^{23}Na composite array receive loops (Q-ratio = 1.5) would typically be considered low (56), the loops were still very effective in improving ^{23}Na -SNR.

Superimposing and intersecting the ^1H loops with ^{23}Na loops in this array design preserves the high SNR of the ^{23}Na receive array while achieving acceptable ^1H image quality. While the composite array has some ^1H transmit inhomogeneities due to the presence of the ^{23}Na receive array, the sensitive volume is reasonably homogeneous (Fig. 9E). Although not presented in the paper, when the scanner's body coil or a smaller 135mm circular coil was used to image ^1H with the ^{23}Na receive array in place, the ^1H images contained signal focusing and signal voids worse than those shown on the ^1H -B1 maps of the composite array. A larger ^1H transmit array will not be more homogeneous, since most of the inhomogeneity arises from the shielding effects of the ^{23}Na loops. A ^1H ladder coil with more elements should also be investigated as this should improve ^1H -SNR due to the smaller ^1H elements. The quality of the ^1H images produced by our current coil design may not be sufficient to replace a conventional ^1H coil for routine breast MRI, but the ^1H -SNR is currently sufficient for registration between ^{23}Na and ^1H images. A single coil capable of both ^{23}Na and ^1H examination with no sacrifice in ^1H image quality compared to commercially available ^1H coils is the ultimate goal of this work, and this will guide future refinements to the current coil prototype.

The coil has similar safety concerns and benefits when compared to other transceive coils used at 3T, such as transceive birdcage coils that have high RF voltages relatively close to the subject. Following similar design principles for birdcage coils, the electronics are well insulated and are positioned at a reasonable distance from the subject. The increased flip

angles for ^1H near the ^{23}Na loops suggests that only low flip angle ^1H sequences should be used to avoid heating the subject. Future versions of the coil will address the increased RF deposition near the ^{23}Na loops by either including ^1H traps on the ^{23}Na loops, or by using more ^1H transceiver segments to reduce the RF focusing caused by the ^{23}Na loops. The use of a transceiver array will result in less RF deposition into the body overall, due the relatively small size of the transceiver array when compared to the scanner's body coil.

Further improvements to ^{23}Na -MRI using the composite array are still possible. Some improvements include: shorter cables between the receive loops and preamplifiers, resulting in reduced cable interactions with the ^1H signal; improving preamplifier decoupling; and, fiberglass formers that conform better to different breast sizes and shapes. It is uncertain whether the decrease in the transmit ^{23}Na -B1 near the nipple results from the decoupling circuits (Fig. 8). If from the decoupling circuits, placing the decoupling circuits at any other location where less ^{23}Na -SNR is obtained may not be desirable. Better ^{23}Na image quality could likely be obtained using a more optimal multi-coil image reconstruction with noise decorrelation (19,39).

Future work will explore whether higher ^{23}Na resolution can improve detection and evaluation of breast cancer in vivo (6). The improvements in ^{23}Na -SNR will allow better ^{23}Na T1 and T2* measurements for the evaluation of lesions, although quantitation of sodium concentrations is still desirable. The experiments in this study did not demonstrate the accuracy with which quantitative measurements of ^{23}Na concentration could be obtained. The low Q-ratios of the ^{23}Na loops suggest that the loops are relatively insensitive to changes in loading (56), so that field profiles obtained with a phantom may potentially be used for accurate quantitation. If necessary, a ^{23}Na transmit flip angle map could be acquired within a few minutes for transmit field correction (54). Receive field profiles could potentially be corrected using sensitivity encoding (SENSE) reconstruction techniques that use the central regions of k-space to estimate coil sensitivities (39,57,58).

The described breast coil is unilateral. However, implementation of a bilateral ^{23}Na receive array for simultaneous imaging of both breasts would be relatively straightforward. Due to the small diameters of the ^{23}Na loops, the separation between the left and right coil receive elements is expected to be sufficient to avoid any significant loss in performance of a bilateral design vs. the demonstrated unilateral design. Bilateral sodium breast imaging is feasible without additional loss in scan time due to the large number of averages typically performed in ^{23}Na imaging. Increasing the FOV has the same SNR advantage as signal averaging, so in any scenario in which signal averaging is needed, the FOV can be increased without a scan time penalty. For instance, doubling the imaging FOV and reducing the number of averages by a factor of two does not change scan time, resolution, or SNR efficiency.

CONCLUSIONS

This work has demonstrated a 2–5x increase in ^{23}Na -SNR using a novel composite array coil design compared to a single-loop trap coil design, significantly improving breast ^{23}Na -MRI image quality at 3T. The coil demonstrates an array superposition technique that can

improve decoupling between ^1H and ^{23}Na array coils, so that excellent ^{23}Na and good ^1H images can be obtained without repositioning the subject. The improved SNR of the ^{23}Na composite array gives breast ^{23}Na images of unprecedented quality in reasonable scan times. High quality ^{23}Na images of the breast may improve the specificity of breast MRI for the detection and characterization of breast cancer.

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Figure 1.

(A) 23Na/1H dual resonant multi-channel composite array. (B) Complete (unpadded) patient setup on the scanner table, with a 10cm spherical phantom placed in the hemispherical fiberglass former. An acrylic ramp and board supports the patient in the prone position. The RF Tx/Rx circuitry is between the scanner table and the patient support device.

Figure 2.

(A) Top view schematic of the composite array coil layout. The proton coil layout is shown in red / dashed. The sodium loops are shown in black / solid. (B) Circuit diagram of the ^{23}Na receive loops. (C) Schematic of the single channel ^{23}Na transmit coil. (D) Enlarged view of the decoupling circuitry for the transmit coil. (E) Warped schematic of the 1H transceive loops. The capacitors for matching, tuning, and decoupling adjacent loops are marked on one loop as C_m , C_t , and C_d , respectively.

Figure 3.

(A) Circuit diagram of the ^{23}Na receive loops and hardware. The active/passive trap was forward-biased during ^{23}Na transmit, detuning the receive loops. The phase shifter completed a half-wave phase shift between the coil and preamp for preamp decoupling of the loop during the receive portion of the pulse sequence. (B) Traps were used to filter the 1H and ^{23}Na Tx RF signal. (C) The 1H 4-way power splitter used three 90° hybrid couplers and a coaxial cable phase shifter to split the transmit power evenly between each element of the 1H transceive array, with each element transmitting at a different phase. (D) The TR switches for the 1H TxRx loops had DC1 forward-biased during transmit, while DC2 was forward-biased during 1H transmit and receive.

Figure 4.
(A) Picture of the single 23Na and single 1H loop trap coil. (B) Circuit diagram of either 23Na or 1H loop on the trap coil.

A. Layout

B. 3D Print

C. Sodium

D. Proton

Figure 5.

(A) Layout of the lines and gaps in the resolution phantom, with sizes indicated on the schematic. (B) Photograph of the 3D print that was inserted into the phantom solution. (C) Sodium image obtained with the composite array in ~20 minutes. The 1.25mm lines of the phantom are distinguishable on the sodium image. (D) Proton image obtained with the composite array.

Figure 6.

Phantom sodium SNR maps of the central (I) sagittal and (II) axial slices using the (A) composite array and (B) trap coil. (C) Ratio of the composite array to trap coil SNRs. (D) Histogram of sodium voxel SNRs obtained from the VOI within the phantom comparing the trap coil (blue) and composite array (red).

Figure 7.
Noise correlation matrix of the sodium receive array.

Figure 8.

(A) Axial, (B) sagittal, and (C) coronal sodium transmit flip angle maps of the central slices from the composite array.

Figure 9. Proton SNR maps of the central sagittal slice using the (A) trap coil and (B) composite array. (C) Ratio of the composite array to trap coil SNRs. (D) Histogram of proton voxel SNRs obtained from the phantom comparing the trap coil (blue) and composite array (red). Flip angle maps of the (E) trap coil and (F) composite array.

Figure 10.

In vivo (A,C) sodium and (B,D) proton breast images of a normal volunteer obtained using the (A,B) trap coil and (C,D) composite array. All images were acquired with the same volunteer on the same day.

A. Sodium

B. Fat

C. Water

Figure 11.

Multiple slices of in vivo (A) sodium, (B) water, and (C) fat breast images of a normal volunteer obtained using the composite array. The (A) sodium images were obtained in a scan time of only 20 minutes. The (B) water and (C) fat images were obtained using 3-point Dixon.

Table 1

Capacitor values and Q-factors for 65mm diameter resonant loops at 32.60MHz.

Thickness (AWG)	# Capacitor segments	Capacitance per segment (pF)	Q _{unloaded}	Q _{loaded}	Q _{ratio}
20 (thin)	3	391	222	161	1.38
20	2	268	213	160	1.33
16	3	465	263	179	1.47
16	2	295	282	187	1.51
14	3	465	254	176	1.48
14 (thick)	2	327	264	176	1.50

Q-unloaded, Q-loaded, and Q-ratio were measured using two stationary magnetic probes and are displayed for 65mm diameter coils with different wire thicknesses and capacitor segments. The coils were all tuned to 32.60MHz.