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Materials and methods for higher performance screen-printed flexible MRI receive coils

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Abstract

Purpose—To develop methods for characterizing materials used in screen-printed MRI coils and improve SNR with new lower-loss materials.

Methods—An experimental apparatus is created to characterize dielectric properties of plastic substrates used in receive coils. Coils are fabricated by screen printing conductive ink onto several plastic substrates. Unloaded and sample loaded quality factor ($Q_{\text{Unloaded}}/Q_{\text{Loaded}}$) measurements and scans on a 3T scanner are used to characterize coil performance. An experimental method is developed to describe the relationship between a coil's Q_{Unloaded} and the SNR it provides in images of a phantom. Additionally, 3T scans of a phantom and the head of a volunteer are obtained with a proof-of-concept printed 8-channel array and the results are compared to a commercial 12-channel array.

Results—Printed coils with optimized substrates can exhibit up to 97% of the image SNR when compared to a traditional coil on a loading phantom. Q_{Unloaded} and the SNR of coils are successfully correlated. The printed array results in images comparable to the quality given by the commercial array.

Conclusion—With the methods and materials proposed, the SNR of printed coils approaches that of commercial coils while using a new fabrication technique that provides more flexibility and close contact to the patient's body.

Keywords

MRI coils; flexible MRI coils; phased-arrays; NMR probes; screen-printing; plastic substrates

Introduction

In magnetic resonance imaging (MRI), surface receive coils are placed in close proximity to the patient's body in order to provide images with high signal-to-noise ratio (SNR) (1-3). These coils consist of loops of wire divided by in-series capacitors that form a resonating LC circuit. By varying the value of the capacitors, the coils are tuned to resonate at the Larmor frequency (4-6). Large coils have an increased field of view but gives less SNR than smaller

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coils (7). Therefore today, high density arrays with small elements are desirable for large field of view coverage with improved SNR (1).

There are several sources of loss that can reduce image SNR. While the materials used in coil fabrication can contribute to these losses, the patient (i.e., the sample) usually contributes the largest portion of total loss in clinical scenarios at 1.5 and 3 T (8). A common way to characterize the origin of the losses (i.e., the effective resistance) generated by a coil-sample system, is to measure the coil's quality factor (Q) with and without the presence of conductive (i.e., loading) materials – Q_{Loaded} and Q_{Unloaded} respectively. The losses associated with materials used in the fabrication of coils can be inferred from the value of Q_{Unloaded} , whereas Q_{Loaded} provides insight on loss contribution from the sample. Higher Q indicates less loss, which translates into better SNR in images (7).

The current strategy used in conventional coil fabrication is to get the highest possible Q_{Unloaded} by using materials with very low electrical loss such as thick bulk copper traces and high quality porcelain capacitors (9). This method works well in cases where the sample noise is not the dominant factor, as is the case in low frequency MR, NMR systems, or preclinical imaging where both sample and coils are small. However in clinical scenarios where the patient is the largest source of loss, as the coil's Q_{Unloaded} continues to increase, only a marginal increase in SNR is seen (8). With a reduced emphasis on the material's electrical properties, unconventional materials and processes, which offer other advantages, can be leveraged without incurring a large SNR penalty. For example, though less conductive than copper foil, conductive traces printed from solution by additive manufacturing techniques such as screen-printing (10) can be used to create electrically active patterns, producing flexible and lightweight coils.

We have shown in prior work that flexible screen-printed coils with relatively low Q_{Unloaded} could generate images within 20% of a high quality control at the same distance from the sample. (11). Furthermore, the high flexibility of printed coils enables them to surpass conventional coils when they fit better and are closer to the sample (11). Printing also represents a quick fabrication pathway, ideal for unique applications, as was shown previously by inkjet printing coils on polyimide film for use in small animal systems (12) and a 3D printed NMR coil (13). In these applications, the sample is usually small, causing less of the total loss to come from the sample. Unlike imaging larger samples found in clinical scanning, this puts more emphasis on the quality of materials used in the coils, potentially impeding SNR. While these previous results have been promising, image SNR from printed coils still lags behind conventional coils when compared head-to-head and the quality of the materials, in particular the capacitor's dielectric, has been identified as an area of possible improvements (11).

In this work, we develop 8.75 cm diameter octagonal printed coils with Q_{Unloaded} as high as 100 and SNR at 3T as high as 97% of the SNR that a traditional coil displays on a representative phantom. This is achieved by using a coil design where the substrate also acts as the capacitor's dielectric and by careful selection of low-loss plastic substrate material. To this end, an experimental apparatus is created to mimic the printed coil geometry by using copper foil for the traces. This apparatus allows rapid characterization of the dielectric

properties of plastic substrates, so several substrates can be rapidly screened in order to identify the most suitable dielectric for printed coils. The influence of the printed silver trace properties on the Q_{Unloaded} of coils is also investigated. To guide the optimization of the printed coils, we develop an experimental procedure that relates a coil's Q_{Unloaded} to its SNR in images of a representative phantom. In particular, this method provides insights of when improvements in Q_{Unloaded} would no longer yield a significant increase in image SNR. Finally, optimized coils are assembled into an 8-channel array for a 3T scanner. The SNR and noise correlation of this array are characterized on a conductive phantom and its performance is demonstrated by scanning the head of a volunteer. These results are compared to a commercial 12-channel array (Siemens 3T Head MATRIX A Tim Coil). Both arrays produce good quality images, the printed array displaying 6 times higher SNR near the surface of the phantom near the array elements. At the center of the sample, phantom and human, the SNR of the printed array is half of the conventional array. The signal difference at the center of the phantom is mostly a result of the differences in the overall size and configuration of the arrays, rather than the materials used, showing how printing can be implemented to build high quality arrays.

Methods

Fabrication

Coils using a plastic substrate as the dielectric are fabricated by printing the metal features on each side of the substrate as illustrated in Fig. 1 A. The coils are 8.75 cm in diameter with trace width of 5 mm (more details available in Supporting Fig. S1A). The metal traces sandwich the substrate, forming integrated tuning and matching capacitors where they overlap as shown in Fig. 1 B (11,14). The area of overlap between the two plates of the capacitors, as well as the substrate's thickness, determines the resulting capacitance. Adding or removing ink on each side of the substrate controls the capacitor area, and in turn, tunes the coil. A photograph of a printed coil made in this way is shown in Fig. 1 C. The total thickness of the coil is 125 μm , contributing to the overall flexibility of the device. Multi-channel coil arrays can be fabricated by shifting the single coil pattern during the printing process to an adjacent position on the substrate, as shown in the photograph of a printed 4-channel flexible array in Fig. 1 D. The critical overlap distance is determined by connecting two tuned printed coils to a network analyzer and measuring what overlap minimizes their $|S_{12}|$ interaction (1,15,16). The overlap between each single coil element in the array is 11.7 mm, which results in approximately 29 cm of lateral coverage for a 4-channel array.

Dielectric Substrates

In the coil configuration used here, the loss characteristics of the plastic substrate directly impact the quality of the printed capacitors, and in turn, image SNR. Dielectric loss is reported for most plastic substrates at limited standard frequencies, but rarely at frequencies common to clinical MRI (i.e., 64, 123.3 and 127.7 MHz). To characterize the dielectric properties of several materials, flat plastic substrates are clamped between two 7 mm thick acrylic plates with 70 μm thick copper traces (resistivity = $1.68 \times 10^{-6} \Omega\text{-cm}$) in the same geometry as our coil (Fig. 2 A). Acrylic was chosen for the plates for its ease of manufacturing, isotropic electrical properties, ability to withstand clamping forces, and

transparency - to aid in positioning of the coils, although other low dielectric constant mechanically stable materials could be used.

For each substrate tested, the lengths of the metal traces are adjusted to obtain coil resonance at 123.3 MHz. To measure Q_{Unloaded} , the coils are placed at least 50 cm away from any conductive material and the $|S_{12}|$ response from two lightly coupled probes is recorded (7). Probes were decoupled by separating them several diameters apart to reduce baseline $|S_{12}|$ response below -70 dB. Light coupling to the coil under test was ensured by keeping any $|S_{11}|$ or $|S_{22}|$ response above 100 mdB. The quality factor, Q , is measured on a calibrated Agilent E5061B ENA network analyzer with 1601 points averaged 16 times, centered at the Larmor frequency (123.3 MHz) with the vector network analyzer set to a frequency span of 25 MHz. It is important to note that the Q measured on a network analyzer in this way is dependent on the position, orientation, calibration, frequency sweep/span, and day-to-day variation of the experimental setup, including hard to control for humidity and temperature dependent film parameters. Our experimental setup was characterized to have a repeatability of 5% by measuring our control coil several times on different days.

The characterized plastic substrates are selected based on their compatibility with the printing process. The substrates studied are commercially available in a thickness between 50-150 μm , able to withstand at least 125°C during annealing, and mechanically stable enough to be handled during processing. The plastic substrates tested are polyethylene (PE), polyethylene terephthalate (PET), polyethylene-naphthalate (PEN), polyethylene-naphthalate with an adhesion layer (PQA1), kapton 100HN (PI-N), kapton 200FN919 (PI-F), polyphenylene sulfide (PPS), polyetherimide (PEI), polyether ether ketone (PEEK), polytetrafluoroethylene (PTFE), cleanroom paper cellulose with acrylic binder (CR-P), tyvek 1025D (Tyvek D), tyvek billion 4173DR (Tyvek DR), and pyralux AP (PI-AP).

It should be noted that the acrylic plates of the testing rig contribute a small amount to the overall coil capacitance since their dielectric constant is higher than air. To characterize this interaction, completed coils with printed conductor were clamped in the test rig without the copper film to measure any shift in resonance frequency or Q . Resonance frequency shifted down 1.4 MHz on average, implying an increase of ~ 0.25 pF from the testing rig. Quality factor with and without the acrylic plates changed less than 0.3%, implying any contribution from the acrylic plates is negligible.

Conductive Inks

Three different silver-based conductive inks are compared. Micro-flake Creative Materials 118-09, nanoparticle InkTek PA-010, and micro-flake DuPont 5064H (with resistivity of $7.99 \times 10^{-5} \Omega\text{-cm}$, $3.94 \times 10^{-5} \Omega\text{-cm}$, and $1.84 \times 10^{-5} \Omega\text{-cm}$ respectively as measured by 4-point probe). Prior to printing, the substrates are preheated to 140 °C for 10 minutes and cooled again at room temperature to relieve any stress and prevent distortion in future processing steps. Next, the conductive inks are screen-printed (ASYS Group ASP01M, Dornstadt: Germany) through a 400 mesh-count stainless steel mesh onto the flexible substrate and annealed at 140 °C for 15 min. This procedure is repeated on the reverse side of the flexible substrates, completing the coils. Printing multiple layers also allows variation of the thickness of the conductive trace, with each layer adding 5 μm to the printed conductor.

Coils with conductive trace thicknesses 5-35 μm are examined to characterize the relationship between conductive film thickness, coil Q_{Unloaded} , and mechanical stability. Conductive trace thickness was measured using a stylus profilometer (Dektak 6M Stylus Profilometer, Veeco: Oyster Bay, New York). This characterization is performed with the highest conductivity ink (DuPont 5064H) on low-loss (PTFE) and higher-loss (PET) substrates to show the relative benefits of conductive ink optimization.

Coils are made for SNR measurements by printing DuPont 5064H silver ink on several types of substrates. During tuning, the coil's resonance frequency and input impedance are monitored with the same VNA as previously described. The coils are tuned to the Larmor frequency of protons of our 3T scanner (123.3 MHz - 3T Siemens TIM Trio, Erlangen: Germany) and an input impedance of 50 Ω when loaded (6).

Imaging

To serve as reference, control coils are fabricated by soldering high quality ATC capacitors on 35 μm -thick copper traces on Pyralux AP substrates. The geometry of the control coil is identical to the printed ones. To predict the SNR of printed coils without scanning, Q_{Loaded} is measured and compared to that of a control coil. To load the coils, they are placed directly on top of a water filled 7L cubic phantom doped with 3.356 g/L $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$ and 2.4 g/L NaCl for conductivity of 0.68 S/m at 123 MHz. The acrylic walls of the phantom were 6.4 mm thick.

To get a direct measurement of SNR, the coils are used to image the conductive phantom on a 3T scanner (3T Siemens TIM Trio, Erlangen: Germany). A 2D gradient echo sequence (GRE) with TE of 10 ms, TR of 438 ms, FA of 25°, and slice thickness of 5 mm is used to scan. The field of view is 200×200 mm^2 with resolution of 256 phase encodes and readouts. Printed portions of the coil were clamped into a test board which contained a PIN diode to deactivate the coil during the transmit phase of each scan (Supporting Fig. S1B). This board is connected to a gateway box (Stark Contrast, Erlangenm Germany) containing preamplifiers via half-wavelength long coaxial RG-316 cable. Several cable traps are placed on the half-wavelength coaxial cable to reduce coupling of the cable to the scanner's body transmit coil. The SNR of images is measured by dividing the average signal (i.e., the average magnitude from a 4×18 cm region of pixels along the entire length of the phantom) by an estimate of the noise standard deviation. Pixels near the center of the coil are chosen to measure the signal to avoid clipping artifacts from saturated regions near the edges of the coil where the conductor traces are. Image analysis is performed on raw data files from the scanner, rather than exported images to avoid compression and signal clipping artifacts. The noise is calculated from an area with no signal containing at least 2800 points at least 5 pixels from the edge of the image with no streaking artifacts. All SNR measurements are normalized to images taken with the traditionally made high quality ($Q_{\text{Unloaded}} = 400$) coils. To show the benefit of using the substrate as the dielectric compared to printing dielectric inks, a coil was fabricated using the same Creative Materials 116-20 dielectric ink as previously reported (11).

Coil Array

An 8-channel printed array is fabricated from printed elements and compared to a 12-channel commercially available coil (Siemens 3T Head MATRIX A Tim Coil, Erlangen: Germany). The commercial array has elements approximately 25 cm long and 8 cm wide in a single row surrounding the enclosed cylindrical volume.

To simplify processing, and to be consistent with the single coil experiments, the same printed coil dimensions used for single coil characterization were kept. The entire printed coil array is laminated in 125 μm of PTFE plastic to prevent skin contact, provide flame resistance, and maintain high breakdown strength for any DC bias that could exist on the coil. The printed array connects to the scanner in the same manner as single coil elements except for the interface between printed and non-printed portions. The printed films were cut to expose conductive leads that are folded over sections of RG-316 and copper braid, locked into place using a brass ring crimp (a detailed photo is available in supporting Fig. S1C). The entire crimped junction was then laminated in PTFE with the rest of the coil. Each of the two gateways accepts 4 channels, providing 8 channels in total. The commercial array uses the built-in commercial preamplifiers and connectors that can accommodate additional channels. Images are taken of a 3.8 L cylindrical loading phantom doped with 4.5 g/L NaCl and 2.9 g/L of $\text{NiCl}_2 \cdot 6\text{H}_2\text{O}$. The coils were placed directly onto of the phantom that had a 6 mm thick acrylic wall containing the solution. SNR maps are obtained by first computing the noise covariance matrices from noise images obtained using no radiofrequency (RF) power or gradients, which are used in a Roemer optimal combination, i.e., SENSE with no acceleration factor (1,17). For this analysis, coil sensitivity maps are estimated with ESPIRiT (18). The noise correlation coefficient matrices are also calculated from noise images obtained using no RF power or gradients and computed using equation 1 in Duensing et al. (19).

Volunteer Imaging

To scan a volunteer, a 3D ultrafast gradient echo sequence is used with 1.1 mm cubic voxels. TR and TE were 1900 ms and 2.51 ms respectively. A 90° flip angle is used with 246 phase and frequency encodes and no averaging. The 8 channel array was wrapped around the head of the volunteer with a 3 mm foam matt placed in between coil and skin to reduce capacitive coupling. Human experiments were performed with approval of the Internal Review Board at the University of California, Berkeley.

Predicting SNR

An experimental method is developed to relate the Q_{Unloaded} of a coil with its SNR in an image of a homogeneous phantom with similar conductivity to human tissue to simplify testing procedures.

If the Q_{Unloaded} and Q_{Loaded} on a particular phantom are known, the efficiency of a coil can be predicted using equation 5.35 from Link in Rudin et al. (20) shown here in equation 1 where SNR_0 is the total available, (i.e. intrinsic) SNR. (21,22)

$$SNR = SNR_0 \sqrt{1 - \frac{Q_{Loaded}}{Q_{Unloaded}}} \quad (1)$$

However, when evaluating materials, particularly if they are in a non-conforming test rig as described here, it is difficult to place it in close contact with phantoms, especially if a curved phantom is desired. While it is possible to measure both $Q_{Unloaded}$ and an approximate Q_{Loaded} with a coil made with the test rig, it is more convenient to only measure $Q_{Unloaded}$. To do this, an alternate means of coil comparison is needed.

The Q_{Loaded} of a coil-sample system is related to the total loss of the system, with contributions from the coil materials - R_{Coil} - and the sample - R_{Sample} . $Q_{Unloaded}$ is only influenced by R_{Coil} . Equation 2 and 3, taken from equation 5.30 and 5.31 from Link in Rudin et al. (20) describe this relationship where L is the inductance of the coil, and ω is the resonance frequency (7,8,23).

$$Q_{Loaded} = \frac{\omega L}{R_{Coil} + R_{Sample}} \propto \frac{1}{R_{Coil} + R_{Sample}} \quad (2)$$

$$Q_{Unloaded} = \frac{\omega L}{R_{Coil}} \propto \frac{1}{R_{Coil}} \quad (3)$$

The SNR of an image obtained from a coil can then be related to its Q_{Loaded} , as described by equation 5.33 from Link in Rudin et al. (20) (7,17,24).

$$SNR \propto \sqrt{Q_{Loaded}} \quad (4)$$

There are a multitude of factors that impact image SNR and which play into the coefficient of proportionality implied in the equation above. To determine a coil's performance without having to estimate this coefficient, one strategy is to compare it with a geometrically identical coil known to have low-loss (i.e., a coil which has an efficiency of approximately 99% or greater). If two coils have the same geometry and are in the same position, then L and R_{Sample} are not significantly changed. Making this assumption, it is then possible to predict the relative SNR in images obtained from a test coil with the SNR obtained from a high quality control coil as shown equation 5.

$$Relative\ SNR\ (\%) = \frac{\sqrt{Q_{LoadedTest}}}{\sqrt{Q_{LoadedControl}}} \quad (5)$$

The Q_{Loaded} of a coil can be estimated from its Q_{Unloaded} with equation 6

$$Q_{\text{Loaded}} \propto \left(\frac{1}{Q_{\text{Unloaded}}} + R_{\text{Sample}} \right)^{-1} \quad (6)$$

Measurements of Q_{Unloaded} , Q_{Loaded} , and SNR of several completed coils allowed estimation of the sample loss factor, R_{Sample} , used to represent the loss contribution from the 7L 0.68 S/m water based phantom. Finally, by combining equations 5 and 6, the relative SNR of a test coil to that of a control coil can be simply estimated from the measurement of its Q_{Unloaded} .

$$\text{Relative SNR (\%)} = \sqrt{\frac{(Q_{\text{UnloadedControl}})^{-1} + R_{\text{Sample}}}{(Q_{\text{UnloadedTest}})^{-1} + R_{\text{Sample}}}} \quad (7)$$

Results

Ink and Substrates Characterization

Figure 2B compares the Q_{Unloaded} of all substrates tested in the copper/acrylic testing rig as well as the Q_{Unloaded} of some printed coils made from the same substrates with the three silver inks (the results are tabulated in supporting Table ST1). The Q_{Unloaded} in the testing rig ranges from 10 for CR-P to 355 with PTFE, which is very close to the Q_{Unloaded} of control coils made with high quality porcelain capacitors. However for printed coils, the Q_{Unloaded} for all 3 silver inks do not reach values higher than 100 despite using substrates that achieve high Q_{Unloaded} in the copper test rig. For example, the PTFE film has the highest Q_{Unloaded} in the copper test rig, but when a printed conductor is used, the Q_{Unloaded} is 100, which is not significantly higher than a printed coil made with PI-F, PEI, or PEEK.

As the thickness of the silver ink was increased from 5 to 35 μm on PET and PEEK substrates (illustrated in Fig. 3 A), a diminishing return in coil Q_{Unloaded} was observed above 20 μm of ink, as shown in Fig. 3 B. Furthermore, it was seen that the silver thickness has a more dramatic influence on Q_{Unloaded} with the PTFE substrate than PET due to the higher dielectric loss of PET.

Image SNR

The SNR extracted from images from printed coils on a conductive phantom – illustrated in Fig. 4 A – are summarized in Fig. 4 B. Figure 4B shows that all coils improve significantly over the printed coil previously reported with a solution processed dielectric (11). Furthermore, all coils are within 8% percent of SNR of the high quality control coil. The diminishing effect a coil's Q_{Unloaded} has on SNR in the conductive phantom images is illustrated in Fig. 4 C, where Q_{Unloaded} of printed coils is plotted against SNR relative to the control coil (tabulated in supporting Table ST2). The results from our model are also shown

in Fig. 4 C with the sample fitting parameter, R_{Sample} , equal to 0.087 ($R^2 = 0.97$). The values obtained from our predictions are in close agreement with the value predicted from Link in Link in Rudin et al. (20).

Printed Array

To clearly compare the SNR between conventional and printed coil arrays, both arrays are first used to image a homogeneous conductive phantom, as shown in Fig. 5 A. The SNR maps produced, shown in Fig. 5 B, illustrate that the SNR for the printed coil is highest at the surface of the phantom. This is especially true for regions near the conductive traces of each element, giving the printed array 6 times higher SNR than in the commercially available array. However, near the center of the phantom, the printed array produced 49% of the SNR compared to the commercial array.

The noise correlation coefficients for each array are characterized by noise images obtained from scans with no RF excitation. Figure 5 C shows the noise correlation coefficients for both the printed and the control array, with the average correlation coefficient for the printed array being 0.105 and the control array 0.041. Coil overlap coupling capacitance was measured to be 21.4 pF. The isolation ($|S_{12}|$) between adjacent coil elements was measured to be 17.2 ± 2.7 dB with a non-adjacent next nearest neighbor value of 20.4 ± 2.3 dB.

Finally, the 8-channel printed array is wrapped around the head of a volunteer as shown in Fig. 6 A. Figure 6B shows brain images taken with the printed 8-channel array and the 12-channel commercial array. Images produced from each array are similar in quality, with SNR trends similar to the ones shown on the phantom in Fig. 5 B.

Discussion

Although previous studies have shown the feasibility of imaging animals (12), and humans (11) with printed coils, further understanding of the materials available to an array designer can increase performance. Printing receive coils for clinical imaging is unique because the primary noise in the system usually comes from the sample, not the coil itself (8). Identifying when increasing coil quality will have diminishing impact in a particular use case can free an array designer to add functionality by choosing a different material set.

Ink and Substrates Characterization

The testing apparatus allows the characterization of Q_{Unloaded} for coils made using several plastic substrates as dielectric material in coil capacitors. Since copper foil is used for the conductive part of the loop, it can be assumed that the coil losses are dominated by the dielectric losses, and therefore Q_{Unloaded} is an indirect measurement of the dielectric quality of each tested material. The gap seen in Q_{Unloaded} between the printed and test rig coils in Fig. 2 B indicates that, for most substrates, the Q_{Unloaded} of printed coils is limited by the conductivity of the printed conductors. However, on the substrates that display the most loss, such as PET, changing the silver ink only raises Q_{Unloaded} from 38 to 57 indicating that the low-values of Q_{Unloaded} stem from losses of both the substrate the conductive ink. Overall, these results indicate the potential of plastic substrates to be used as dielectric in very high Q_{Unloaded} coils. However, when printed conductors are used, they tend to limit the maximum

values attainable for Q_{Unloaded} . As a result, other substrate properties, such as mechanical or thermal stability, can be taken into account to select the best suitable substrate for a particular application of printed coils. Reducing the losses generated by the printed conductors represents a major avenue for improving the Q_{Unloaded} of printed coils.

One method to boost the conductivity of printed silver traces, and in turn coil Q_{Unloaded} , is to increase the trace thickness - as illustrated in Fig. 3 A. As is shown in Fig. 3 B, very thick films of printed conductor do not drastically improve Q_{Unloaded} because of the diminishing contribution on the trace's cross sectional area, and because the skin effect limits the effective thickness of material which contributes to charge conduction. A slight decrease in Q_{Unloaded} is even seen for very thick conductive silver films on PTFE, most likely due to the occurrence of the film cracking and thus reducing conductivity. Based on those observations, a thickness of 15 μm of silver printed from the DuPont 5064H ink is chosen as the optimal fabrication parameter for subsequent coils to maintain sufficient SNR while avoiding film cracking.

Image SNR

All coils made with flexible substrates show higher Q_{Unloaded} and image SNR than the fully printed dielectric coils described in the previous study. This difference is attributed to the low loss the substrates display and to the higher loss in binders present in most types of printable dielectric ink.

The results shown in Fig. 4 C of the single printed coil testing along with the trend from our model, suggests that coils with a Q_{Unloaded} of 100 or more would only see a marginal increase ($\sim 3\%$) in SNR compared to the control. Even a printed coil made of PET, which has a Q_{Unloaded} of 58, displays 92% of the SNR of the control coil. To put it in perspective, a 3-8% decrease in SNR is roughly equivalent to a coil being offset from the body by only 1-1.8 cm (11). In a worse case non-clinical scenario where the imaged sample would be non-conductive (and thus non-loading), the PTFE coil with Q_{Unloaded} of 100 would still show 50% of the SNR of the control coil. Therefore, even with improvements in the properties of printed materials, only marginal improvements in SNR can be expected. However the high flexibility of printed coils may provide patients with more comfort while consistently delivering images with high SNR.

Printed Array

From our proof-of-concept printed 8-channel array, fabricated using DuPont 5064H and a PEEK substrate, we were able to produce SNR and noise correlation numbers comparable to a commercially available 12-channel array. Other substrates such as PTFE, PEI, and PI-F show better coil performance, but lack mechanical and environmental stability preventing their use in arrays that are suitable and safe for human imaging without additional steps. Specifically, PTFE had poor film adhesion causing cracking to occur at any printed/non-printed interface that existed between the coil substrate and the cables used for connection, while PEI and PI-F absorbed enough moisture from the air to significantly change tuning over time.

The close proximity to the sample and the smaller element size of the printed array gives higher SNR at the surface, but did not give as much penetration as the 12-channel commercial array, achieving only 49% of the SNR at the center of the phantom. The differences in the SNR maps for the printed and control coils are most likely due to the differences in the surface areas between the arrays, positioning, layout, and preamplifier noise figure.

While the printed array shows higher noise correlation than the control array, the results shown here are comparable to literature on 3T surface coil arrays (25-27). The higher correlation coefficient is attributed to the close proximity to the sample in addition to the relatively large mutual capacitance between channels. We have identified other pathways for improvement of printed arrays, such as better cable management, optimized coil placement, and additional input baluns. Such optimizations would produce lower noise correlation numbers, but is beyond the scope of the work that is presented here.

The image quality is similar with both coil arrays despite the vast difference in component quality and number of channels. While this is not a rigorous comparison between the two coil sets (e.g., coil shape, layout, and connection to scanner effected overall performance) this illustrates how close the performance of printed coils can be to conventional, commercial systems.

Conclusion

As materials and methods for printing continue to improve, the characterization techniques described here can serve as practical tools to enable quick identification of highly performing flexible substrates and conductors for printing MRI receive coil arrays. In this work, we demonstrated a printed coil with a SNR within 3 % of a conventionally fabricated control coil. The tests performed highlight the importance of optimizing materials used in printed coils. Using this approach, we have shown that receive coil arrays fabricated using printing techniques are comparable to commercially available arrays, while offering new form-factors and lightweight construction.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1.

A. Illustration of the printed coils fabrication process. Conductive features are printed on each side of the substrate to form tuning and matching capacitors. **B.** Cross-section schematic of coil capacitors formed between two printed metal features and the plastic substrate along with the equivalent circuit of the printed portion of our receive coil. **C.** Photograph of a single element printed flexible coil. **D.** Photograph of coil array fabricated using the process flow illustrated in A. Single coil elements are shifted to adjacent positions to form a multi-channel array. Transmission lines connecting the coils to the MRI scanner are also shown coming out of the bottom of coil array.

Fig. 2.

A. Illustration of the testing rig developed to characterize the dielectric loss of flexible substrates prior to printing. This setup mimics the geometry of the printed coils. The stack of materials is highlighted in lower right. **B.** Summary of Q_{Unloaded} of coils obtained with flexible substrate from the testing rig and printed conductive inks. Black points are results from testing rig with copper traces. Blue, green, and red points are from coils printed with CM 118-09, PA-010 and Dupont 5046H respectively. † Denotes the use of the dielectric ink from previous work (10). PTFE is identified as the dielectric with the lowest loss, as it has a highest Q_{Unloaded} in the testing rig at 355 while PET shows the lowest at 87.

Fig. 3.

A. Illustration of a printed coil with inset showing an increase of the printed layer (grey) thickness, obtained by printing additional layers of conductor. **B.** Q_{Unloaded} of coils made with increasing thickness of DuPont 5064H silver ink for highest (PTFE) and lowest (PET) performing flexible substrates in blue and red.

Fig. 4.

A. Illustration showing a coil on a 7L conductive phantom for measurements of Q_{Loaded} and image SNR. A slice of phantom is highlighted with an overlay of a scan showing the sensitivity profile for printed coils at 3T. **B.** Relative SNR of printed coils made with several substrates and Dupont 5064H silver ink loaded with a conductive phantom and measured at 3T as a percentage of the control. Measurements were taken at center of the coils along the length of the phantom. **C.** SNR relative to control vs. Q_{Unloaded} for all coils as predicted from bench top measurements (blue) and on 3T scanner (red). Black dotted trend-line is fit of data using model.

Fig. 5.

A. Illustration of a cylindrical conductive phantom (green) showing the placement of (top) 8-channel printed array and (bottom) a commercially available 12-channel head array. **B.** SNR maps of the phantom using (top) printed coil array and (bottom) commercial array. Blue and red dotted lines highlight the location of SNR cross-sections shown in the plot below. SNR is normalized to the max SNR seen with the printed array. The printed array shows higher SNR near the surface and slightly lower in the center compared to the commercial array due to different element geometry and layout. **C.** Noise correlation coefficient matrix for (top)

printed array and (bottom) commercial array from images without RF excitation. The average off-diagonal coefficient for the printed matrix is 0.105 (min: 0.007 and max: 0.357) while the commercial array is 0.041 (min: 0.002 and max: 0.113).

Fig. 6.

A. Photograph of two printed 4-channel arrays combined to make an 8-channel array and placement on a volunteer. **B.** Axial slice of a volunteers' brain using (left) the printed coil array and (right) the commercially available 12-channel head array. Similar to phantom images, printed coil shows higher SNR near surface of the brain and slightly lower in the center compared to commercial 12-channel array.