Novel Inductive Decoupling Technique for Flexible Transceiver Arrays of Monolithic Transmission Line Resonators

Roberta Kriegl, 1,2,3 Jean-Christophe Ginefri, 1 Marie Poirier-Quinot, 1 Luc Darrasse, 1 Sigrun Goluch, 2,3 Andre Kuehne, 2,3 Ewald Moser, 2,3 and Elmar Laistler 2,3*

Purpose: This article presents a novel inductive decoupling technique for form-fitting coil arrays of monolithic transmission line resonators, which target biomedical applications requiring high signal-to-noise ratio over a large field of view to image anatomical structures varying in size and shape from patient to patient.

Methods: Individual transmission line resonator elements are mutually decoupled using magnetic flux sharing by overlapping annexes. This decoupling technique was evaluated by electromagnetic simulations and bench measurements for two- and four-element arrays, comparing single- and double-gap transmission line resonator designs, combined either with a basic capacitive matching scheme or inductive pickup loop matching. The best performing array was used in 7T MRI experiments demonstrating its form-fitting ability and parallel imaging potential.

Results: The inductively matched double-gap transmission line resonator array provided the best decoupling efficiency in simulations and bench measurements (<−15 dB). The decoupling and parallel imaging performance proved robust against mechanical deformation of the array.

Conclusion: The presented decoupling technique combines the robustness of conventional overlap decoupling regarding coil loading and operating frequency with the extended field of view of nonoverlapped coils. While demonstrated on four-element arrays, it can be easily expanded to fabricate readily decoupled form-fitting 2D processes with an arbitrary number of elements in a single etching process. Magn Reson Med 000:000–000, 2014. © 2014 Wiley Periodicals, Inc.

Key words: surface coil; coil array; transmission line resonator; mutual decoupling; pickup loop matching; ultrahigh field

INTRODUCTION

Many biomedical applications of MRI on humans and small animals require high image resolution, together with high signal-to-noise ratio (SNR) and reasonably short acquisition time. From an instrumental point of view, these requirements call for highly sensitive radiofrequency (RF) probes adapted in size and shape to the region of interest. To achieve this, several strategies known to improve detection sensitivity and speed can be combined, such as operating at high field, using arrays of small sized RF coils, and using flexible materials for coil fabrication to enable form-fitting of the coil to the target region.

Applying a higher static magnetic field strength is one of the most common strategies in NMR research to increase the amount of detectable nuclear magnetization, and thus, to achieve high spatial resolution with sufficiently high SNR (1–3). Currently, the highest field strength available for human whole-body MRI is 9.4 T, with head scanners up to 10.3 T.

The use of small surface coils in the regime of sample dominated noise enables strong sensitivity improvement because it provides both, stronger magnetic coupling with the sample and noise reduction due to the smaller volume of tissue visible for the coil (4,5).

The concept of coil miniaturization is of particular interest for high field (3 T ≤ B0 < 7 T), and ultrahigh field (≥ 7 T) applications and has been used to improve the SNR in several studies (6,7), as the coil size defining the threshold between sample and coil noise domain decreases with increasing frequency. For instance, at 300 MHz, i.e. the proton Larmor frequency at 7 T, this threshold should be reached for a coil diameter of 12 mm (8).

Mechanical flexibility of the RF detection system is advantageous for imaging samples with nonplanar surfaces or anatomical regions that can vary in shape and size from one subject to the other. Form-fitting RF coils to the sample improves the magnetic coupling between sample and coil, provides a higher filling factor and better RF transmission efficiency, and thus, leads to a significant SNR gain (9).

RF coil arrays have several advantages over large single element coils for imaging large anatomical regions. In receive mode, arrays can achieve a large field of view (FOV) while preserving the intrinsically high detection sensitivity of small coils (10). Combined with parallel imaging techniques they allow for accelerated image acquisition (11,12). In addition, in transmit mode, coil arrays give access to B1+ shimming (13,14), enabling homogenization or specific shaping of the transmit RF field. This provides a way to compensate for transmit field inhomogeneity occurring at high field due to the
shorter wavelength at high Larmor frequency, which usually induces undesired spatial variation of image contrast and intensity. Finally, the concept of parallel excitation permits to shorten the duration of applied spatially selective RF pulses, thus further, speeding up the MRI experiment (15,16).

While efficient principles and techniques are available for these strategies when followed individually, combining all of them to develop a flexible transceiver array composed of small-sized coils for high-field MRI evokes several technical constraints as well as more fundamental issues.

Standard coil technology using lumped resistive (R), inductive (L), and capacitive (C) components imposes practical limits on the design and fabrication of miniaturized flexible RF devices. This is due to the rigidity of the coils themselves and to the minimum space required by discrete capacitors. Furthermore, even for coils fabricated on flexible substrate or made of semirigid copper, the use of lumped capacitors involves rigid solder joints that might crack on bending, may cause susceptibility artifacts (despite the use of nonmagnetic capacitors), and induces electrical stray fields increasing dielectric losses (17). This is especially important at ultrahigh field, where multiple lumped capacitors per coil are needed to generate a uniform current distribution along the loop (18). These constraints can be overcome by the concept of monolithic transmission line resonators (TLRs) (19,20). TLRs consist of two circular conducting bands intersected by diagonally opposite gaps and deposited on both sides of a low-loss dielectric substrate, which may be flexible. They are autoresonant and can be tuned...
over a wide range of NMR frequencies without the use of lumped elements by adjusting the geometrical parameters of the coil, such as substrate thickness and permittivity, or conductor width. After fabrication the TLR’s resonance frequency is fixed; however, appropriate fine-tuning under variable loading conditions can be achieved by resonant inductive matching. The $B_0$ field of the TLR is generated by the common mode current, given by the sum of the currents flowing in the two conductors. This current is intrinsically constant even if the length of the rings is comparable to the wavelength (21). With the TLR design, in comparison to standard RLC coils, dielectric losses are reduced, the RF homogeneity is improved (22), and when combined with an inductive coupling technique, no solder joints on the coil are needed.

A major technical challenge in coil array design is the mutual decoupling between individual coil elements. Conventional decoupling techniques use either geometrical overlap (10), with the drawback of reduced overall FOV and higher $g$-factors for parallel imaging due to the overlapping sensitivity profiles. Another decoupling strategy includes LC-networks between nonoverlapping coils (23,24), with the disadvantage of frequency and load-dependent decoupling efficiency. Some authors proposed strategies to decouple physically separated coils by magnetic flux sharing to combine the advantages of overlap and LC-network decoupling. Avdieievich and Hetherington (25) used a pair of overlapping annex loops with opposite winding orientation connected in series with two neighboring surface coil elements. Constantinides and Angeli (26) placed closed copper loops proximal to the array, partially overlapping with the mutually interacting surface coils, and thereby eliminating the magnetic coupling. Low-impedance preamplifiers are widely used for inter-element decoupling in receiver arrays (10). In transmit arrays, the mutual coupling can be reduced with the current source RF amplifier method (27,28), although it is currently not available for most MRI systems.

However, the above decoupling techniques are not well suited for double-sided monolithic structures. They are either restricted to standard single layer coils, using lumped elements, and therefore, contradict the monolithic feature of TLRs (e.g., LC-component decoupling), or they require three or more conductive layers, which implies a more complex fabrication process and also complicates handling after fabrication (i.e., existing inductive methods). Furthermore, none of them is readily implementable for flexible coil arrays. Hence, no flexible array of TLRs exists so far due to the lack of a suitable decoupling strategy.

The goal of this work is the development of an original transceiver array composed of small monolithic TLRs fabricated on flexible substrate for MRI at 7 T. To this end, a novel decoupling technique suitable for TLR arrays is proposed. This work aims at establishing the proof of concept that the new decoupling technique combined with monolithic design and microtechnological processes can be used to produce flexible two-dimensional arrays of TLRs with an arbitrary number of elements that enhance the RF detection sensitivity.

### METHODS

#### Design and Simulation

**Novel Decoupling Technique for TLR Arrays**

To decouple the elements within an array of TLRs, the basic TLR geometry (Fig. 1a,b) is extended by small circular annexes connected in series with the main windings (Fig. 1c,d). Neighboring elements are decoupled by overlapping a front-sided annex of one element with a back-sided annex of the other element (Fig. 1e,f). Four annexes per TLR permit decoupling from nearest neighbors in 2D-arrays (Fig. 1g,h) (29).

The coil arrays investigated in this study were composed of single-turn TLRs fabricated on flexible Teflon substrate providing low dielectric losses. Two different TLR geometries were compared, the first design being a 30-mm single-gap TLR self-resonating well above the Larmor frequency of interest, i.e. 297.2 MHz, and to be tuned and matched capacitively. For the second design, the geometric parameters of a 40-mm double-gap TLR were chosen in a way to closely approach the Larmor frequency, with an accuracy of a few MHz, to completely avoid

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**Table 1: Coil Geometries, Resonance Frequencies, and Quality Factors**

<table>
<thead>
<tr>
<th>Single TLR element</th>
<th>$d_{\text{ext}}$ (mm)</th>
<th>$w$ (μm)</th>
<th>$h$ (mm)</th>
<th>$\varepsilon_{\text{ext}}$</th>
<th>$w_{\text{ann}}$ (mm)</th>
<th>$l_{\text{ann}}$ (mm)</th>
<th>$f_0$ 3D EMS (MHz)</th>
<th>$f_0$ unloaded (MHz)</th>
<th>$f_0$ loaded (MHz)</th>
<th>$Q$ unloaded</th>
<th>$Q$ loaded</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single-gap</td>
<td>30</td>
<td>2.0</td>
<td>510</td>
<td>2.05</td>
<td>6.4</td>
<td>0.8</td>
<td>2.0</td>
<td>413.7</td>
<td>423.4</td>
<td>416.0</td>
<td>250</td>
</tr>
<tr>
<td>Double-gap</td>
<td>40</td>
<td>2.1</td>
<td>127</td>
<td>2.2</td>
<td>8.0</td>
<td>0.8</td>
<td>2.7</td>
<td>307.9</td>
<td>315.7</td>
<td>310.0</td>
<td>280</td>
</tr>
<tr>
<td>Double-gap</td>
<td>40</td>
<td>2.1</td>
<td>127</td>
<td>2.2</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>344.0</td>
<td>339.7</td>
<td>335.0</td>
<td>370</td>
</tr>
</tbody>
</table>

The table also includes the dimensions of the four-element arrays for single-gap and double-gap TLR design as well as the respective interelement spacing. TLRs were loaded by the torso phantom in respective experiments. $d_{\text{ext}}$ external TLR diameter, $w$ conductor width, $h$ substrate thickness, $\varepsilon$ relative permittivity of the substrate material, $a_{\text{ext}}$ external diameter of the decoupling annexes, $w_{\text{ann}}$ conductor width for the annexes, $l_{\text{ann}}$ length of the linear segment connecting the annexes to the main windings.
lumped element tuning. This coil could then be fine-tuned and matched inductively with a coupling loop.

Exact geometric parameters of single TLR elements and arrays are given in the results section (Table 1).

Matching Networks

Capacitive matching networks consisted of a variable tuning capacitor (6.5–30 pF) and two series matching capacitors (6.5–30 pF) connected in symmetric configuration between the tuning capacitor and the coaxial feed cable. The question, how a TLR can be optimally matched capacitively to the receiver has not been answered yet. Here, a configuration was chosen, where the tuning capacitor is connected in parallel to the coil capacitance, as in a conventional RLC coil circuit. In the case of a TLR, the coil capacitance is distributed across the substrate, and hence, the feed points were positioned on either side of the TLR. The feed point position along the winding was chosen to achieve the highest voltage across the two conductors, which is at a gap on one side and in the center of the conductor on the other side (Fig. 2a).

For inductive matching, pickup loops were placed at a distance of 6.5 mm above each TLR. The pickup loops were tuned and matched with lumped element capacitors (Fig. 2b; \( C_T, C_M = 3–10 \) pF). Each pair of TLR and pickup loop was operated in over-coupled mode to permit fine-tuning (30), since the free resonance frequency of the fabricated TLRs was few MHz above the Larmor frequency. To shift the lower resonance peak of the coupled two-coil system to the Larmor frequency, the pickup loop had to be tuned above the free resonance frequency of the TLR (Supporting Information Fig. S1).

To investigate the effect of mutual coupling between pickup loops on the overall decoupling efficiency, the coupling between two neighboring inductively matched double-gap TLRs, was simulated for pickup loop diameters of 10–30 mm in 5 mm steps.

Decoupling Performance

The size of the decoupling annexes was optimized for single- and double-gap TLR designs by simulating the transmission scattering parameter \( S_{21} \) of two neighboring elements as a function of the annex diameter. For each annex size, the TLRs were tuned and matched by circuit cosimulation before recording the \( S_{21} \) values. The annex size resulting in the best isolation between neighboring channels was considered optimal. The width of the conducting bands forming the decoupling annexes was reduced in comparison to the width of the main winding and set to 0.8 mm as the space for placing the annexes in four-element arrays is limited (Fig. 1g,h).

After predicting the resonance frequencies for the basic TLR designs with an analytical model (Eq. (1)), the shift in resonance frequency induced by the decoupling annexes was accounted for by 3D electromagnetic simulations (EMS) for finding the final TLR geometries (Results section, Table 1).

The single element TLR designs, annex sizes and pickup loop diameters as determined in the previous steps, were used to demonstrate the decoupling technique in four-element arrays with the individual elements arranged to form a square with an interelement distance of 2 mm. The decoupling efficiency in arrays of single- and double-gap TLRs, each with either capacitive or inductive matching, was compared by simulating the full S-parameter matrices. To explain variations in decoupling efficiency between the different configurations, current density distributions in single TLR elements were simulated at the resonance frequency (31).

Specific Absorption Rate

To evaluate how adding the decoupling annexes influences the performance of the TLRs in terms of specific absorption rate (SAR), local unaveraged SAR distributions were derived from EMS for single TLR elements with and without decoupling annexes, and for a four-element array decoupled by overlapping annexes (All channels were driven with the same phase and amplitude). Postprocessing was performed using a dedicated toolbox (SimOpTx, Research Studio Austria, MedUni Vienna, Austria) using local power correlation matrices (15,32) computed by an ultrafast convolution based SAR averaging algorithm (33).
Simulation Tools

Starting values for the geometrical parameters of the TLR coils were determined using an analytical model for the resonance condition of TLRs (Eq. (1)) (21).

\[
\frac{L_{tot} \omega_0}{4\pi n_0 Z_0} \tan \left( \frac{\omega_0 \sqrt{\varepsilon \mu}}{4n_0 c} \right) = 1
\]  

with the angular resonance frequency \(\omega_0\), the coil’s equivalent inductance \(L_{tot}\), the length of one conducting band \(l\), and the parallel-plate transmission line characteristic impedance \(Z_0\), which is a function of the conductor width \(w\), the substrate thickness \(h\), and its relative permittivity \(\varepsilon\). \(c\) denotes the vacuum speed of light and \(n_0\) the number of gaps per conductor. \(L_{tot}\), which is the sum of the individual inductances of the windings on both sides of the substrate and their respective mutual inductance, as well as \(Z_0\) can be calculated with semiempirical models (20).

TLR geometries and decoupling efficiency were studied by full wave 3D EMS (XFtfd 7.3, Remcom, State College, PA) in combination with circuit cosimulation (ADS, Agilent, Santa Clara, CA). For 3D EMS a basic mesh resolution of \(2 \times 2 \times 2 \, \text{mm}^3\) was used. In the vicinity of the coil, the resolution in the coil plane was increased to 0.5 mm for \(S\)-parameter simulations and to 0.25 mm for current density simulations; the resolution along the coil axis inside the substrate was set to half the substrate thickness. The XACT-mesh technology allowing for conformal modeling (34) embedded in the simulation software was enabled for improved meshing accuracy. A rectangular block phantom with the electric and magnetic properties of muscle tissue (0.72 S/m conductivity, 64 relative permittivity) placed 5 mm below the coil was used as load in all EMS. The phantom was 5 cm thick and its lateral dimensions were chosen in a way that the phantom exceeds the simulated single TLR element, two-element or four-element array by 5 cm. Reduced sample dimensions were chosen in comparison to bench and MR measurements (see next section) to save simulations time; it was verified that results were not substantially altered by this simplification. For rapid tuning and matching, the corresponding capacitors were modeled as 50 \(\Omega\) ports in 3D EMS and the resulting \(S\)-parameters were postprocessed using circuit cosimulation (35). Current density simulation data were analyzed using Matlab (Mathworks, Natick, MA).

Hardware and Phantoms

For step-to-step characterization of the novel TLR designs, single coil elements with and without annexes, as well as two- and four-element arrays of single- and double-gap design were fabricated. Single-gap TLRs were etched from standard 1.5 mm thick FR4 printed circuit board material. Nonmagnetic trimmer capacitors (Murata Manufacturing Company, Kyoto, Japan) were used and shortened “bazooka” type baluns (Fig. 2c) were placed on the coaxial cables at a distance smaller than one eighth of the wavelength from the coils to reduce shield currents.

Bench measurements were performed using a four port vector network analyzer (E5071C, Agilent, Santa Clara, CA).

For MR imaging, one of the four tested four-element arrays was selected based on bench measurements and simulation results. MRI experiments were carried out on a 7T whole-body MRI system (Magnetom 7 T MRI, Siemens Medical Solutions, Erlangen, Germany) equipped with a SC72d gradient coil with maximum gradient strength of 70 mT/m and slew rate of 200 T/m/s. All coil elements were used in transmit/receive mode, driven with the same amplitude and phase during transmission. Additional hardware, including power splitters, transmit/receive switches, and low-noise preamplifiers were placed on a separate interface board.

For bench and MRI experiments with the TLRs in planar configuration, a torso phantom with dimensions and electromagnetic properties as specified in the ASTM F2182-11a standard was used. The phantom is box-shaped \((65 \times 42 \times 9 \, \text{cm}^3)\) and filled with 25 L polyacryl acid gel. To test the ability of form-fitting and the applicability for various target regions, the before selected four-element array was wrapped onto a cylindrical phantom \((7.5 \, \text{cm diameter}, 17.5 \, \text{cm long})\) filled with the same gel, representing, for example, a human arm or lower leg. Further, to investigate the performance of the developed array when loaded less than by the phantoms, a kiwano fruit \((Cucumis metuliferus)\) was used in bench and MRI experiments mimicking, for instance, wrist or small animal loading conditions. In addition, the tuning and matching capability as well as the decoupling performance of the selected array were evaluated on the bench when it was placed on the torso of a volunteer (male, 39 years, body mass index = 23 kg/m²). For all configurations, 5 mm thick acrylic glass was located between sample and TLR array.

Bench Measurements

Decoupling Performance

On the workbench, the decoupling efficiency was evaluated by measuring the transmission scattering parameters of two- and four-element arrays of single- and double-gap TLRs. \(S\)-parameter matrices were recorded at the Larmor frequency after a typical impedance match better than –30 dB had been achieved for all elements. From these measurements and from simulation results, the array configuration showing the best decoupling efficiency was chosen for further experimental evaluation.

To select a suitable pickup loop size for the inductive matching setup, the coupling of neighboring inductively matched double-gap TLRs in two-element arrays was measured for pickup loop diameters from 10 to 30 mm in 5 mm steps. Bench measurements were compared to the results from EMS.
Form-Fitting

The selected array was wrapped on an acrylic glass former suitable for experiments with the cylindrical phantom and the kiwano fruit. Tuning and matching capacitors were adjusted and full S-parameter matrices were measured to evaluate the matching and decoupling performance in form-fitted configuration.

**Pickup Loop Noise Factor.** The noise degradation associated with pickup loop matching (36) was studied as a function of pickup loop size by calculating the noise factor $F$ for single double-gap TLR elements loaded by the torso phantom:

$$F = 1 + \frac{k_c^2}{K^2} + \frac{Q_{TLR}}{K^2 Q_P} + \left(1 - \frac{f_0^2}{f_L^2}\right)^2$$

with the coupling coefficient $k$

$$k = \frac{1}{2} \left(\frac{f_0}{f_L} + \frac{f_0}{f_p}\right) \sqrt{\left(\frac{f_L^2 - f_s^2}{f_L^2 + f_s^2}\right)^2 - \left(\frac{f_0^2 - f_s^2}{f_0^2 + f_s^2}\right)^2}$$

and the critical coupling coefficient $k_c$

$$k_c = \frac{1}{\sqrt{Q_{TLR}/Q_P}}$$

These equations include the resonance frequency of the isolated TLR $f_0$, the lower ($f_s$), and the higher resonance frequency ($f_L$) of the over-coupled system of TLR and pickup loop (see Supporting Information Fig. S1), and the Larmor frequency $f_L$. $Q_{TLR}$ and $Q_P$ denote the isolated quality factors of the TLR when loaded with the torso phantom and the pickup loop, respectively.

**Influence of the Decoupling Annexes.** The influence of the decoupling annexes on the coils’ resonance frequencies and quality factors was evaluated by comparing isolated double-gap elements with and without annexes. Measurements were done in unloaded configuration and when the coils were loaded by the torso phantom using the single-loop probe method (37), while not connected to a matching network. The influence of the single-loop probe was considered negligible when the reflection coefficient measured at its terminal was $<-40$ dB.

**MRI Experiments**

**Preparatory Measurements**

To examine the $B_1$ field distortion potentially induced by the pickup loops (38), MR measurements with a single, inductively matched, double-gap TLR were performed. Transversal 2D gradient echo (GRE) images ($T_R/T_E = 140$ ms/7.74 ms, $0.375 \times 0.375$ mm$^2$ resolution, 1 mm slice thickness, 256 x 128 matrix) with a flip angle $>180^\circ$ close to the TLR were acquired using the five different pickup loops placed 6.5 mm above the TLR, and compared to those acquired using a 30-mm pickup loop placed at a distance of 20 mm, for which the induced distortion is assumed to be negligible (38).

The pickup loop diameter, for which the best compromise between preserved decoupling efficiency, low noise degradation and low $B_1$ field distortion is achieved, was selected for further experiments.

**Noise Correlation and Parallel Imaging Performance**

The parallel imaging performance of the selected array in planar and bent configuration was evaluated in terms of the GRAPPA $g$-factor applying the pseudomultiple replica method (39) and off-line GRAPPA reconstruction as described by Breuer et al. (40). Therefore, noise-only data, for computing the noise correlation matrix, and fully encoded 2D GRE images ($T_R/T_E = 500$ ms/7.74 ms, $80^\circ$ nominal flip angle, $0.52 \times 0.52$ mm$^2$ in-plane resolution, 1 mm slice thickness) of the cylindrical (transversal slices) and the torso phantom (transversal and coronal slices) were acquired. Acceleration factors of $R = 1$ (no acceleration), $R = 2$, and $R = 3$, were mimicked during reconstruction by eliminating not required phase encoding steps. Resulting $g$-factors were computed for sum-of-squares combined images. To compare the parallel imaging performance in flat and bent array configuration, mean and maximum $g$-factors were calculated for an elliptical region of interest (ROI) (major axis 60 mm, minor axis 40 mm) drawn on transversal images of both, the torso and the cylindrical phantom.

**High-Resolution MRI**

High-resolution images of the kiwano fruit were acquired with the selected array in form-fitted configuration applying a 3D GRE sequence ($T_R/T_E = 150$ ms/6.56 ms, $76 \times 84$ mm$^2$ FOV, $220 \times 220$ mm$^2$ in-plane resolution, 52 slices, 1 mm slice thickness, GRAPPA with $R = 2 \times 2$, $T_{acq} = 7$ min 15 s).

**RESULTS**

**Coil Geometries and Matching Setup**

The geometrical parameters of the fabricated single TLR elements and four-element arrays including the
optimized annex sizes are summarized in Table 1. The table also provides simulated and measured resonance frequencies, measured \( Q \) factors for the TLRs in unloaded condition and when they are loaded by the torso phantom, and the comparison of the double-gap TLRs’ RF characteristics with and without decoupling annexes.

The simulated transmission scattering parameters of two neighboring TLRs (corresponding to elements 3 and 4 in four-element arrays, Fig. 1) as a function of the annex diameter are shown in Figure 3 for single-gap (capacitively matched) and double-gap (inductively matched) design. The simulated decoupling levels with size-optimized annexes were \(-17.5 \text{ dB} \) for single-gap TLR and \(-16.5 \text{ dB} \) for double-gap TLR, respectively.

For the final array configuration, 15-mm pickup loops were selected after comparing matching performance, decoupling efficiency, noise factor, and \( B_1 \) field distortion for pickup loops with diameters ranging from 10 to 30 mm in 5 mm steps. Typical matching levels better than \(-30 \text{ dB} \) at the Larmor frequency could be achieved using any of the five tested pickup loops. However, a decrease in decoupling efficiency with increasing pickup loop size was observed in simulations (transmission increased from \(-16.2 \) to \(-12.5 \text{ dB} \)) and bench measurements (from \(-14.7 \) to \(-12 \text{ dB} \)), as shown in Figure 4. Conversely, the noise performance improved with increasing pickup loop diameter. The calculated noise factors decreased from 3.1 (10 mm pickup loop diameter) to 1.1 (30 mm pickup loop diameter) corresponding to 4.9 and 0.4 dB noise degradation, respectively (Fig. 4).

High-flip-angle images revealed an asymmetry in \( B_1 \) distribution in comparison to the reference image obtained with the 30-mm pickup loop placed 20 mm above the TLR, which decreased with increasing pickup loop diameter.

The 15- and 20-mm pickup loops performed sufficiently well to be used in the developed four-element array; the 15-mm pickup loops were then selected as they provided a higher decoupling efficiency, which is the primary objective of this study. The corresponding pickup loop noise factor was 1.6, and the measured isolation between neighboring TLRs was \(-14.3 \text{ dB} \).

**Choice of the Array Design**

Simulations of the current density (Fig. 5) show that the current is not equally distributed among all decoupling annexes with the single-gap TLR design whatever the matching configuration. As the current exhibits a minimum at the gap, two annexes placed at different distances from the gap do not carry the same current. This can be overcome with the double-gap design for which all annexes are placed at the same distance from the gaps. The simulated current density distributions also show that connecting capacitive matching networks to the TLRs induces an asymmetry in current density between front and back conductor, while inductive matching provides the same current distribution on both faces of the TLR.

The simulated S-parameter matrix (Fig. 5) of the capacitively matched four-element array of single-gap TLRs shows unequal decoupling efficiencies for pairs of
TLR elements ranging from $-7$ to $-18$ dB. A comparable asymmetry is observed when the single-gap array is inductively matched. Also for the double-gap TLRs a variation in decoupling efficiency is observed using capacitive coupling while the simulated transmission parameters are equilibrated with the inductive matching technique. Resulting transmission parameters for the latter were $-16$ dB for nearest neighbors, and $-14$ dB for diagonal elements, which are not decoupled by overlapping annexes.

The S-parameter matrices of four-element arrays recorded in bench measurements (Fig. 5) basically reflect the behavior observed in EMS with slightly lower coupling values. For single-gap arrays and the capacitively matched double-gap array, the resonance peaks of the individual elements showed severe asymmetry in bench measurements due to insufficient interelement decoupling. No peak splitting or asymmetry was observed for the inductively matched array of double-gap TLRs for which an isolation of $-15$ dB or better was measured when loaded by the torso phantom. The observed decoupling performance proved robust when the same array was loaded by the torso of the volunteer, and also tuning and matching at the Larmor frequency could easily be achieved. In unloaded condition, the coupling between diagonal elements, which are not decoupled with the proposed technique, increased to $-6$ dB while the isolation between direct neighbors remained below $-15$ dB.

FIG. 5. Choice of the array configuration. Current density distributions in single TLR elements are shown on the left. The averaged current density in each annex normalized to the mean $J$ over all four annexes is plotted in the center. On the right, simulated and measured decoupling levels are depicted.

FIG. 6. Unaveraged SAR distributions derived from 3D EMS. Coronal maximum intensity projections of unaveraged SAR distributions are shown for double-gap TLRs, a: without decoupling annexes (Fig. 1b), b: with decoupling annexes (Fig. 1d), and c: for the selected four-element array (Fig. 1h). All elements of the four-element array were driven with equal amplitudes and phases, resulting in destructive interference of $E$-fields, and thus, negligible SAR between elements. SAR values are normalized to 1 W input power.
Following the above results, the inductively matched array of double-gap TLRs was then chosen for MRI experiments.

Specific Absorption Rate

Figure 6 depicts maximum intensity projections of the simulated unaveraged SAR distributions for single double-gap TLRs with and without decoupling annexes, and for the selected four-element array. Adding the decoupling annexes leads to a 14.5% lower peak SAR value. Further, it is demonstrated that no SAR hot spots are introduced at the location of the annexes, neither for the single TLR element (Fig. 6b), nor for the four-element array (Fig. 6c). For the four-element array, all TLRs were driven in-phase resulting in destructive interference of E-fields in the center.

Performance of the Flexible TLR Array

Coronal GRE images and corresponding g-factor maps acquired with the selected four-element array in flat configuration are shown in Figure 7. The depicted slices are located 2 and 20 mm below the phantom surface. For the top slice, a signal void between neighboring elements can be observed along the z-direction, where the produced $B_1$ field is parallel to $B_0$. This effect is greatly reduced for the slice located deeper inside the phantom. Signal related to the decoupling annexes cannot be clearly distinguished.

The measured transmission scattering parameters and corresponding noise correlation matrices of the inductively matched (15-mm pickup loop), four-element array of double-gap TLRs in bent and flat configuration are shown in Figure 8 for direct comparison. In bent array configuration, slightly increased transmission S-parameters between neighboring elements are observed. Noise correlation values are comparable to those for the flat configuration with peak values of 0.31 (bent) and 0.34 (flat), respectively.

Figure 8 also shows transversal phantom MR images acquired in flat and bent configuration and the calculated g-factor maps for acceleration factors $R=2$ and $R=3$. No degradation in parallel imaging performance due to bending of the array was observed. Mean g-factors calculated for the elliptical ROI were $1.2 \pm 0.2$ ($R=2$) and $1.9 \pm 0.4$ ($R=3$) for the flat configuration and $1.1 \pm 0.1$ ($R=2$) and $1.6 \pm 0.4$ ($R=3$) for the bent configuration, respectively.

In Figure 9, a high-resolution image of a kiwano fruit acquired in bent array configuration is shown. The decoupling performance of the form-fitted array when loaded by the kiwano fruit was comparable to that observed with the cylindrical phantom.

DISCUSSION AND CONCLUSIONS

In this article, a novel technique for interelement decoupling in TLR arrays based on mutual magnetic flux sharing via overlapping annexes is introduced. This new decoupling technique is frequency independent over a wide range and robust against variations in loading, similar to conventional overlap decoupling. Although this new decoupling technique was first demonstrated for 7T MRI, it is fully applicable to other static field strengths and can be implemented for various coil-element sizes.

The decoupling efficiency of the proposed technique was optimized and evaluated for two- and four-element arrays with single- or double-gap designs, combined with capacitive or inductive matching.

It should be noticed that the difference in diameter between single-gap TLRs (30 mm) and double-gap TLRs (40 mm) used in this study does not prevent the comparison of the two designs in terms of mutual decoupling. The decoupling performance was optimized separately for both TLR types and has been shown to depend on the current density in the annexes and on the size of the annexes but not on the TLR diameter. A comparison in terms of imaging performances (SNR, $B_1$ homogeneity, FOV) of the two designs would require the use of single-gap and double-gap TLRs with equal diameters. Since the aim of this work was to evaluate the performance of the presented novel decoupling method, however, the difference in diameters is not relevant.
Using 3D EMS, it is demonstrated that the decoupling efficiency in TLR arrays is closely related to the current density distribution along the transmission line. In particular, the current density has to be made equal for all decoupling annexes to equilibrate decoupling levels between all nearest neighbors in 2D-arrays. It is shown, that this can be achieved using a double-gap TLR design together with resonant inductive matching. Using the double-gap design, the respective distances between each annex and the closest gap (at which the current density is forced to zero) are equal. As long as the symmetry of the half-wave sinusoidal current density along the conductor of the transmission line (21) is not broken, it follows that the current density in each annex is the same. An inductive matching scheme conserves this intrinsic symmetry, since it does not introduce a defined electrical ground at any position of the TLR. The capacitive matching network used in this work did not fulfill this criterion, and therefore, resulted in asymmetric current distribution. However, other approaches for positioning of the feed points on the TLR and possibly using asymmetric matching capacitors, could be studied in future work in view of equilibrating the current distribution in the TLR. But even if a solution can be found, resonant inductive matching still offers the intrinsic advantage for flexible arrays, that no solder joints have to be added onto the coils. It was demonstrated that tuning and matching may be achieved with this technique for various loading conditions (e.g., human torso, torso phantom, cylindrical phantom, and kiwano fruit).
The double-gap TLR geometry enables not only an equally distributed current density among the decoupling annexes but also a symmetric array layout regarding the relative gap position for the individual elements. The relative gap orientation may strongly influence the mutual coupling behavior, as Fang et al. (43) demonstrated for spiral surface coils. The proposed design avoids these effects, since each coil element has the same geometric relation to its four nearest neighbors (Fig. 1g). Further, the investigated four-element arrays cover all nearest-neighbor interactions in tetragonally arranged arrays. Therefore, the proposed decoupling principle can be easily expanded to multielement linear or 2D arrays without restriction regarding the number of elements.

It was found that the influence of adding the decoupling annexes to the basic TLR geometry is not a limiting factor in terms of imaging and SAR performance. The resulting decrease of the resonance frequency can be accounted for by proper choice of the TLR geometry. The unloaded quality factor is decreased by adding the annexes, but is still well higher than the loaded Q indicating that sample noise is the dominating loss mechanism. When using this decoupling technique for small TLRs at lower field strength, e.g., 1.5 T, it should be considered that adding the decoupling annexes increases the coil noise, and therefore, also increases the coil diameter for which coil noise becomes dominant. As shown in Table 1, the optimal size of the annexes is not fixed in general. It has to be specifically optimized for a given array configuration since the magnetic flux to be cancelled mainly depends on the size of the TLRs, on the distance between TLRs and on the arrangement of the array elements. The magnetic flux shared by the overlapped annexes depends on the thickness of the substrate and the annex size.

A theoretical limitation of the presented array design is that coupling between diagonal elements, which are not decoupled by overlapping annexes, may induce splitting of the resonance peak. In practice, however, a single peak is observed for each TLR element within the array when sufficiently loaded; this holds true for all investigated loading conditions. For applications where loading is minimal, further investigations might be needed to ensure proper tuning and decoupling.

Pickup loop matching in over-coupling mode can be implemented for transceiver coil arrays, but requires careful choice of the position and size of the pickup loops. These can be determined by finding a reasonable trade-off between pickup loop noise factor, coupling between neighboring pickup loops, and $B_1$ distortion. The closer the free resonance frequency of a fabricated TLR matches the Larmor frequency, the less it has to be retuned and the smaller the noise contribution of the pickup loop is (Eq. (2)).

A slight asymmetry in signal intensity between the left and right side of the array was observed in 7 T MR images (Figs. 7 and 8). We believe that the major source for this artifact is high-frequency effects introducing asymmetry in $B_1^+$ and $B_1^-$ (44). Note that in the present work, all transmit elements were driven with equal amplitude and phase. Such asymmetry could in future implementations be alleviated by the use of optimized amplitude and phase settings between the coil elements, using, e.g., a parallel transmission system. This would be particularly beneficial for bent configurations, since the relative phases could be easily adapted to the target geometry without hardware changes.

The form-fitting ability of the developed array was successfully demonstrated in bench and MR experiments when wrapped on a cylindrical former where the measured transmission scattering parameters, noise correlation matrices, and g-factors proved robust concerning this mechanical deformation of the TLR array. The proposed decoupling technique is especially favorable for form-fitting TLR arrays comprising a large number of elements, because readily decoupled arrays can be fabricated on a single flexible substrate in one standard photo-lithographic etching process. Also, in contrast to previous work, proposing the principle of magnetic flux sharing for physically separated coils (25,26), no soldering is necessary.

**FIG. 9.** High-resolution images of a kiwano fruit (*Cucumis metuliferus*). A transversal (a) and a coronal slice (b) are shown. GRE images (220 × 220 × 1000 μm³ resolution) were acquired at 7 T with the inductively matched array of double-gap TLRs in form-fitted configuration in an acquisition time of 7 min 15 s. The pulp and the seeds of the fruit as well as the inner structure of the paring can be observed. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]
Such flexible arrays are well-suited for studying anatomical regions, which may vary strongly in size and shape from patient to patient and require both a large FOV and high SNR. Potential biomedical applications include high-resolution imaging of skin and joints like wrist, elbow or knee, or dynamic imaging of moving organs such as the heart. In this respect, the performance of the developed prototype array will be further improved by increasing the number of coil elements and by adapting the size of the individual elements to the targeted organ or structure.

Considering, for instance, skin imaging (45,46), the achievable SNR could be further increased by miniaturizing the TLR elements until the threshold between sample and coil noise dominance is reached. In the coil noise domain, further SNR improvement could be achieved by reducing internal coil noise, e.g., by using superconducting coil technology (8). The concepts presented here are particularly attractive for this field of applications since both the TLR coil design and the proposed decoupling technique are fully monolithic and none of them imposes limits in terms of coil miniaturization. Furthermore, the inductive matching approach used here avoids direct soldering on the coil, and thus, allows preserving low noise features of superconducting coils.

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