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Over-overlapped Loop Arrays: A Numerical Study

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Abstract

Arrays of coils are commonly used in MRI both for reception and in parallel transmission to alleviate radiofrequency field inhomogeneities at high fields. Most designs typically overlap loop elements by a critical area (approximately 10 %) to minimize mutual inductive couplings. With this geometrical constraint, loop sizes have to be reduced to accommodate large numbers of coils for a given coverage. However, the contribution of coil noise to total noise increases as each coil size decreases, which reduces overall signal-to-noise ratio (SNR), especially in deeper regions of the sample volume. Here we propose arrays designs using elements that overlap much more (over-overlapped), and using numerical calculations we investigate their performance compared to two kinds of conventionally overlapped arrays (one with the same coil size but smaller coil number, and one with the same coil number but smaller coil size). Our simulation results show that the over-overlapped array can considerably increase the central SNR when coil noise dominates.

Keywords

RF coil; loop; array; over-overlap; SNR; parallel imaging

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Author Contributions

XY conceived the presented idea and designed the analysis. ML performed the simulation, analyzed the data and prepared figures for the manuscript. XY and JCG supervised this study and wrote the manuscript.

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Introduction

Arrays of coils have been used as receivers in MRI since they were introduced in the late 1980s' and early 90s' [1], as well as in parallel transmission (pTx) systems to alleviate transmit field (B_1^+) inhomogeneities at high fields since the 2000s' [2, 3]. Simple loop coils are by far the most widely used array elements in receiver arrays used in clinical scanners (3T and lower) as well as brain imaging applications in ultrahigh field scanners (7 T and above). Loops are typically overlapped by a critical area (approximately 10%) [1], or separated by gaps with decoupling provided by components such as shields, capacitive/inductive networks and passive resonators, which minimize the mutual electromagnetic (EM) coupling [4–7]. The small but critical degree of overlap leads to geometrical constraints on loop coils in arrays. As the number of elements in the array increases, the loop size has to be reduced. And to cover a given volume of interest (VOI) using a large number of loops, the size of each loop has to be kept small.

For MR signal reception, increasing the number of coil elements increases the SNR. However, increasing the number of loops while decreasing their size increases the SNR at peripheral areas but may lower the SNR in deeper areas compared to arrays with a smaller number of loops but larger sizes [8, 9]. This happens because the coil itself has resistance and generates a thermal noise which becomes dominant as loop size decreases. It is therefore desirable to increase the number of coils while still using larger elements, which in turn requires larger degrees of overlap. In addition to MR signal reception, coil element size and number play important roles in transmit arrays, affecting B_1 and the special absorption ratio (SAR) because their EM fields provide the boundary conditions for RF shimming and tailored pulse designs.

Loop arrays in use today are based on the standard overlapping or gap designs, and the use of over-overlapped arrays has not previously been reported. A limitation of the over-overlapped coil is that the mutual coupling is more complex and is not easily reduced by current decoupling methods. Recently, several novel decoupling methods have been proposed for loop arrays in which the loops can be over-overlapped with no significant loss of SNR. For example, the AIR coil [10] and high-impedance coil [11] proposed novel preamplifier decoupling methods that allow coil overlap area to vary from a gap to up to 40%. Yan et al [12] proposed the impedance distribution can be intentionally re-distributed to provide another degree of freedom to optimize the overlapped area. With these novel and flexible decoupling approaches, we believe it is possible to design and build practical over-overlapped loop arrays. Here we report the results of full-wave EM simulations to investigate the behavior of over-overlapped arrays as receivers, including the SNR and g-factor in parallel imaging, and compares the performance with conventionally-overlapped arrays.

Methods

1. Over-overlapped coil and EM simulation model

Figures 1a and 1b show the layout of two circular loops with a conventional overlapped area of ~10% and loops with an over-overlapped area of ~40%. Note that the mentioned

overlapped ratios are for the area rather than the distance. Loop arrays consisting of 12 overlapped (adjacent loops overlapped 40%) circular loops were modeled on a cylindrical surface, as shown in Figure 1d. The coil diameter varies across 5 cm to 9 cm at different static magnetic (B_0) fields (1.5 T, 3 T and 7 T). A cylindrical phantom was positioned inside each loop array as loading. The radius of the phantom varies across 4 cm to 8 cm accordingly to ensure the phantom is always 1 cm apart from the coils. The EM properties of the phantom were set to match the average value of human brain at each static magnetic field, with relative permittivity $\epsilon_r = 82.8 / 63.1 / 52$, and conductivity σ (S/m) = $0.41 / 0.46 / 0.55$ for 1.5 T / 3 T / 7 T respectively. The coils' conductors have a width of 0.5 cm for 9-cm-diameter coils and a width of 0.3 cm for other sized coils. As a comparison, we also modeled arrays of conventionally-overlapped loops with the same coil size but smaller coil number (8 coils, Figure 1c), and arrays of conventionally-overlapped loops with the same coil number but smaller coil sizes (Figure 1e). For the latter case, each loop is shrunk to ~70% along the overlapping direction to ensure adjacent loops are overlapped ~10%.

Simulations were performed using commercial finite-element (FEM) solver EM software (Ansys HFSS, Canonsburg, PA, USA) on a DELL workstation with 192 GB RAM and two 12-core Intel Xeon 3.0-GHz CPUs [13, 14]. Unit current excitation sources were used to drive each loop individually [15]. The loop conductors were modeled as copper sheets with finite conductivity of 5.8×10^7 S/m. Note that the current-source-driven mode assumes perfect reactive decoupling among all loop elements. In each loop, current excitation sources are distributed less than 1/10 wavelength apart to realize uniform current distribution along each coils' conductors. The number of current excitation sources in a single coil varies across 1 (5-cm-diameter coil at 1.5 T) to 6 (9-cm-diameter coil at 7 T). Manual initial mesh definition was used to ensure the reliability and accelerate simulation convergence. The H- and E- fields were exported from Ansys HFSS to Matlab to calculate the receive sensitivity and sample noise matrix.

2. SNR and g-factor calculation

The reception field of each coil element (B_1^-) were calculated using Eq. 1, where B_x and B_y are transverse RF magnetic fields produced by unit current excitation sources [16].

$$B_1^- = (B_x - iB_y)^*/2 \quad [1]$$

The sample noise matrices (R_s) were calculated by integrating the power dissipation over the entire sample volume, as shown in Eq. 2, where E_{km} is the local electric field of voxel k from coil element m , E_{kn} is the local conductivity of voxel k from coil element n , x , y and z are the voxel size in x , y , and z directions, receptivity [17, 18].

$$R_s = \Delta x \Delta y \Delta z \times \sum_k \sigma_k (E_{km} E_{kn}^*) \quad [2]$$

The voxel size was set to $2 \times 2 \times 2$ mm³ in this study. The coil noise matrices (R_c), including the conductor noise and components (capacitors) noise, were calculated by integrating

power dissipation over all conductors (P_{cond}) and all components (P_{compon}), as shown in Eq. 3.

$$\mathbf{R}_c = \oint\!\!\!\oint (P_{cond} + P_{compon}) \quad [3]$$

With the assumption of perfect reactive decoupling, no RF power is dissipated on other coils when one coil is excited, so $\mathbf{R}_c$ is a diagonal matrix. P_{cond} and P_{compon} were calculated based on separate single coil simulations with tuning, matching and distributed capacitors/ inductors that make the coil resonate at the corresponding Larmor frequency and match to the characteristic impedance.

We evaluated two kinds of SNR in this work; one is the intrinsic SNR (iSNR) without the coil noise (Eq. 4), and the other is the “real” SNR (rSNR) accounting for the coil noise (Eq. 5).

$$iSNR \propto \sqrt{\mathbf{B}_1^{-H} \times \mathbf{R}_s \times \mathbf{B}_1^-} \quad [4]$$

$$rSNR \propto \sqrt{\mathbf{B}_1^{-H} \times (\mathbf{R}_s + \mathbf{R}_c) \times \mathbf{B}_1^-} \quad [5]$$

Both kinds of SNR are normalized to the maximum value for a given coil-diameter and a given $\mathbf{B}_0$ field. The SNR in the central and peripheral areas exhibit different behaviors as the coil number and size vary, so the central and peripheral SNRs were evaluated and compared separately. The central SNR was calculated by averaging the SNR over a central round area that has 0.2 diameter of the cylindrical phantom’s diameter, while the peripheral SNR is calculated by averaging SNR over a peripheral ring area that has 0.34 cross-sectional area of the cylindrical phantom, as shown in Figure 2.

Another important factor in evaluating a receive array is the ability to complement the reduced gradient encoding of parallel imaging. The geometry (g-) factor affects the SNR in the reconstructed images and depends on the number and geometrical configurations of the array elements. We used a standard definition of the SENSE g-factor [19]. Similar to the SNR, both peripheral and central g-factors were calculated for various acceleration factors ($R = 2, 3$ and 4). The reliability and accuracy of the above simulation approach for SNR calculation have been validated in previous works [20–22].

Results

1. Intrinsic SNR without considering coil noise (iSNR)

Figure 3 shows the normalized iSNR maps in the central axial slice for coil sizes 5 cm, 7 cm and 9 cm, and $\mathbf{B}_0$ fields 1.5 T, 3 T and 7 T. Figure 4 summarizes the peripheral and central iSNR ratios of the 12-ch over-overlapped, the 12-ch conventionally-overlapped, and the 8-ch conventionally-overlapped arrays. For all cases, the peripheral iSNR increases significantly (41% to 67%) as the coil number increased, while the central iSNR stays at the same level.

The over-overlapped array exhibits higher peripheral iSNR compared to the 8-ch array (41% to 58% higher), but lower peripheral iSNR when compared to the 12-ch conventionally-overlapped array (5.1% to 21% lower). It should be noted that the iSNR does not consider coil noise and is determined by the current patterns [23, 24].

2. SNR accounting for the real effects of coil noise (rSNR)

With real coils, both the coil and sample contribute to the total noise in an MR image. The coil noise typically comes from the finite-conductive copper, lossy components, solder joints and imperfect decoupling. For simplicity, the solder joint was not considered and coils were assumed to be perfectly reactively decoupled. Figure 5 shows the sample noise ratio for different coil sizes and different B_0 fields. The sample noise ratio is defined as the ratio of sample noise to total noise [25]. As expected, the coils used for the 12-ch conventionally-overlapped array (elliptical) have a lower sample noise ratio compared to the coils in the other arrays due to their smaller size. The sample resistance is approximately proportional to the square of the Larmor frequency while the coil noise increases approximately as the square root of the Larmor frequency [26], so the sample noise increases much faster than the coil noise as the frequency increases. As loop size and B_0 fields increase, the sample noise becomes more dominant and the sample noise ratio factors of elliptical and circular loops converge. For 7- and 9-cm-diameter coils at 7 T, the sample noise ratios of the circular coil are respectively only 5% and 2% higher than that of the elliptical coil. At lower B_0 fields such as 1.5 T, however, the sample noise factor of the circular coil is considerably higher than that of the elliptical coil, yielding an average increase of 35% across different coil sizes.

As shown in Eqs. 4 and 5, SNR is inversely proportional to the sample noise ratio when calculating rSNR. Therefore, the rSNR of a 12-ch conventionally-overlapped array may decrease much more than that of the over-overlapped array, especially when the coil noise dominates (such as the 5- and 7-cm-diameter coils at 1.5 T and the 5-cm-diameter coil at 3T). Figure 6 shows the rSNR maps, and Figure 7 summarizes the peripheral and central rSNR ratios of the 12-ch over-overlapped, the 12-ch conventionally-overlapped, and the 8-ch conventionally-overlapped array. At 7 T, the over-overlapped array does not enhance either the peripheral or the central SNR, and in fact the 12-ch over-overlapped array exhibits slightly lower peripheral SNR compared to the 12-ch conventionally-overlapped array (6.7%~21% lower). At 1.5 T and 3 T, the over-overlapped design increases the central SNR by 4.1%~23.2 %, depending on the coil size and B_0 field. Therefore, at 1.5 T and 3 T, the over-overlapped design has better performance when the target is located deep inside the field of view.

When the sample noise does not dominate, simply increasing the coil number with ~10% overlapping will decrease the central SNR as smaller coils have reduced sample noise ratio. This is consistent with the practical findings in previous publications. For example, a 96-channel 3 T head coil has 5%–20% central SNR decrease compared to a 32-channel coil, depending on imaging combination method [8]; and a 256-channel 7 T head coil shows an average central SNR reduction by a factor of 0.8 compared to a 32-channel coil [9].

3. G-factors in parallel SENSE imaging

The g- or geometry factor acts as a noise amplification factor so a lower g-factor is desirable in parallel imaging. Figure 8 shows maps of the values of $1/g$ with left-right accelerations up to 3 for 8-ch conventionally-overlapped coils, 12-ch conventionally-overlapped coils, and 12-ch over-overlapped coils. The average and maximum g-factors for different coil sizes and B_0 fields are summarized in Figure 9 and demonstrate that coil number plays a crucial role in parallel imaging performance. For example, the average g-factors of 8-ch and 12-ch conventionally-overlapped arrays for 5/7/9-cm-diameter-coils at 3T are respectively 1.95/1.84/1.82 and 1.39/1.36/1.36 in the peripheral area, and 1.63/1.62/1.65 and 1.39/1.36/1.36 in the central area.

The overlapped geometry also affects the g-factor although its impact is much lower than the coil number. With greater overlapped areas, the over-overlapped array has worse (higher) g-factor. This is consistent with previous findings that the less that coils are overlapped, the better g-factor the array can achieve [27]. However, it should be noted that while the increase in g-factor is significant for the peripheral area, in the center the over-overlapped array has almost the same g-factor as the conventionally overlapped array. For example, the average g-factors of 12-ch conventionally-overlapped and over-overlapped arrays for 5/7/9-cm-diameter-coils at 3T are 1.39/1.36/1.36 and 1.51/1.44/1.45 in the peripheral area but 1.39/1.36/1.36 for conventionally-overlapped array and 1.4/1.38/1.37 for over-overlapped array in the center.

Discussion

Loop surface coils are commonly used for multi-coil arrays to maximize the SNR and parallel imaging performance. The loop coils are conventionally overlapped by ~10% to cancel the mutual inductive coupling. This critical 10%-overlapping requirement, however, limits the choice of coil elements for a specified imaging coverage. The relative noise contributions from the coils themselves (Figure 5) increase as loop size decreases so we hypothesized there might be SNR benefits from over-overlapped coils that allow the same coil size to be used while increasing the coil number. In this work, we numerically investigated 12-ch over-overlapped arrays across different coil sizes and different B_0 fields and compared their performances with conventionally-overlapped arrays.

Our simulation results reveal: 1) using an over-overlapped geometry does not benefit intrinsic SNR, either in the peripheral area or in a central area of a cylindrical object; 2) for real situations accounting for coil noise, over-overlapped geometry exhibits higher SNR in the central area compared to conventionally-overlapped arrays when the sample noise is not dominant. Increasing the central SNR is more challenging than improving the peripheral SNR, so that the over-overlapped design is attractive for several MR imaging applications including the deep brain and the prostate; 3) although over-overlapped design increases g-factors compared to conventionally-overlapped design, the increase is only small and mainly affects the peripheral areas (Figure 9).

There are several realistic approaches to implement over-overlapping. The most straightforward and general method is to use low input-impedance preamplifiers to form a

tank circuit and thus suppress the current due to inductive coupling, which also known as “preamplifier decoupling”. The decoupling capability of preamplifier decoupling is dominated by how low the preamplifiers’ input impedance could be. A broadly-accepted criterion of this value is less than 5 ohm. Nowadays, vendor even provides preamplifier with ultra-low input-impedance of 0.1 ohm (WMA3RA-R1, WanTCom, Chanhassen, MN, USA), with which the inductive coupling can be reduced to a sufficient value. In addition to pushing the preamplifier’s input impedance to extremely low values, researchers recently proposed advanced preamplifier decoupling methods such as the novel class E-mode preamplifiers [10] and high impedance/reverse preamplifier decoupling method [11]. In this work, we choose 40% over-overlapped area as this is close to the upper limit of using the preamplifier decoupling methods reported in high impedance coils [11].

We investigated loop arrays that are over-overlapped in only one direction. In this case, loop size increases in only one direction. For arrays that are over-overlapped in multiple directions, the loop size can be further increased in which case the sample to noise factor can also be further increased and the central SNR can be further improved. Loop coils with rectangular shape are also widely used and are preferred in extremity imaging for better coverage along the z-direction. Figures S1–S3 (in the supplemental materials) show simulated SNR maps of rectangular loop arrays in the central transverse slice, and these results are almost the same as those of circular loop arrays (Figures 3 and 6).

Although the over-overlapped design is demonstrated only for receive arrays, it can also be used for transmit arrays. From the principle of reciprocity, the over-overlapped coil is expected to have less power dissipation in the coil itself and thus higher transmit efficiency. The main challenge of the transmit array is to maximize the transmit field’s uniformity for a given SAR or minimize the SAR for a given transmit uniformity. In Deniz et al [28], the transmit coils were lifted so larger elements could be used with the same coil number. They demonstrated that using larger coils produced lower SAR while maintaining a homogenous excitation but at the cost of lower transmit efficiency. Over-overlapping also enables larger coils without lifting them off, so it is likely beneficial to transmit performance without costing transmit efficiency. However, additional decoupling treatments will also likely be needed.

Conclusion

We numerically investigated the receive performance of over-overlapped arrays with different sizes across different static magnetic fields. Our simulation results revealed that over-overlapped designs improve the SNR in deep areas when the coil noise dominates, which might lead to improved coil designs for some special applications.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Diagrams and simulation models of the conventionally overlapped coils (10 % overlapped area) and over-overlapped coils (40% overlapped area). (a) Diagram of 2ch conventionally overlapped coils. (b) 2ch over-overlapped coils. (c) Simulation model of an 8-ch array with circular loops. (d) Simulation model of a 12-ch over-overlap array with the same circular loops as Figure 1c. (e) Simulation model of a 12-ch loop array with elliptical loops. For a conventionally overlapped array, the coil size has to shrink to accommodate more coils.

Figure 2.
Illustration of how to define areas for peripheral SNR and central SNR calculation.

Figure 3.

Normalized intrinsic SNR (iSNR, coil noise not considered) maps in the central axial slice for coil sizes 5 cm, 7 cm and 9 cm, and B_0 fields 1.5 T, 3T and 7T.

Figure 4.

Intrinsic SNR (iSNR) ratios of the 12-ch over-overlapped and the 12-ch conventionally-overlapped arrays to the 8-ch conventionally-overlapped array. (a) iSNR ratio in the peripheral area. (b) iSNR ratio in the central area.

Figure 5. Sample noise ratio of circular loops and elliptical loops (shrink horizontally) for different coil sizes and B_0 fields.

Figure 6.

Normalized real SNR (rSNR, coil noise is considered) maps in the central axial slice for coil size across 5 cm, 7 cm and 9 cm, and B_0 field across 1.5 T, 3 T and 7 T.

Figure 7.

Real SNR (rSNR) ratio of the 12-ch over-overlapped and the 12-ch conventionally-overlapped array to the 8-ch conventionally-overlapped array. (a) rSNR ratio at the peripheral area. (b) rSNR ratio at the central area.

Figure 8.

1/g-factor maps of 8-ch conventionally-overlapped array, 12-ch conventionally-overlapped array with elliptical loop and the 12-ch over-overlapped array. The acceleration factor (R) varies from 2 to 4 in left-right direction. Geometry (g -) factor is a mathematical expression of the ability of a receiver coil array to retain SNR in accelerated MR acquisitions. The lower g -factor means higher SNR in accelerated images.

Figure 9.

G-factor comparisons for different coil sizes and B_0 fields. (a) Averaged g-factor at the peripheral area. (b) Averaged g-factor at the central area. (c) Maximum g-factor in the whole slice.