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Effects of proximity and noise level of phased array coil elements on overall signal-to-noise in parallel MR spectroscopy

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Abstract

Parallel imaging using phased array coils facilitates accelerated magnetic resonance imaging (MRI) and spectroscopy (MRS). Parallel data reconstruction requires the combination of data from individual coil elements, but limited combination algorithms currently exist for higher-order phased arrays and MRS data. Here, we present a systematic framework for identifying coil proximity-related signal inhomogeneities and noise levels in phased array coils that may affect sensitivity of parallel MRS. Single-voxel MRS was acquired in nine voxel positions in a brain spectroscopy phantom on a 3T whole-body MR scanner using commercially available 64-, 32-, and 20-channel phased array coils. Spectra produced by individual coil elements were combined using both a signal-to-noise ratio (SNR) threshold and based on the position of individual coil elements. SNR and metabolite Cramer-Rao lower bounds (CRLBs) from the final combined spectra were used as metrics to compare combination strategies and the effects of the phased array geometry and individual coil proximity. Comparisons were performed using one-way repeated measures ANOVA and post-hoc Tukey’s range test (p<.05). The 32-channel phased array coil produced the highest overall SNR compared to the 64-channel (p=.0009) or 20-channel coils (p=.003). Low SNR spectra from individual coil elements in the 64-channel coil can reduce the overall SNR when simply combining spectra from all elements. SNR varied significantly as a function of voxel position (F=58.3, p<.0001) and SNR threshold for all phased arrays (p<.05 for 64-, 32-, and 20-channel coils). Metabolite CRLBs were dependent on the combination strategy. We demonstrate the importance of the sampling voxel position and coil proximity on overall SNR in parallel MRS data acquisition, with significant SNR improvements after selectively filtering individual spectra based on pre-determined SNR thresholds which must be optimized for each phased array coil element and volume of interest.

Keywords

magnetic resonance spectroscopy; parallel imaging; phased array; signal-to-noise

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1. INTRODUCTION

Parallel imaging, facilitated by multi-channel phased array receive coils, is widely implemented in magnetic resonance imaging (MRI) and spectroscopy (MRS) for accelerated acquisition [1–3]. An important step for phased array data reconstruction is the combination of signals collected from individual coil elements to maximize the overall signal [2–4]. Current combination methods are often optimized for parallel imaging based on the strong water signal but fewer methods, particularly for higher-order phased arrays, exist specifically for MRS that targets metabolites with concentrations much lower than water [5–7]. Increasing the number of independent coil elements (N) theoretically improves the signal-to-noise ratio (SNR) in the signal averaged spectrum by $\sqrt{N}$ [8], but empirical limitations such as geometry-dependent coil sensitivity and high levels of heterogeneous noise reduce the observed SNR. As net SNR contributions from individual coil elements vary, coil elements that add mostly noise may reduce the overall combined SNR. This noise propagation is particularly challenging for MRS due to the relatively low concentrations of metabolites and high levels of noise when recording tissue spectra. Identification of the most important factors contributing to the sensitivity and SNR of the combined spectra is necessary for metabolite quantitation and development of a widely-applicable and robust coil combination algorithm for parallel MRS. Improved collection and analysis of phased array data will enable full exploitation of current phased array and parallel data recording technologies. Here, we systematically investigated the effects of the nonuniformity of phased array coil construction, number of individual coil elements, coil-specific noise levels, and proximity of the coil elements to the sampling voxel on spectral SNR and metabolite quantification using a brain spectroscopy phantom.

2. MATERIALS AND METHODS

Single voxel MRS was performed in a spherical brain spectroscopy phantom [9] (#2152220, GE Healthcare, Chicago, IL; see Supplementary Material) with a 32-channel (12 anterior and 20 posterior elements) phased array head coil on a Siemens 3 T Tim Trio MR scanner, and a 64-channel (24 anterior and 40 posterior elements) phased array head and neck coil and a 20-channel (10 anterior and 10 posterior elements) phased array head and neck coil on a Siemens 3 T Prisma MR scanner (Siemens Healthcare, Erlangen, Germany). MRS data were acquired with the following parameters: TR=2 s; TE=30 ms; 64 averages; 1200 Hz bandwidth; 2048 complex data points; and nominal voxel size of 2×2×2 cm$^3$. A water suppressed scan with the same parameters was also collected (50 Hz water suppression bandwidth) for phase and eddy-current corrections. A 3-plane localizer image (TR=7.7 ms; TE=3.67 ms; 20° flip angle, 280×280 mm$^2$ field of view, 512×512 pixels, 6 mm slice thickness) was used to place the 7 sampling voxel. The phantom was positioned in the phased array coils such that the center of the phantom was as close to the isocenter as possible. Although the arrangement of individual coil elements differs between phased array coils [10], spectra were acquired from the voxel at the isocenter and from voxels shifted 30 mm from the isocenter in anterior, posterior, right, left, head, and feet directions, and 50 mm in anterior and posterior directions (Figure 1). For the 64-channel phased array coil, all 64 coil elements were selected manually to collect data from every element, as the scanner
Software automatically turns on select, but not all, coil elements depending on the position of the sampling voxel.

Data was processed with LCModel [11] (version 6.3-1H) and Matlab (R2015a, Mathworks, Natick, MA). SNR was calculated in the frequency domain using the maximum signal of the N-acetylaspartate (NAA) peak divided by the standard deviation of the noise measured from the last 100 points of the spectrum, and was used to assess data quality and as a threshold to select spectra from individual coil elements. Using Equation 1, n spectra from individual coil elements ($S_i(\omega)$) were phased and baseline corrected. Individual spectra above an SNR threshold, $A$, (0, 10, 20, or 30) were included in the final combined spectrum ($S(\omega)$) and spectra below the threshold were excluded.

$$S_t(\omega) = \sum_{i=1}^{n} S_i(\omega) \cdot \chi_A(SNR)$$

Individually, $\chi_A(SNR) = \begin{cases} 1 & \text{if } SNR \geq A \\ 0 & \text{if } SNR < A \end{cases}$

Individual spectra were also combined based on the position of the individual coil elements (e.g., coil elements in the neck, anterior, or posterior sections of the phased array).

To identify the effects of coil proximity and noise levels on metabolite quantification, data were processed with LCModel after combining individual time domain data to include spectra from 1) all individual coil elements (SNR > 0); 2) individual coil elements that produced spectra with SNR above a threshold; and 3) either all anterior or all posterior individual coil elements without SNR discrimination. Cramer-Rao lower bounds (CRLB), which is the standard deviation of the peak fit, and the absolute difference between observed and actual metabolite/creatine (Cr) ratios were used as metrics to determine the effects of the threshold and combination strategy on metabolite quantification. For the voxels placed in the isocenter, anterior, and posterior positions, metabolite CRLBs and the metabolite/Cr ratios were calculated for the combined spectra. Differences in the SNR of combined spectra as a function of voxel position and SNR threshold; CRLB differences as a function of the signal combination strategy; and metabolite/Cr ratio differences as a function of the signal combination strategy were calculated using one-way repeated measures analysis of variance (ANOVA) and post-hoc analysis was performed with Tukey’s range test using SPSS Statistics (version 24.0, IBM, Armonk, NY). The exception was the post-hoc analysis of CRLB differences for the voxel 30 mm anterior to the isocenter, where the Bonferroni correction was used as the Tukey’s range test could not be applied. Significance was determined with $p < .05$.

3. RESULTS

Figure 2 shows representative normalized spectra from the isocenter voxel produced by combining signals from selected coil elements of the 64-channel phased array coil using Equation 1. The maximum SNR was observed when combining spectra with SNR ≥ 20 (Figure 2; Table S1, Supplementary Material). Individual spectra produced from three unique coil elements in the 64-channel coil are shown in Figure S1, demonstrating...
individual coil element differences in spectral SNR. Several coil elements (e.g., coil 22) produced little signal and mostly noise, suggesting that these data may only contribute noise to the overall combined spectra if included. Improvements in SNR of the combined spectra were achieved by applying higher SNR thresholds which effectively eliminated the contribution of noise-heavy spectra. Furthermore, the position of the sampling voxel also affects spectra produced by individual coil elements, demonstrated by comparing the normalized combined spectra obtained from the isocenter voxel (Figure 3A–C) and a voxel shifted 30 mm anterior from the isocenter (Figure 3D–F) acquired with 64-, 32-, and 20-channel phased array coils using a threshold of SNR ≥ 10 for selecting individual coil elements for data combination. The combined spectra obtained from the isocenter exhibited higher SNR for all phased array coils tested compared to the voxel shifted 30 mm anterior. However, for this off-center voxel, the 20-channel phased array coil yielded the highest SNR in contrast to the higher order phased array coils (i.e., 32 and 64-channel) when using the SNR threshold for signal combination.

For the phased array coils and sampling voxel positions tested, as shown in Figure 4, the overall maximum combined spectral SNR was influenced by phased array coil geometry and coil element arrangement, sampling voxel position, and the SNR threshold used for combining individual data. Data including SNR values, percentage and number of coil elements used in the combined spectrum, and relative changes in SNRs for all voxel positions are presented in Table S1. Comparing the SNRs of combined spectra for all voxel positions and an SNR threshold > 0 (i.e., all individual spectra included in combined spectrum), the mean SNRs ± standard error of the 64-channel (53.5 ± 8.6), 32-channel (111.9 ± 16.9), and 20-channel (60.1 ± 9.8) phased array coils were significantly different (F=12.4, p=.0006). The SNRs of combined spectra also varied significantly between the voxel positions (F=58.3, p<.0001) for all phased array coils. Post-hoc analysis revealed that, despite expected theoretical SNR improvements with more coil elements, the 32-channel coil produced significantly higher overall spectral SNRs compared to that of the 64-channel (p=.0009) or the 20-channel coils (p=.003) when using all coil elements (SNR > 0). No difference in SNR was observed between the 64-channel and 20-channel coils (p=.9).

However, increasing the number of the coil elements did not yield a net gain in SNR as shown in the comparison of 64- and 32-channel coils. For all voxel positions, >90% of the coil elements in the 32-channel coil produced spectra with SNR ≥ 10. In comparison, the 64-channel and 20-channel coils had as few as 10% and 70%, respectively, of individual coil elements that produced spectra with SNR ≥ 10, depending on the voxel position.

For comparison, we also examined the effect of coil proximity to the sampling voxel. For the voxels shifted in the anterior and posterior directions, spectra acquired from the 64-channel coil were combined using only anterior or posterior coil elements to determine the effect of coil proximity, rather than individual SNRs, on the overall SNRs of the combined spectra (Figure S2, Table S2). Similarly, for voxels shifted in the head and feet directions, spectra from only the neck coil elements were combined (Table S2). For a voxel shifted 30 mm anterior from the isocenter, the use of only posterior elements produced spectra with a higher SNR than using only anterior elements despite the fact that these coil elements were further away from the voxel (Figures S2A). This was attributed to the addition of signal when using more coil elements (40 posterior versus 24 anterior). However, as the voxel was
further shifted from the isocenter (50 mm anterior), there was little or no SNR improvement as some posterior coil elements may contribute less signal but more noise to the combined spectrum (Figure S2B). For voxels shifted in the posterior direction from the isocenter, the use of data from only posterior coil elements in the combined spectrum led to increased SNRs for both voxel positions (Figure S2C and D).

When using SNR thresholds (Equation 1) and individual coil positions (i.e., anterior and posterior) as criteria to combine individual spectra, we observed that the combination strategy contributes most to the overall SNR of the combined spectra. Significant differences in SNRs using varying SNR thresholds were observed for all phased array coils (Table 1). For all voxel positions, SNR was maximized when using an SNR threshold (Table S1) rather than using the data from coil elements closer in proximity to the voxel position (Table S2), indicating the impact of noise contributions to the overall SNR.

In addition to the SNR metric, we also measured metabolite concentrations to examine effects of coil proximity and combination method on spectral quantitation (Table S3). CRLB values were dependent on both the SNR as well as coil proximity. Not surprisingly, for voxels shifted in the anterior direction (30 or 50 mm), using only anterior coil elements resulted in lower CRLBs compared to using only posterior coil elements (30 mm, p<.0001; 50 mm, p=.03). Using only anterior coil elements for the voxel shifted 30 mm anterior also produced spectra with the lowest CRLBs, despite the fact that the SNR ≥ 20 threshold produced the highest SNR for the same voxel. Similar results were obtained for the posterior voxels, where using only posterior coil elements reduced CRLB values compared to using only anterior coil elements (30 mm, p=.0008; 50 mm, p=.0005) and the CRLBs were comparable to that observed for spectra combined based on their SNRs. No changes were observed between combination methods for metabolite/Cr quantification, possibly due to the high quality spectra acquired in the spectroscopy phantom.

4. DISCUSSION

Pursuit of high sensitivity continues to be a major technical development focus for in vivo MRS applications. Parallel MRS using phased array coils is an attractive and rational approach. Our results from a systematic analysis of SNR contributions from individual phased array coil elements as a function of coil proximity to the sampling voxel and combination strategies suggest that future development of data combination strategies and design of phased array coils for parallel MRS may need to include the following considerations: 1) empirical determination of the phased array coil rather than default use of the array with the highest number of coil elements (e.g., 32-channel gave the highest overall SNR in our study); 2) selecting data from individual coil elements that produce the highest SNR when there is SNR non-uniformity in the phased array coils; 3) eliminating noise-dominated spectra prior to combination; 4) using data from individual coil elements that are in proximity to the sampling voxel; and 5) understanding the relationship between high SNR and low metabolite CRLBs for the region and metabolites of interest.

Recovery of SNR in the overall combined spectra can be achieved by discarding individual spectra that contribute mostly noise using an SNR threshold. However, we also observed that
metabolite quantification and spectral fit measured by CRLBs do not improve linearly with SNR. Spectra with the lowest CRLBs are not necessarily those with the highest SNR, which must be investigated further particularly for data combination strategies that rely solely on SNR. While it is conceivable that individual coil elements in close proximity to the sampling voxel and with high sensitivity should contribute to better spectral quality, this was not always the case for the commercial array coils tested. For example, using only neck coils of the 64-channel phased array coil for a voxel closer to the neck did not produce spectra with the highest SNR as the neck coils in the array may be less sensitive than coils closer to the isocenter. Similarly, using only anterior coils for a voxel shifted in the anterior direction did not necessarily improve SNR compared to using only posterior coils, particularly in the case where the number of posterior coils is much higher (40 versus 24) and the voxel is close to the isocenter. Further optimization of the spatial arrangement of coil elements and architecture of phased array coils may improve or maximize the coil performance and data quality.

While several previous studies report empirical and data driven combination methods that often apply a non-zero amplitude weighting to all individual signals [5, 7, 12, 13], we utilized a simple combination method that excludes spectra entirely with SNR values below a given threshold or based on the position of individual coil elements. By systematically comparing the effects of the non-uniform spatial arrangement of the coil elements (i.e., both the 64- and 32-channel phased array coils contain more posterior than anterior coil elements) and the number of “effective” or high SNR coil elements, we observed that adding spectra to increase net signal can, but does not necessarily, increase the SNR of the combined spectrum. Empirical selection of phased array coils as well as individual coil elements within an array, followed by subsequent elimination of spectra that contribute only noise, can improve overall spectral SNR. This is important when studying localized lesions in the brain where the sampling voxel is rarely at the isocenter, and the influence of noise-only spectra may have larger effects on metabolite detection sensitivity.

We acknowledge several limitations in our study. Notably, current analysis was performed using a phantom rather than in vivo. However, challenges associated with in vivo experiments, such as partial volume effects, motion, and other artifacts, were largely eliminated in the motionless and homogeneous brain spectroscopy phantom studies. The use of the phantom also demonstrated that even in the case when the substrates are homogenous, uniform, and produce spectra with high SNR and narrow linewidths, the effects of varying coil sensitivity, coil arrangement and construction of phased array coils, and data combination strategies on overall SNR are significant. Whereas the use of single voxel MRS was an optimal starting point to identify the effects of physical coil proximity and sensitivity using phased arrays, future studies may need to be performed using multi-voxel MRS imaging to facilitate accelerated parallel imaging across multiple voxels. The noise correlation between individual coils was assumed to be negligible, which may need to be considered in future studies [5, 13].

In conclusion, we demonstrated the importance of noise reduction and SNR uniformity in parallel spectroscopy data acquisition and combination. We also showed several pitfalls in current parallel MRS implementation using commercial state-of-the-art phased array coils,
and presented potential approaches to improve coil combination algorithms and the design of new phased array coils.

**Supplementary Material**

Refer to Web version on PubMed Central for supplementary material.

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**References**

### Highlights

- A framework for identifying phased array coil signal inhomogeneities is proposed.
- Sampling voxel position and coil proximity must be optimized for each phased array.
- Higher-order phased arrays do not always produce the highest signal-to-noise ratio.
Figure 1.
MRS voxel positions in a brain spectroscopy phantom. iso=isocenter; A=anterior; P=posterior; R=right; L=left; H=head; F=feet; #=unidirectional shift, in mm, of voxel position from isocenter.
Figure 2.
Normalized combined spectra acquired using a 64-channel phased array head coil from a voxel placed at the isocenter using SNR thresholds of A) >0; B) ≥10; C) ≥20; D) ≥30. Inset is the region between 1.4 – 1.0 ppm. # Coils indicates the number of individual coil elements that produced spectra within the SNR threshold.
Figure 3.
Comparison of normalized combined spectra acquired using 64-, 32-, and 20-channel phased array head coils from a voxel placed at the isocenter (A, B, C) and 30 mm anterior from the isocenter (A30; D, E, F), respectively. All spectra were combined using an SNR threshold ≥10. # Coils indicates the number of individual coil elements that produced spectra within the SNR threshold.
Figure 4.
Comparison of combined spectral SNR as a function of voxel position (see Figure 1) using 64-, 32-, and 20-channel phased array head coils and SNR thresholds of A) >0; B) ≥10; C) ≥20; D) ≥30. iso=isocenter; A=anterior; P=posterior; R=right; L=left; H=head; F=feet; #=unidirectional shift, in mm, of voxel position from the isocenter.
Table 1

Statistical comparison of SNR differences based on voxel position and SNR threshold.

<table>
<thead>
<tr>
<th>Phased Array Coil</th>
<th>Voxel Position</th>
<th>SNR Threshold</th>
<th>Post-Hoc Analysis</th>
</tr>
</thead>
<tbody>
<tr>
<td>64-channel</td>
<td>F=55.1</td>
<td>F=21.7</td>
<td>SNR 20 &gt; 10 (p&lt;.02)</td>
</tr>
<tr>
<td></td>
<td>p&lt;.0001</td>
<td>p&lt;.0001</td>
<td>SNR 30 &gt; 10 (p&lt;.001)</td>
</tr>
<tr>
<td>32-channel</td>
<td>F=82.0</td>
<td>F=20.1</td>
<td>SNR 30 &gt; 10 (p&lt;.002)</td>
</tr>
<tr>
<td></td>
<td>p&lt;.0001</td>
<td>p&lt;.0001</td>
<td>SNR 20 &gt; 0 (p&lt;.0001)</td>
</tr>
<tr>
<td>20-channel</td>
<td>F=95.0</td>
<td>F=30.1</td>
<td>SNR 30 &gt; 0 (p&lt;.001)</td>
</tr>
<tr>
<td></td>
<td>p&lt;.0001</td>
<td>p&lt;.0001</td>
<td>SNR 20 &gt; 0 (p&lt;.0001)</td>
</tr>
</tbody>
</table>

F values correspond to differences in SNR as a function of position (Voxel Position) and as a function of SNR threshold within voxel positions (SNR Threshold) calculated with one-way repeated measures ANOVA. Post-hoc analysis to compare SNR as a function of the SNR threshold for each phased array was performed with Tukey’s range test, p<.05.