Central and Peripheral SNR as a Function of Number of Active Coils for 32 and 96 Channel Receive Coils at 3 Tesla

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Introduction: When using arrays with large numbers of small coils there is commonly a concern about the sensitivity provided by the small elements at distances far from the array. Theoretical arguments and experimental evidence show that for body noise dominance and low coupling, a planar array is as sensitive deep in the sample as a single coil of the same overall dimensions as the complete array, but benefits from the higher sensitivity of the small elements close to the array [1]. Similarly, a cylindrical array should provide the same central SNR as a birdcage coil of similar dimensions, but considerably higher SNR in the periphery [1]. Helmet-shaped array coils with elements arranged over the entire dome of the head have been described with 23, 32 and 96 channels [2,3,4]. As the number of elements increases, and the size of the individual elements therefore shrinks, it is important to understand how the SNR in the center of the sample increases as a function of the number and distribution of coil elements around the head, whether in fact the SNR may decrease beyond a certain number of coils due to increased opportunities for noise coupling, and therefore how much benefit is gained from costly and complicated designs with 96 channels or more.

Methods: Data were acquired with 32 and 96 channel helmet arrays constructed on identical close-fitting formers [3,4] using a prototype Tim Trio 3T clinical scanner with 128 receive channels (Siemens Medical Solutions, Erlangen Germany). The coil elements were critically overlapped to minimize inductive coupling, with element diameters of approximately 85mm for the 32 channel coil and 48 mm for the 96 channel coil. For the 32 channel coil it was possible to acquire multiple SNR data sets with different subsets of coil elements selected, and the other coils detuned. This approach was not possible with the 96 channel coil because only 32 PIN bias lines are used for element detuning, each split to control three different coil elements. For this coil a full 96 channel data set was acquired, plus a noise scan, and the noise correlation matrix was calculated. To synthesize the SNR that would be provided by different subsets of coils, only those channels were combined, and in the case of optimum SNR reconstruction, the noise correlation between those elements was taken from the full 96 channel noise correlation matrix. The validity of this method was examined by performing the same synthesized reconstructions of sequentially more elements for the 32 channel coil from the full 32 channel data set, where this could be compared to the data acquired with only specific subsets of coils active. The order in which channels were added in sequential reconstructions was the same for both coils- first adding only those coil elements in the imaging plane, then the coil elements over the dome of the helmet, and finally the lower rows of coils at the bottom of the helmet (Fig. 1). SNR comparisons were made using proton density gradient echo images (TR/TE/flip/slice = 200ms/4.07ms/20deg/3mm, 256x256, FoV=220mm) obtained in human scans, acquiring scans both with and without RF excitation, in an axial plane passing just above the Corpus Collusum.

Results: Figure 2 shows the central SNR for Optimum SNR (SNR_{opt}) reconstruction as a function of coil number for the two coil arrays. Data from the 32 channel array are shown both for the "true" case where only that subset of elements were switched on during the acquisition (dotted line), and for "synthesized" SNR calculated by reconstructing only selected channels from a full 32 channel data set (dashed line). For low coil numbers the synthesized data underestimates the SNR by as much as 25%, but both curves follow the same trend and naturally converge for 32 channels. Examination of the noise correlation coefficient matrices for the different data sets (not shown) show that the correlation between a given pair of coils only fluctuates about 5% as other coils are added, suggesting that using values from the full data set is reasonable. What is not taken into account is that the SNR of the individual coil elements also decreases as more coils are added because of residual coupling. Using these reduced SNR uncombined images to synthesize SNR from subsets of coils results in an underestimate of the true SNR that would be obtained if those were the only coils active.

One key feature of the plots in Fig. 2 is that the central SNR_{opt} is monotonically increasing with added elements, with roughly equal contribution from all but the lower coils (in red). The SNR increase offered by the addition of the "top coils' (in blue) is significant. These "top coils" have been the subject of some debate, since most of them are oriented with their main axis along the Z-direction, which is generally considered inefficient and gives rise to individual coil profiles with nulls. However, the 32 and 96 channel coil designs have demonstrated that a continuous overlapped array over the dome of the head provides very high SNR in the vertex, an area typically poorly served by other coil designs. It is of interest here to note that these coils even contribute to the central SNR of this deeper slice (approximately 55mm from the top of the head). In contrast the coils added to the bottom row brings little SNR to the central ROI, especially for the 32ch array.

Figure 3 shows the same plots but for Root Sum of Squares (RSS) reconstruction, where the noise correlation is not taken into account. Here the SNR is no longer monotonically increasing with the addition of new coils. The 96 channel data in particular show two kinks where the addition of the next couple of coils actually decreased the central SNR. Also, the addition of coil elements lower on the helmet only decreased the central SNR. This underscores the importance of moving to more routine use of optimal array combinations than is currently used. The penalty for including too many coils for this particular slice is about 5%.

Figure 4 shows SNR profiles through the axial slice from the "true" 32 channel data set, for the cases of a single active element, 10 elements (all the coils in the imaging plane), 17 elements (including the dome) and the full 32. Once all the coils in the imaging plane are active, the additional gains are mainly in central SNR.

Conclusions: While there are decreasing gains in central SNR from adding additional coils around a sample, adding a coil never decreases SNR_{opt}. In contrast, it is possible to see a drop in central SNR (~5%) with RSS from added elements. Although central SNR will be penalized, SNR will be increased near the added coil elements. Finally, with increasing numbers of coil elements, the benefits of Optimum SNR reconstruction become more and more compelling.

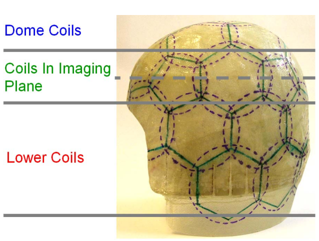

Fig. 1 Schematic antenna geometry for 32 channel coil and classification of coils

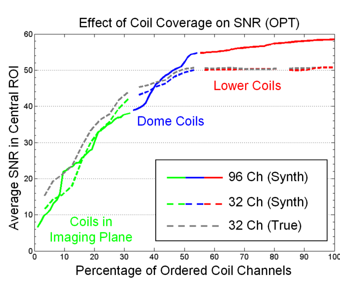

Fig. 2. Optimum SNR in the center of the slice as a function of the number of coil channels used

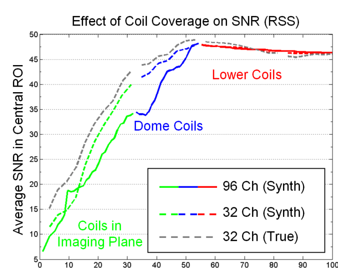

Fig. 3. Root Sum of Squares (RSS) SNR as a function of channels used

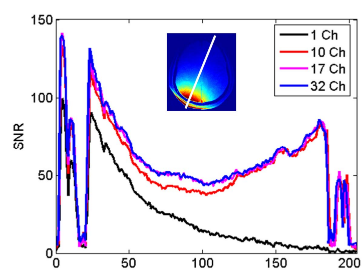

Fig. 4. Profiles through Optimum SNR reconstructions for the 32 channel "true" data set.

[1] Wright, SM et.al. NMR Biomed, 1997 **10**(8) p.394-410 [2] Wiggins GC et. al. Proc ISMRM 2005, p671 [3] Wiggins GC et. al. Mag Reson Med. 56:216-223 (2006) [4] Wiggins GC et. al. Proc ISMRM 2007,