Design of a 96-Channel Bilateral Prone Breast Array for High Performance Parallel MRI

Mike J Smith[1] and Scott B King[1]
[1]*National Research Council of Canada, Winnipeg, Manitoba, Canada*

Introduction

Array coils with many receive channels can provide parallel imaging benefits and improved SNR, enabling faster imaging sequences through increased acceleration factors. The SNR benefits are only realized when the coil losses remain negligible relative to the sample losses [1], forming a relationship between ideal coil size/geometry, system frequency and the depth of the structures of interest. Implementations of 128 and 96-Channel receive array coils include applications for torso and head imaging [2-4]. Here, we investigate prone positioned MR breast imaging, an application where the surface area to anatomical volume relationship provides a natural fit for high channel count receive arrays. To investigate SNR, we have modeled 96-Channel bilateral receive arrays for 3T MRI applications, using a realistic phantom for comparisons to a 16-Channel bilateral design based on the same array footprint and conformal surface. We consider different design structures to investigate how important the design step is for high density receive arrays by considering the SNR and parallel imaging implications.

Theory/Methods

The 96-Channel bilateral breast array was equally split for each lateral and medial side of the breasts. Simulations were performed using SEMCAD X (Schmid & Partner Engineering, Zurich). The simulation phantom consisted of uniform tissue ($\sigma = 0.17 S/m, \varepsilon_r = 5.2$), with electrical parameters designed to match a fat/skin mixed tissue at our target frequency of 123.2MHz. The 16-Channel benchmark bilateral breast array (BM) used 4-channels on the lateral and medial side of each breast. Conductors for the 96-Channel designs were placed on the same surface as the benchmark coil, with 8mm separation from the phantom. Typically, A-P phase encoding is avoided for prone breast imaging due to cardiac and respiratory motion [5]. For lateral design, we investigated overlap based regular grid arrays of 8x3 and 6x4 (S-I x A-P) sets of coils. For the 48-channel medial portion, we considered two independent 8x3 panels and a 7x7 structure (6 sternum loops). For SNR coverage and g-factor robustness, octagonal loops with staggered-rows were used for all designs. We also investigated hybrid structures of these arrays between the lateral and medial sections. SNR and g-factor analysis was performed using Musaik (Schmid & Partner Engineering, Zurich).

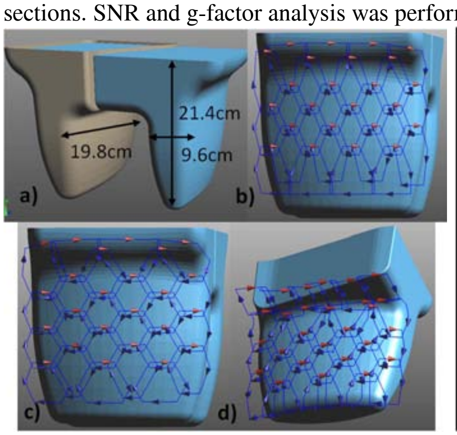


Fig. 1: a) Bilateral breast phantom, **b)** 8x3 lateral, **c)** 6x4 lateral, **d)** 7x7 medial.

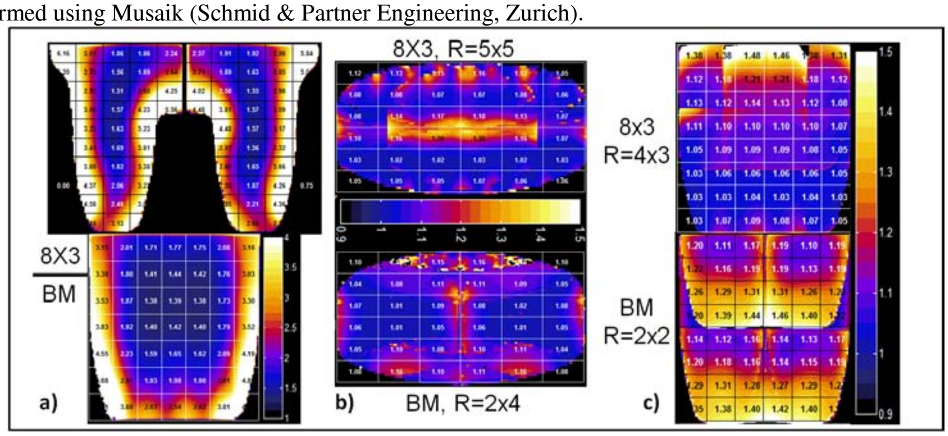
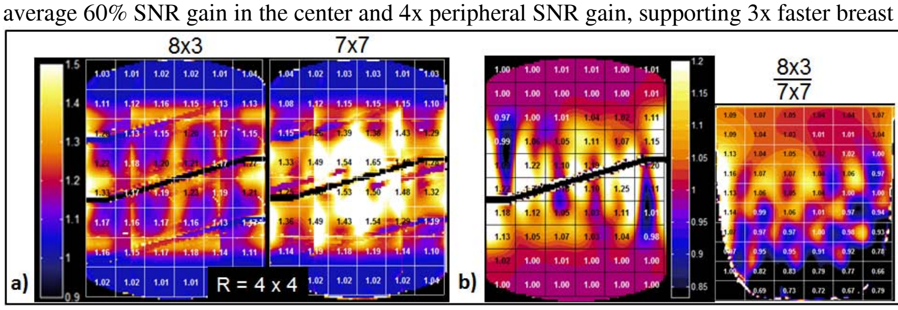


Fig. 2: a) SNR ratio maps for central axial and sagittal slices (96-Ch/16-Ch), **b)** Central breast coronal 2D g-factor maps (only single breast shown), **c)** Central sagittal 2D g-factor maps.

Results/Conclusions

We show a small subset of the data to give insight into SNR and g-factor behavior between the designs. Fig. 3 compares medial layouts for the 96-Channel array. We see better g-factor and SNR performance from the 8x3 architecture in the sternum, but the 7x7 array has better SNR in the nipple region. Fig. 4 shows that the 8x3 lateral design has better S-I acceleration capabilities compared to the 6x4 design. Fig. 2b) and c) show reasonable g-factors from 3x higher accelerations for the 96-Channel array relative to the 16-Channel benchmark. Fig. 2a) demonstrates the 96-Channel array will average 60% SNR gain in the center and 4x peripheral SNR gain, supporting 3x faster breast MRI than currently available with 16-Channel coils.



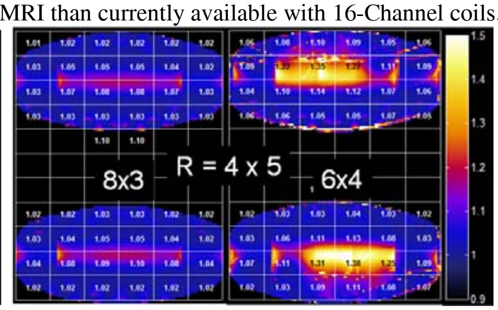


Fig. 3: a) Coronal (sternum) g-factor maps comparing medial designs, **b)** SNR ratio maps for (8x3 vs 7x7) 96-Ch arrays, left - coronal (sternum) and right - sagittal (2cm medial of center) slices.

Fig. 4: g-factor maps comparing lateral designs for R = 4 x 5 (L-R x S-I) for mid-breast coronal slice.

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