SNR and parallel imaging performance of a 32-channel array for human brain imaging at 7 Tesla

Jacco A. de Zwart*, Peter van Gelderen, Shumin Wang, Jeff H. Duyn

LFMI, NINDS, National Institutes of Health, Bethesda, MD, USA

Introduction

Multi-channel coil arrays [1] have led to substantial sensitivity and resolution improvements in MRI of human brain. These benefits generally increase with the number of coil elements but level off at large element counts. Another advantage of increased array size is increased parallel imaging (PI) performance. Here, actual data from a 32-channel array at 7 T were compared with a simulation of the array and with numerically derived coils with lower element counts to investigate these two aspects.

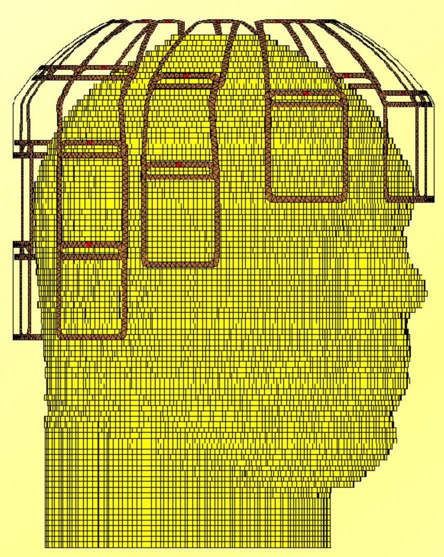

Figure 1: Coil element layout.

Methods

The 32 channel array was designed in cooperation with, and built by, Nova Medical (Wilmington, MA) [2]. It consists of a 4-by-5 grid of elements in the posterior half, and 5 columns with respectively 3, 2, 2, 2 and 3 elements in the anterior half (Figure 1). Columns are gapped, perpendicular to z (gap size is 30% of element width), whereas elements in a column overlap (in the z-direction) to avoid signal dropouts. Finite element simulations of this and related coil layout were also performed [3]. In these simulations, per-element coil noise power was assumed to decrease to the power 0.5 with the number of elements, whereas sample noise power was decreased to the power 1.5.

Image- and noise-data were acquired on normal volunteers (n=2) on a 7 T GE scanner (120×96; twelve 1-mm slices; 9 mm gap; 225×180×111 mm^3 coverage). Since comparisons between different coil designs are difficult, acquired coil data were numerically combined as described earlier [4] to yield a 15-, a 10-, two 8-, two 5-, two 2- and a 1-element coil. Note that such combinations suffer from more component noise than actual arrays that consist of less (larger) elements. SNR over the entire brain and a central 15×15×21 mm^3 region were evaluated, as well as the average g-factor (over whole brain) for 20 different PI acceleration rates. Since the scanner was equipped with only 16 receivers, image data were acquired in 2 groups of 16, and coil noise data in 6 different groups of 16 to allow complete coil noise correlation assessment. (All 32 elements were connected to pre-amplifiers at all times.)

Results and Discussion

Figure 2 shows SNR (relative to 1-channel) as a function of the number of elements. SNR in the center remains relatively constant, but improves about a factor of 4 on average over the brain for 32 independent channels. Note however that the gain from 15 to 32 channels is only 19%, demonstrating that SNR gain levels off when many elements are used.

Figure 3 shows the average PI g-factor in the head. Multi-dimensional acceleration outperforms a similar overall rate in 1D, and acceleration in the largest dimension (AP>LR>SI) performs best. Up to ~6-fold acceleration can be achieved with a mean PI penalty of <~20%. This corresponded very well to the simulated array's PI performance; the correlation coefficient of the mean g-factor for the 20 acceleration rates is 0.93 when compared to the actual array.

channels	average g-factor (for rate $[R_{AP},R_{LR},R_{SI}]$)											
	[2,1,1]	[1,2,1]	[1,1,2]	[3,1,1]	[1,3,1]	[2,2,1]	[2,1,2]	[1,2,2]	[3,2,1]	[2,3,1]	[3,1,2]	[1,3,2]
32	1.02	1.02	1.05	1.09	1.11	1.05	1.08	1.07	1.14	1.17	1.17	1.20
15	1.03	1.02	1.16	1.11	1.15	1.06	1.21	1.20	1.20	1.23	1.35	1.41
10	1.03	1.03	1.96	1.13	1.18	1.08	2.19	2.53	1.29	1.32	2.90	3.95

Table 1: Average g-factor for the actual 32- and numerically-combined 15- and 10-channel arrays. In the 10-channel array the coils in each row are combined. The 15-element coil is similar, but the elements in the occipital half consist of two rows of 5 elements.

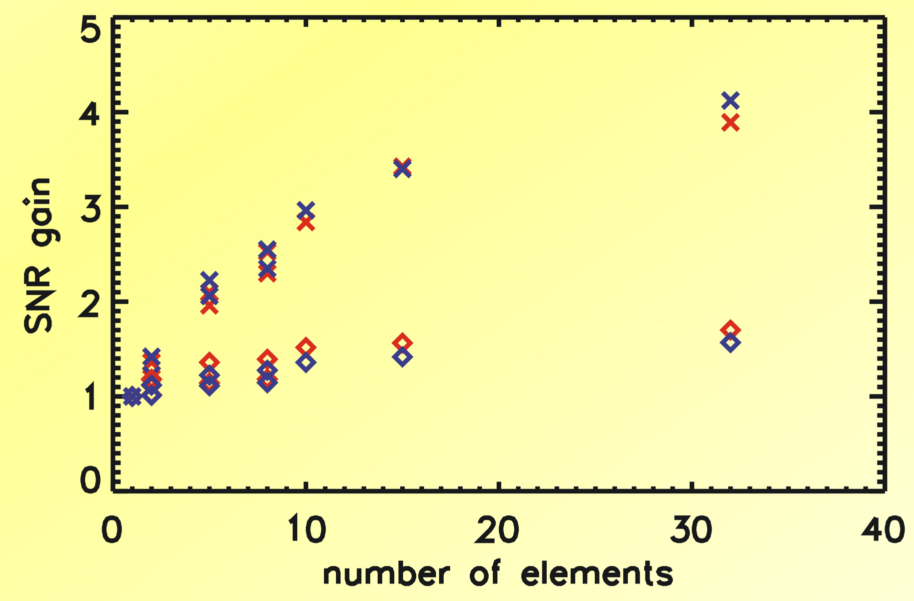

Figure 2: SNR gain as a function of the number of coil elements, averaged over the entire brain (×) and in the center (◊) for the two volunteers (indicated by symbol color).

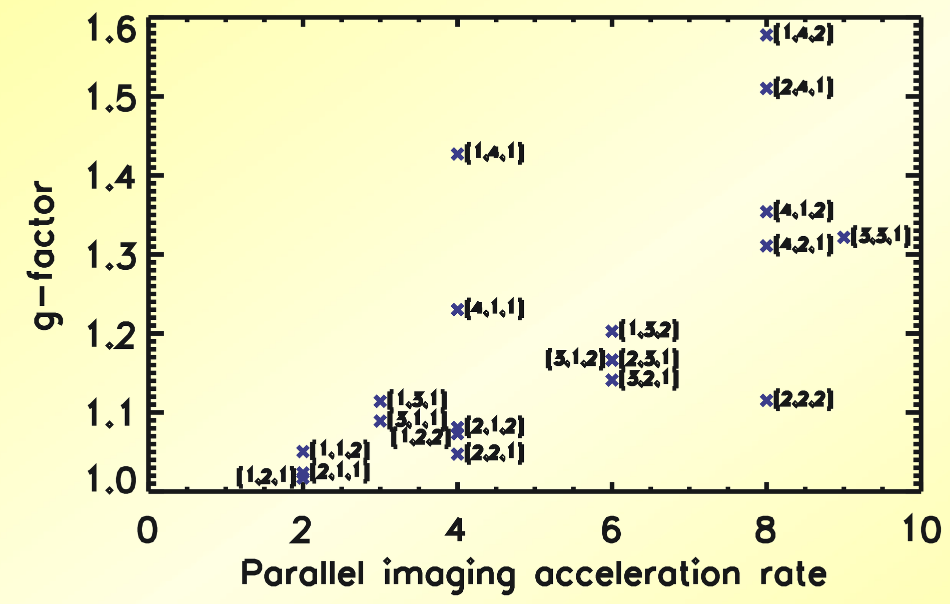

Figure 3: Average (first over brain, then volunteers) g-factor for 20 different PI acceleration rates for the 32-channel coil. Labels at the right of each symbol indicate acceleration in the AP, LR and SI direction as $[R_{AP},R_{LR},R_{SI}]$. (Except [1,2,1], [1,2,2] and [3,1,2], placed left of symbol due to space constraints).

Table 1 compares the mean g-factors shown in Figure 3 with the numerically-combined 10- and 15-channel array. Not surprisingly, the 32-element array outperforms the smaller arrays in all cases, most notably for R>3. (The 10-channel array performs especially poorly for R_{SI}>1 since it consists of only a single row of elements.)

Simulations were also used to assess performance potential of more than 32 elements (Figure 4). Benefits largely depend on coil loading (the relative contribution of coil and sample noise). For well-loading coils (large unloaded-to-loaded Q ratio, see Figure 4), substantial performance benefits still exist. The actual 32-channel coil loads approximately 2:1.

Conclusion

The high-performance 32-channel array yields approximately 4-fold SNR gain compared to a similarly sized single-channel coil and up to 6-fold PI acceleration can be achieved with minor PI penalty. A large number of elements is particularly advantageous for image acceleration, while benefits for average SNR start to level off. Low-noise coil electronics, resulting in increased coil loading, are essential to yield the full potential of even larger arrays.

References

[1] MagnResonMed 16 1990 p. 192
[2] ISMRM 2007, p. 242
[3] ISMRM 2007, p. 1008
[4] MagnResonMed 51 2004, p. 22

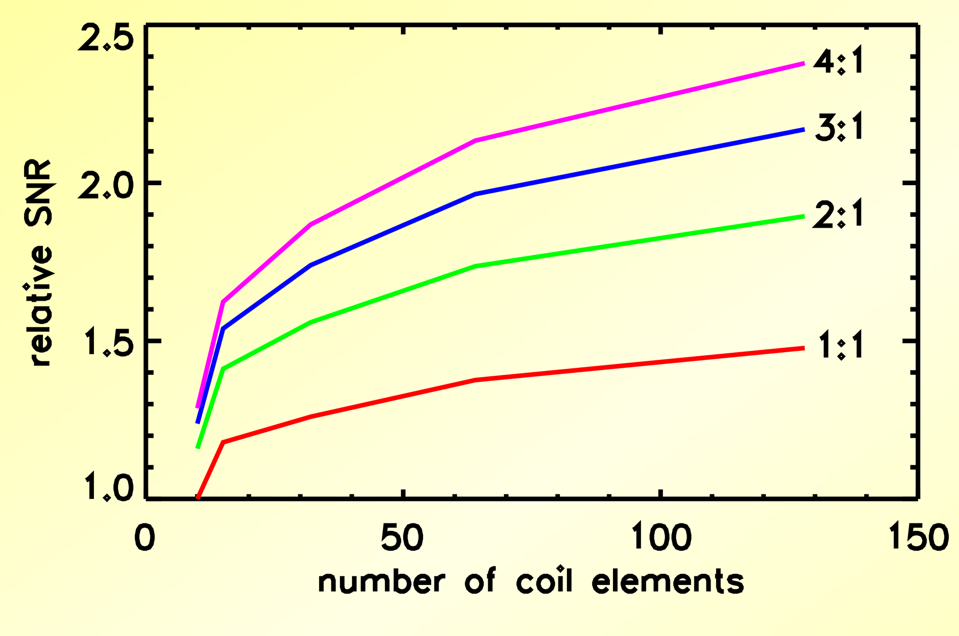

Figure 4: Electromagnetic simulations of estimated gapped-array SNR. Gain is strongly dependent on the contribution of noise from the electronics (NE). When NE for 10 coil elements (n) is less than ¼ of the sample noise (NS), gains level off beyond n=64 (top curve, ratio NS:NE for n=10 indicated). In more realistic cases of higher NE, leveling-off occurs at lower n, with little gain achieved beyond n=32 when NE=NS for n=10.

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*dezwart@nih.gov