Parallel Imaging Update: How Many Elements Do We Need?

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Introduction While advances in the “primary” technologies of MR such as increased field strength and improved gradient performance have been substantial, advances in the third component of the triad, RF receive technology, has proved to be an effective and likely cost-effective way to improve MR’s sensitivity and encoding capabilities. Initial experience with array coils quickly showed that signal detection efficiency improved with even a modest number of surface coils and that the reception uniformity and reconstruction burden could be handled with simple post-processing algorithms.(1) The use of surface coils was thus able to expand into the traditional role of volume coil structures. Today virtually all clinical coils (except the body excitation coil) are available as receive-only array coils. Even as the array coil principles were being worked out, it was shown (2-4) that multiple detector geometries could be used to aid MR encoding; the foundation for the widely used methods of SENSE and GRAPPA. The sensitivity and acceleration potential both drive exploration of the logical expansion of the array approach: increasing the number of elements and reducing the size of the receive coils (to cover a fixed anatomical territory).

With current clinical systems (e.g. 32 channel), the reduction of a 10 minute volume scan to <1minute with 12 fold acceleration is possible with minimal increase in noise beyond the standard tradeoff between SNR and acquisition time. In addition to shortening total exam times, the capability of order-of-magnitude accelerations allows multiple contrasts to be obtained within a clinically relevant scan time. With even more channels (64 and 96), credible imaging can take place with truly minimal use of the gradients (for example a single readout). Using a single readout encoding and MEG-like reconstruction, fMRI has been performed at 50 fps with a 90 channel head coil. With 1% or less of the spatial encoding coming from the gradients and reconstruction with either extreme versions of MR parallel reconstructions or MEG-based methods, there is potential for new types of ultra-fast MRI such as nearly silent MRI with whole head volumes coming at 200 fps.

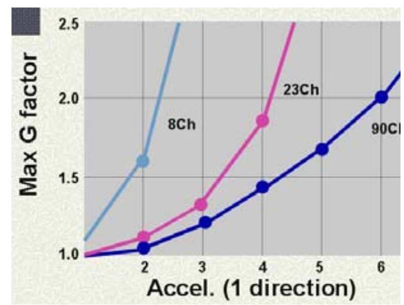

Fig. 1 Maximum g factor vs acceleration rate based on measured noise corr. and sensitivity maps for a 8Ch, 23ch, and 90 channel brain array.

Fig.2 90 channel 1.5T helmet array with preamps and the coil’s builder (Graham Wiggins).

Simulations Three related figures of merit for the array coil are of interest: a) the SNR as a function of position for un-accelerated scans, b) a similar SNR map for accelerated scans and c) the maximum acceleration achievable with an acceptable g-factor penalty. Simulation studies of increasing number of elements have concentrated on fixed geometries such as planar (5) or spherical arrays(6). The latter study also examined unconstrained current distributions; the so-called “ultimate SNR”. For the planar and spherical geometry the detection sensitivity is seen to increase with an increasing number of channels (n) with large benefits near the small detectors and asymptotically decreasing gains for deeper tissue. At a radius approximately 80% of that of the spherical coil model, the array showed an approximately linear increase in sensitivity as the number of coil elements was increased from 8 to 32. Importantly, there was still a factor of 7 to go before reaching the ultimate SNR limit at this location. As in the planar model, deeper positions showed SNR reaching an asymptote as the number of elements increased.

Experiment In order to explore the potential capabilities and practical limitations of large-n arrays (32 and beyond), multiple groups have initiated prototypes for the brain (23, 32, 90 and 96 channels), torso arrays (32ch), and heart (32ch and 128ch systems initiated). These preliminary efforts show promising potential for high sensitivity and acceleration. As the coil elements become smaller, maintaining the dominance of body noise over unloaded losses becomes more difficult to achieve, especially for 1.5T. A coil former that is tightly fitting to the body becomes critical and flexible and stretchable coil formers will be required to extend many of these designs to the clinical arena. Other important engineering concerns include producing a preamplifier and cabling system that is sufficiently small, transparent to the body coil excitation and robust enough that failures are minimized.

References

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