What can coils do?

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Introduction: Advances in field strength and improved gradient performance have been substantial and are the image most practitioners have of advancing MR technology. Nonetheless, advances in the third component of the triad, RF technology, have proved as valuable and perhaps more cost-effective for improved sensitivity and encoding capabilities in MR imaging. The coils which come standard on a state of the art scanner today look very different from those of 15 years ago. For example, the single channel volume coil was a standard receive coil for brain, extremity and even body imaging. Today, array coils of 8 to 32 channels perform the receive function and birdcage structures are found only as transmit coils (and even the single channel transmit coil appears to have a limited future thanks to parallel transmit array technology.)

The complexity (and associated cost) of highly parallel detection is easy to see with a glance inside the covers of a 32 or higher channel coil array (e.g. Fig. 1, a 64 channel head/neck/C-spine array). So what is the benefit? The answers are sensitivity and encoding ability and the ability to trade-off these two desirable goals in a flexible way. This talk will largely focus on what can currently be done in this arena and what types of applications might appear with the routine ability to accelerate image encoding by more than 10 fold.

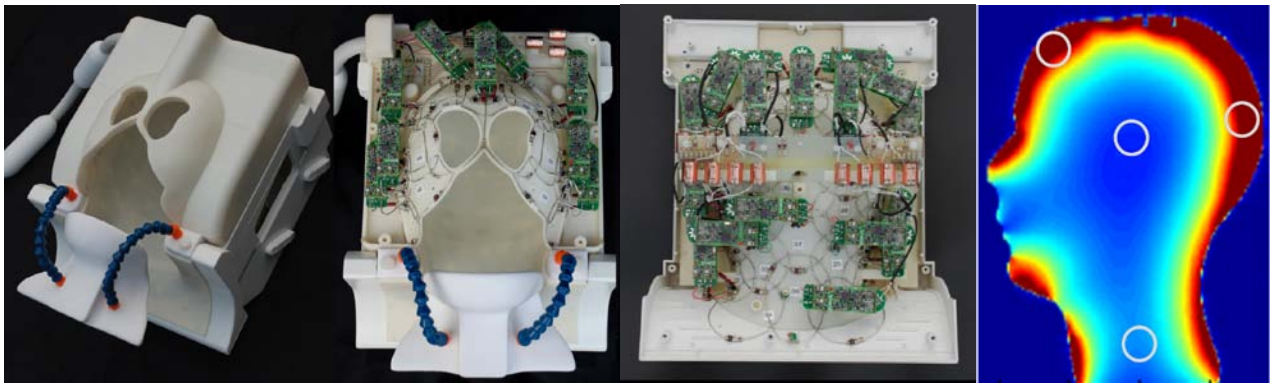

Fig. 1 The loop layout of a close-fitting 64 channel head/neck/C-spine array for 3T. Left) with covers. Right) top and bottom section without covers. Far Right) SNR map. The SNR was improved in the brain cortex by 2.4x compared to a larger 20ch array and by 1.2x compared to a similar sized 32channel array. In the C-spine region the SNR was increased 1.8x compared to the 20ch head/neck array. Courtesy Boris Keil, MGH.

Receive Sensitivity

Your coils today provide 2-5 fold increases over volume coils with the larger increases at the periphery of the body.

The ability to detect MR signals with Faraday detection (the generator effect) is ultimately limited for wire arrangements which must be placed external to the body; the so-called “ultimate SNR.” [1, 2] It seems that we are pretty close to the ultimate SNR in the center of the body [3], even with traditional volume coils (as long as they are tight fitting). The trend in coil construction has been to 1) squeeze more SNR out of the center by making the coil former more closely fit the body (this improves volume coil designs as well), and 2) gain SNR in the periphery by using an array of multiple small coils.

While the gains are not uniform across the body, the localized gains can be impressive. For example a 96 channel brain array achieved roughly 10x higher SNR in the superficial cortex

than a traditional volume coil.[4] An attempt to achieve this gain via bigger magnets would require a 30T magnet!

Parallel imaging acceleration

Coil arrays are not limited to detecting signal, they are also part of image encoding process (through SENSE, GRAPPA). Therefore the conventional SNR metric must be updated to include not only the SNR of un-accelerated images, but also the SNR of accelerated acquisitions.

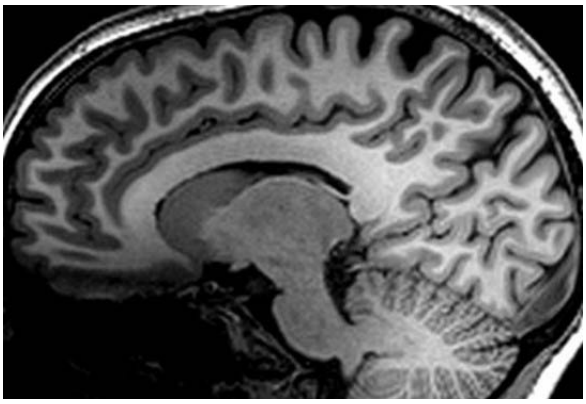

Fig. 2 3min 30second 1mm isotropic MPRAGE acquired with a 32ch pediatric coil sized for a 4 year old.

The ultimate limitation of the ability of the array for spatial encoding (e.g. through SENSE or GRAPPA) is likely imposed by the fundamental smoothness to the coil sensitivity patterns in regions free of current sources. Studies examining the ultimate acceleration limits point out a steep drop in the SNR for accelerations above about R= 4 or 5.[2, 5] For example, after this rate, the g-factor penalty was seen to rise exponentially with acceleration rate for locations near the center of the spherical sample [5] Both studies found that moving to higher B_0 field strength postponed the rapid deterioration of accelerated SNR. As with the unaccelerated SNR, the situation improves as you move closer to the array elements. Here the proximity to the wires provides more rapidly varying spatial profiles. This proximity effect allowed the 64 element array of McDougall and Wright [6] to achieve credible images with an acceleration rates of 64 fold; considerably higher than the 4 to 5 fold limit suggested by the ultimate sensitivity analysis which focused on regions far from the array.

What can your coils do for pediatric neuroimaging Pediatric imaging has the potential to greatly benefit from the high acceleration abilities of modern high field coils with 32ch or more. Figure 2 shows a 1mm isotropic brain volume acquired in 3.5 minutes. Both the resolution and imaging time are substantially better than volume coils and smaller channel count arrays. Figure 3 shows SNR maps comparing images acquired in 32 channel arrays sized for adults, 7 year olds and 4 year olds as well as the g-factors for these coils showing the benefit (and high acceleration and SNR possible) with appropriately sized pediatric arrays. The increased acceleration provides the means to reduce scan times considerably (e.g. down from 9 minutes to 2:15min for a 3D MPRAGE acquisition, allowing the possibility of pediatric imaging without sedation.

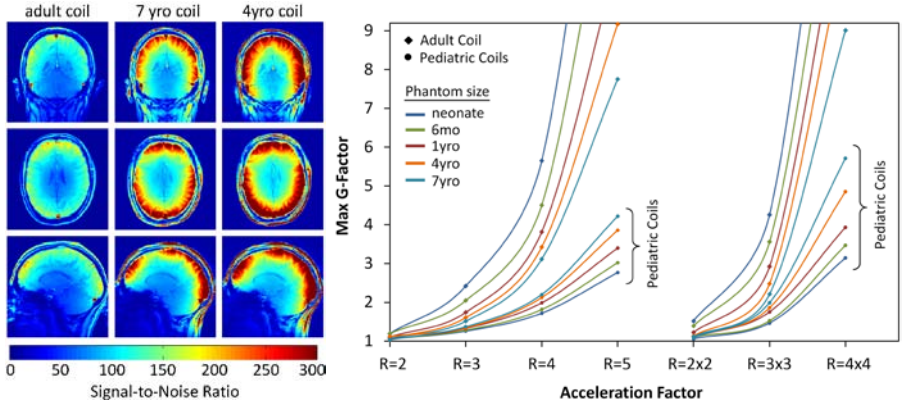

Fig. 3 *In vivo* SNR and g-factor comparisons between 32ch coils designed specifically for pediatric sized heads and a commercial available adult 32ch coil.

What can your coils do for Distortion Mitigation in EPI Accelerated array coil imaging has also emerged as the principal tool for fighting susceptibility induced distortion in EPI based images used for diffusion and fMRI. For example, the geometric distortions in the image (measured in mm) scale with the echo spacing (esp) of the EPI readout. In general esp increases with higher resolution readouts. Parallel imaging shortens the effective esp by a factor of the acceleration rate R, greatly reducing distortion. While a factor of 2x can be helpful, larger accelerations, requiring larger array coils, can enable a large reduction in distortion. For example, the single shot 1mm isotropic, TE=30ms, R=3, 3T EPI (Fig 4) shows anatomical-like resolution, which enables activation to be cross referenced with anatomical information from the EPI itself, solving registration problems. This degree of distortion mitigation is not likely to be matched by increased gradient performance (which is currently limited by nerve stimulation.)

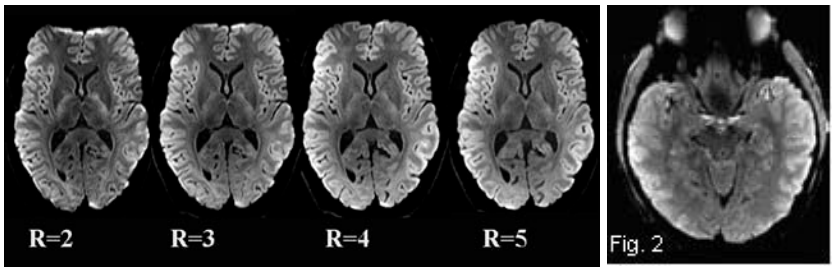

Fig 4. Left) High resol (1mm) 3T DWI, 32 ch array and GRAPPA R= 2, 3, 4, and 5. Note frontal distortion. Right) 1mm isotropic EPI, TE=30ms for fMRI at R=3. Note reduced dropout (from the 1mm slice) and low distortion in orbital frontal cortex.

What can your coils do for Diffusion Imaging The time- efficiency of diffusion imaging is relatively poor since a large fraction of the sequence is spent encoding the diffusion for that slice. Thus, lengthy TRs (e.g. TR<8s) are required for full brain coverage at high isotropic resolution (<2mm). In this situation, acquiring multiple slices simultaneously and teasing apart the aliased slices using parallel imaging can be very beneficial. For example, 3x slice accelerated simultaneous multi-slice will reduce a TR = 9s acquisition to TR=3s, retaining a fully relaxed image but acquiring the stack of slices and diffusion directions 3 fold faster. Unlike in-plane parallel imaging, the only penalty is the g-factor of the parallel imaging reconstruction (there is not a SQRT(R) penalty in simultaneous multi-slice since the data is not under-sampled. Recent work on multi-slice acquisitions for EPI based diffusion has resulted in a blipped-CAIPIRHINA scheme that speeds up diffusion acquisitions 3 fold (3 slices simultaneously), with a g-factor penalty on the order of 10%. [7, 8] Successful propagation of methods like this essentially allow high angular resolution diffusion to be acquired 3x faster with minimal penalty. Note that in order to unalias axial slices, coil elements must be arranged along the z direction, as well as the more conventional x and y directions.

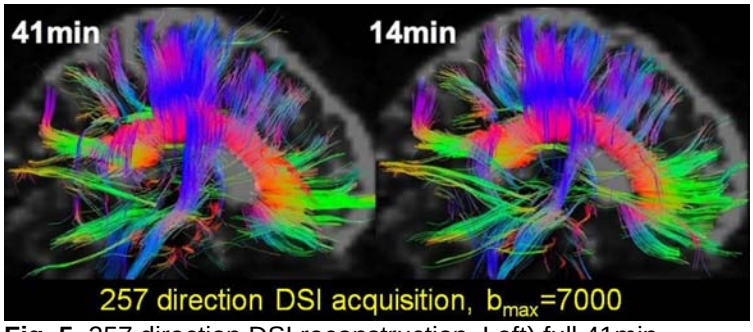

Fig. 5 257 direction DSI reconstruction, Left) full 41min conventional acquisition, Right) nearly identical 14min, R=3 simultaneous multi-slice acquisition; 3 fold faster with little SNR .degradation.

What can your coils do for Ultra-fast Imaging Single shot 2D encoding methods such as Echo-Planar Imaging (EPI) and spiral imaging are used for the vast majority of functional imaging studies to reduced motion artifact and physiologic fluctuations derived from the cardiac and respiratory activity. EPI can provide full brain coverage in less than 5 seconds, but is unfortunately not fast enough to Nyquist sample the cardiac cycle. The desirability of extending EPI to three spatial dimensions to speed up acquisition even further was realized in the first

description of EPI. [9] The resulting technique, termed Echo Volumar Imaging (EVI) was realized by Mansfield and colleagues a decade later. [10]

The main challenge of single-shot EVI is the difficulty of performing all of the gradient encoding within the range of T_2* of venous blood for BOLD imaging. Furthermore, the long echo spacing between samples in the "slow" phase encoded direction (along the second phase encoding direction, k_z) can cause severe image distortions due to susceptibility effects. For example, a whole 3mm isotropic resolution brain acquisition requires a 64x64x48 image matrix. Even with half-Fourier encoding, this still requires 1536 phase encode steps. With a typical echo spacing of T_{ES} = *0.5* ms for state-of-the-art body gradients, this would result in a readout duration of 768 ms, much longer than the T_2* relaxation time. Additionally, the traversal in the k_z direction would be 32 fold slower than in the k_y phase encode direction yielding susceptibility induced image distortions 32 times larger than EPI. In other words, EVI is only feasible with very high acceleration factors, 10 fold or greater.

An example of a single-shot EVI acquisition is shown in Fig. 6, acquired at 3T with 3.1 mm isotropic nominal spatial resolution and a matrix of 64x64x56. The acquisition used 6/8 Partial Fourier coverage and R = 4 x 4 acceleration to shorten the readout and speed up traversal in the slow encode direction 21 fold (and thus reduce susceptibility distortion 21 fold.), The EVI sequence allows whole-brain acquisition at 8.3 frames per second, easily allowing cardiac modulations to be fully time-sampled.

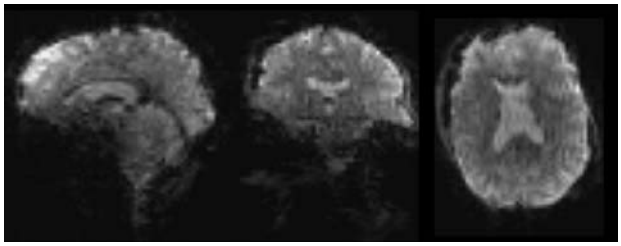

Fig. 6. Single shot Echo Volume Image of the head, matrix=64x64x56, taken in 120ms with TE=36ms. 21 fold acceleration from a 32ch 3T brain array is used. (work of T. Witzel, MGH)

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