128-Channel Body MRI with a Flexible High-Density Receiver-Coil Array

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Introduction: MR imaging with a large number of receiver channels enables higher acceleration factors for parallel imaging and improved SNR, provided losses from the coils and electronics are negligible. Channel count has been steadily increasing in recent years, with a number of groups reporting 32-channel results (1), and more recently 90-channels (2), and raising the question of the point at which benefits no longer accrue. Recent studies (3) have shown that body losses still dominate at 1.5 T for coil diameters down to around 5 cm. A hexagonal overlapped body array covering a 40 cm FOV with 5 cm diameter elements corresponds to roughly 128 channels. However, issues associated with packing receiver-coil electronics into a small region can compound coupling and losses, and affect flexibility and weight. In order to explore these limits, we have developed a 128-channel MRI system and body receiver-coil array. The array comprises two clamshells containing 64 coils each, with a fixed posterior array built to maximize SNR, and a relatively lightweight and flexible anterior array.

Methods: A 128-channel prototype MRI system was constructed that synchronizes the clocks on four 32-channel SIGNA HDx (GE Healthcare, Milwaukee, WI) transceiver systems, and gates data acquisition from the three auxiliary systems to that of the main system. 64-element anterior and posterior sections of the receiver-coil array were built, each section comprising a hexagonal lattice of 6.4 cm diameter coils covering a 40 cm x 47 cm surface, with coils overlapped to minimize coupling between nearest neighbors. The coils in the posterior array were water-jetted from 20-oz copper sheets and set in 3 overlapped layers of milled Lexan, and curved to conform to and embedded into the patient cradle. Coils were tuned and matched on a normal subject, and blocking networks and baluns were mounted on each coil (Fig. 1). Coils on the anterior array were etched from 10-oz Cu on 8 separate double-sided printed-circuit boards, each comprising a semi-rigid column of 8 coils (Fig. 2). Electronic components including baluns, capacitors, and blocking networks were all mounted on 8 separate printed-circuit-board spines aligned orthogonal to and soldered to each coil column. Rigid coax cables were run along the side of each spine to connector cables on the edge of the array. Each column of 8 coils was positioned with optimal overlaps relative to its neighbors and attached to neighboring columns via cloth hinges made of synthetic leather. These hinges allowed neighboring columns to tilt relative to one another, enabling the array to better conform to the subject. (Prototype experiments showed little change in coupling with moderate tilting.) Isolated coils had Q=370, and a loading factor of 4.0. SNR maps from a variety of scan planes were generated from a loading phantom for this array and for a 32-element array covering the same footprint (4) by acquiring 30 identical single-shot fast-spin-echo (SSFSE) images and dividing the average intensity of each pixel by the standard deviation of intensities. G-factor maps were generated for each array using 2D and 3D fast-gradient-echo data from the same phantom.

Results and Discussion: Figure 3A shows an axial SNR map of a loading phantom acquired with the array, and Fig. 3B shows a map of the ratio of SNRs between the 128-element and 32-element arrays. SNR is improved in the anterior and posterior surface regions by an average factor of 2, and in the central sphere by 25%, relative to the 32-element array. Average SNR in regions near the 64-element posterior section was noticeably better than near the anterior section, suggesting the flexible design has resulted in some SNR penalty. Maximum g-factors over the entire volume are given in Table 1, for two different 2D acceleration factors. Substantial improvements are seen for 128 channels.

We have successfully demonstrated image acquisition using a 128-channel MR system and a high-density 128-element receiver-coil array. Substantially improved performance was noted over a similar 32-element array. This system should help enable applications requiring extremely high speeds using previously unattainable accelerations.

Table 1: Maximum g factor vs acceleration

Array	R = 4x4	R = 5x5
32-element	6.7	26
128-element	2.1	3.2

References:
1) Zhu Y, et al. MRM 2004; 52:869-77. 2) Wiggins GC, et al, ISMRM 13th meeting, 2005, p. 671. 3) Boskamp E, et al, ISMRM 13th meeting, 2005, p. 916. 4) Hardy CJ, et al. MRM 2004; 52:878-84.

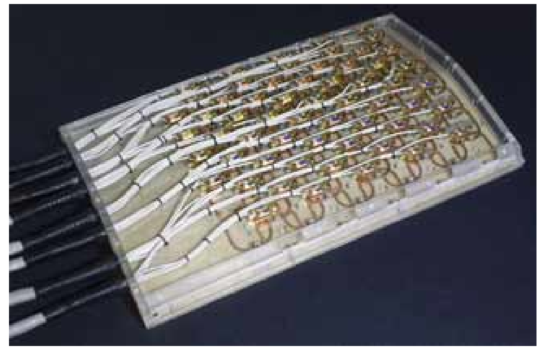

Figure 1. 64-element posterior rf receiver coil array (bottom view) is mounted in patient cradle.

Figure 2. Flexible 64-element anterior rf receiver coil array, with covers removed, on subject lying on posterior array.

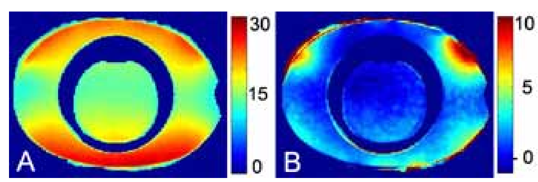

Figure 3. A) SNR map of a central axial slice from 128-element array. B) Map of relative SNR compared to a 32-element torso array, displayed on $10\log_{10}$ scale (0 on scale indicates relative SNR of 1).

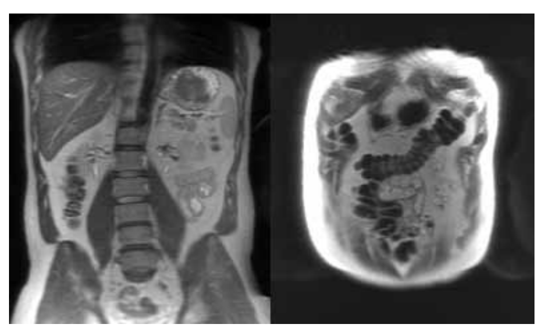

Figure 4. Coronal SSFSE images at different depths acquired with array.