64-Channel Array Coil for Single Echo Acquisition
Magnetic Resonance Imaging

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A 64-channel array coil for magnetic resonance imaging (MRI) has been designed and constructed. The coil was built to enable the testing of a new imaging method, single echo acquisition (SEA) MRI, in which an independent full image is acquired with every echo. This is accomplished by entirely eliminating phase encoding and instead using the spatial information obtained from an array of very narrow, long, parallel coils. The planar pair element design proved to be key in achieving well-localized field sensitivity patterns and isolated elements, the crucial requirements for performing SEA. The matching and tuning of the array elements were accomplished on the coil array printed circuit board using varactor diodes biased over the RF lines. The array was successfully used to obtain SEA images as well as conventional partially parallel images at unprecedented acceleration factors. Magn Reson Med 54: 386–392, 2005. © 2005 Wiley-Liss, Inc.

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For some time, the use of arrays of coils to partially replace phase encoding in MR images has offered assistance in the efforts to decrease scan time. As gradient-based methods to speed imaging operate at the threshold of hardware and safety limitations, the use of phased arrays and partially parallel imaging techniques that were introduced and discussed theoretically as early as the mid-1980s are now well-recognized for their clinical effectiveness (1–10). Using arrays of 4 to 8 coils, methods such as SMASH and SENSE have made acceleration factors of 2 to 4 now common clinically, and research sites routinely exceed this. The number of coils used in arrays has tracked (and pushed) the number of receivers available. Sixteen-channel head coil work was presented as early as 1998 (11,12), and 32-channel surface coil work has recently been presented (13). Our group has reported a 64-channel prototype portable receiver, enabling the use of 64-element arrays and potentially corresponding acceleration factors (14).

Our primary goal regarding 64-channel research was to explore MR imaging at the limit of acceleration factors: acquire an image in a single echo, with phase encoding completely eliminated, replaced by the spatial localization of long and very narrow planar coils. Using this imaging method (single echo acquisition or SEA), slice selection, and frequency encoding are performed using standard gradient methods, with the frequency encoding along the long axis of the array elements. No phase encoding is required; instead, the echo from each coil is 1D Fourier transformed and becomes one column of the final image (15). This requires a high degree of independence in the B1 patterns of the array elements, a condition that is met parallel to and relatively near the array (16).

This paper describes the first 64-channel array coil for MR imaging, constructed to enable SEA imaging at 4.7 T. The nature of the intended use for the coil dictated a clear but nonstandard set of design parameters where SNR, Q factor, penetration depth, and other characteristics, while all desirable, were not the highest priorities or defining features. While decoupling array coils is always a significant issue, in this case it was of particular consequence due to the fact that space and expense considerations led to the use of standard, 50-ohm chip preamplifiers (Agilent INA-01170 Low Noise MMIC Amplifier, noise figure 1.7 dB) instead of isolation preamplifiers (5,17). This consideration, combined with the large number of elements and the intention to use the array to completely perform the localization in the phase encoding direction, defined the crucial design characteristics: The complexity, size, and expense inherent to 64 channels needed to be minimized, and the elements needed to be long and narrow, parallel, closely spaced, and decoupled with highly localized field patterns. The array coil developed has been used to demonstrate successful single echo acquisition imaging as well as imaging at extremely high acceleration factors.

METHODS

Element Design

The two most reasonable designs for the array elements, considering the requirements, were a conventional loop and a planar pair (see Fig. 1) (18). The SNR of the two designs is generally considered comparable near the surface; hence, the use of planar pairs in quadrature pair coils (19). The nature of SEA ensured that imaging would occur almost exclusively near the surface; thus, the two designs were evaluated based on their element-to-element coupling characteristics and coil sensitivity pattern localization. Both were modeled using a full-wave method-of-moments software package developed in-house. The relative field patterns of the two coils are compared in Fig. 2a. The loop coil has a 30% broader pattern, measured at the point where the field falls to half intensity (full-width half-maximum, FWHM). For SEA imaging, where each coil is responsible for the strip of the final image directly above it, the broader pattern of the loop coil is undesirable.
Figure 2b compares the element-to-element coupling between two loops and two planar pairs as a function of separation distance. Coupling was quantified using the magnetic coupling coefficient as defined by Roemer et al. (5). The loop coils exhibited far stronger coupling than the planar pair elements, with a coupling coefficient more than four times greater at a distance corresponding to the “nearest neighbor” separation. Avoiding the complexity of 64 decoupling networks, as well as being mindful of the need for the narrowest possible field pattern, made the planar pair design the clear choice for the prototype SEA array construction.

Array Design

A 64-element array of planar pair coils, shown in Fig. 3, was constructed for imaging on our 4.7 T/33 cm Bruker/GE Omega system. Because the receive-only array had to fit inside the homogenous region of a volume coil inside an 18-cm-id Accustar S-180 gradient coil, the portion of the array to be used for imaging was designed to have an overall dimension of $13 \times 8.1$ cm. Each element consisted of three traces, each 10 mil (0.254 mm) wide with 20-mil (0.508-mm) gaps between them. Ten-mil spacing between each coil yielded total element footprints of only 2 mm (80 mil) $\times$ 8.1 cm. To realize as much space as possible for matching and tuning the 2-mm coils, the array fanned out to 16 cm, completely filling the volume coil. This transition from the array elements as well as the matching and tuning networks were fabricated over a ground plane in an attempt to keep the active area of the array limited to the linear portion of the elements. Two single-sided varactors (one for each loop of the planar pair) in a nonmagnetic SOD-323 package (Infineon BB639) were used for tuning on each coil. The decision was based on several factors: first, the coil needed to be tuned in a balanced configuration across the paired loops; second, the tuning mechanism needed to be very small; third, it was desired to tune, at least approximately, all 64 simultaneously in “bulk.” The varactors were biased over the RF lines by placing a 10-kΩ resistor under the matching capacitor, as there was not enough space to put it in parallel. This provided a simple method of providing DC bias to the varactors with minimal additional components. Four 1.5-m ultrasound cables, each containing twenty 50-Ω coaxial lines (Precision Interconnect “Blue Ribbon,” Wilsonville, OR, USA), were used to connect the coil to two 32-channel “bias insertion boards” located outside the bore. The cables were preassembled with low-profile header connectors, and matching surface-mount receptacles (Samtec QSE/QTE series) were installed on the array and bias insertion.
boards. The bias insertion boards contained the DC block and conversion to the preamp lines and also connected to high-resistance carbon wire lines leading to the potentiometer boards that were used for individually controlling the bias voltages. The layout of the entire system is diagrammed in Fig. 4. The boards were all mechanically etched in the Magnetic Resonance Systems Lab using a C30 PC board prototyper (LPKF, Wilsonville, OR, USA) and the final coil was outsourced for etching (PCB Express, Mulino, OR, USA).

In order to assess the effect of the ultrasound cables and varactors on the SNR, a “test array” was fabricated with planar pair coils in three different configurations. The first set of coils was tuned with capacitors instead of varactors and connected to the conversion board by standard RG-174 coaxial cable instead of the ultrasound cable. The second configuration was intended to test the effect of the varactors and was thus connected to the conversion board by standard coax but was tuned with varactors instead of capacitors. The third set was designed in the final 64-channel array scheme—tuned with varactors and connection to the conversion board with the ultrasound cable—in order to have a comparison of the overall system to the “ideal” as well assess the effect of the cabling. Q measurements were made on the bench using the VSWR 2:1 bandwidth of the coils in the three different configurations (20), and the relative SNR was calculated as $\sqrt{\frac{Q}{\text{number of channels}}}$.

The system as designed and operated, using varactors and the blue ribbon cable, showed a 41% decrease in SNR from the ideal capacitively tuned-coax-connected case, with the ultrasound cable responsible for a 26% decrease and the varactors responsible for the remaining 15%.

**Coupling**

For coupling measurements, the coils were impedance matched to 50 ohm and $S_{21}$ measured on an HP4195 Net-
work Analyzer. The calibration and measurement point was the input of the system, the bias-insertion board. In order to verify that the $S_{21}$ measurement was not significantly affected by cable coupling or coupling on the insertion board, several $S_{21}$ measurements were also made after calibrating though the entire system to the input of the array using calibration standards especially fabricated for the Samtec connectors. Measurements at the two calibration points, labeled in Fig. 4, indicated a variation of less than 1.5 dB. Nearest neighbor $S_{21}$ values on the 4.7-T array were $16.8 \text{ dB}$ and fell to better than $20 \text{ dB}$ by third neighbor. The coils tuned easily and independently; therefore, these values were deemed adequate for initial evaluation of SEA imaging.

Imaging and Reconstruction
Both fully encoded and SEA images were obtained on our 4.7 T/33 cm Bruker/GE Omega system. The prototype 64-channel receiver system previously reported was used to acquire signals from all coils simultaneously. For purposes of testing and evaluation at different acceleration factors, 64 fully encoded images were obtained using a standard, spin-echo pulse sequence with resolution $256 \times 256$, TR 250 msec, TE 13 msec, 1 average, spectral width 50 kHz, FOV 14 cm. A phantom was constructed from a 13-cm-diameter dish containing spiraled compartments filled with various objects and alternatingly filled with distilled water, 1 g/liter copper sulfate, and 0.5 g/liter copper sulfate. Imaging was performed in the coronal plane, parallel to the plane of the array, slice thickness of 2 mm, centered 4 mm above the array. For single echo acquisition imaging, SEA, a gradient echo sequence was modified to eliminate the phase encoding table and acquire a $64 \times 256$ [(number of coils) $\times$ (readout resolution)] image with every excitation. The signal from each coil was mixed to an intermediate frequency of 0.5 MHz where it was digitized with 16-bit resolution at a 2.5-MHz sample rate. All 64 channels were digitized simultaneously. Digital I/Q demodulation of the sampled signals was performed using in-house software following all data acquisition.

To evaluate the performance of the array at high acceleration factors, appropriate lines in the full k-space data set were set to zero. Accelerated images were reconstructed using a method similar to PILS (21); the 64 full data sets were decimated, Fourier transformed, and masked according to the coil profile to eliminate aliasing, and a sum-of-squares reconstruction was implemented. For SEA imaging reconstruction, a 1D FFT was performed on the echo received from each coil, the resulting 64 images were stacked into a $64 \times 256$ matrix, and the matrix was interpolated to $256 \times 256$ for display.

RESULTS AND DISCUSSION
While the coil array was built to enable SEA imaging, it was evaluated for its performance in high acceleration factor imaging as well. As described above, beginning from a full $256 \times 256$ data set, imaging at acceleration factors from 2 to 64 were simulated by using only 128, 64, 32, 16, 8, and 4 k-space lines. Images were also constructed with only 2 and a single k-space line (SEA imaging). While the imaging time has been accelerated by factors of 128 or 256 in these images, it seems appropriate and necessary to distinguish acceleration due to coils and acceleration due to k-space manipulation. We therefore refer to coil acceleration as $L_c$, strictly k-space acceleration as $L_k$, and total acceleration in the standard notation $L$. While both result in SNR loss according to the square root of the acceleration factor, each has a different effect on resolution and it is therefore necessary to distinguish between the two. The results of the accelerated imaging are shown in Fig. 5. While the SNR follows the $\sqrt{L}$ loss expected, the resolution does not begin to significantly degrade until we use only 2 lines of k-space for reconstruction. This is due to

![FIG. 5. Accelerated images made using the 64-channel array coil. Resolution is maintained through an acceleration factor of 64, the maximum coil-based acceleration ($L_c$) possible from a 64-element array using conventional reconstruction techniques. Reductions in imaging time by factors greater than the number of coils can be obtained by further reducing the amount of k-space data obtained, but as this information is not replaced by coil pattern information, resolution degrades accordingly. This is evident in the lower right image, where a total reduction of 128 is obtained from a factor of 64 coil acceleration ($L_c = 64$) and a factor of 2 k-space reduction ($L_k = 2$).](image-url)
the fact that there is an overlap in the coil patterns and it requires at least 4 lines of $k$-space (corresponding to $L_c/H$) to effectively separate these spatially—i.e., to avoid aliasing into the masked region of another coil. Once the acceleration factor exceeds the number of coils, as is the case when 2 lines of $k$-space are used for reconstruction (corresponding to $L_c = 64, L_k = 2$), then the information lost in the acceleration is unrecoverable and the resolution suffers notably. The SEA image, reconstructed from a single echo, is shown in Fig. 6. As would be expected, the resolution in the vertical direction, determined by the coil patterns, is lower in the SEA image, as evidenced by the blurring of some features such as the rightmost point of the star. In addition, the effects of individual coil patterns, seen as horizontal lines in the sum-of-squares image ($L = 1$), are blurred at higher accelerations and the SEA image.

It is worth noting that the intuitive loss of resolution expected in a $64 \times 256$ image and seen in the blurring of smaller features is of a (visually and physically) different type than the aliasing-based loss of resolution seen in the PILS highly accelerated images. The width of the coil pattern is responsible for both, but because of the differences in the reconstruction methods, the overlap manifests itself as two different violations. Specifically, in SEA imaging, the point spread function is determined by the individual coil pattern. This is illustrated in Fig. 7 by a fully encoded image and profile from the set of 64 used to form the accelerated images in the previous figure. The coil sensitivity falls to half intensity by the center of the adjacent elements. Thus, the bulk of the signal in the echo is received from the region directly over the coil, as is needed for SEA imaging. Features that are smaller than the width of the coil pattern shown will suffer loss of contrast, similar to partial volume averaging in the slice encode direction, as well as effective blurring to the physical coil width (the nominal resolution in the original phase encoding direction). These effects increase with distance from the array, likely limiting the effectiveness of the coil for SEA imaging to relatively shallow penetration depths, where all of our work has occurred. To illustrate these effects, however, several profiles at increasing depths are shown in Fig. 8 at a smaller field of view than the previous figure. Immediately apparent is the decrease in SNR with depth. In addition, while not as obvious as if the profiles were normalized, the FWHM of the patterns (and thus the point spread function) broadens considerably with depth.

It is commonly accepted that the most useful region of sensitivity for a coil exists within approximately 1 coil width above it, consistent with the results in Fig. 8, which shows that the SNR has dropped by 80% at a depth of 2.25 coil widths. The penetration depth, then, of coils small enough to effectively replace a pixel in the phase encoding direction (and maintain a standard degree of resolution) will understandably not be more than a few millimeters at best. In this case, however, any initial uncertainty regarding the clinical utility of the array or the method was outweighed by the opportunity to demonstrate and explore a new imaging technique, one that offered the possibility of imaging at rates approaching 1000 frames per second using recalled echoes. The power of the SEA method lies in the fact that, in contrast to parallel imaging
techniques where image acquisition time is temporally blurred over the time required to acquire multiple lines of $k$-space, this technique acquires essentially a "snapshot"—a full image formed in as little as the few hundred microseconds of a single acquisition. Snapshot imaging by eliminating phase encoding has been proposed before (3,6), but the coil described here has enabled the first successful implementation of this concept. This ability is new to the field of MRI and therefore by its very nature will offer the benefits of the imaging modality to new, unexplored, and even unknown (at least to the authors) applications. Broadly speaking, with the sequencing of the human genome came a redefinition of the direction of medical research toward the molecular level (22). In order to continue to play an integral part in this movement, MRI is evolving to meet smaller and faster requirements. SEA (and the associated hardware described in this paper) represents a movement of MRI, we hope, in that direction. The field of microfluidics, for example, with its eventual lab-on-a-chip goal, is already actively adapting MRI to meet its needs with coil miniaturization (23,24) and even adapting itself to meet MRI's capabilities. For example, microfluidic devices are being scaled to sizes suitable for MR imaging (25) and fabricated with MR compatible materials (26). MRI is particularly attractive to this field as it overcomes two main disadvantages of the optical techniques that are most commonly used for fluidic imaging: the need for exogenous contrast agents that disturb the molecular environment and the ability to image only through optically transparent materials (27,28). When imaging flow channels for microfluidic techniques, the penetration depth of interest is on the order of millimeters or less, and the nonperiodic chemical kinetics and flow dynamics (where gating is not possible) can require kilohertz frame rates or greater to observe (29–31).

The coil could also easily prove useful to more traditional MR surface imaging applications. While this coil was specifically designed to enable the SEA method, it stands alone as a tool to provide extremely high acceleration factors in surface MRI using partially parallel imaging techniques. This is of particular benefit as surface imaging is traditionally very slow since it is typically used in conjunction with microscopy and 3D techniques to gain not only in-plane but also thru-plane resolution of fine structures such as the skin and ex vivo tissue samples (32–35). High acceleration factors using a very large number of array elements would enable a large field of view while maintaining high resolution, important in these applications. On the other hand, the penetration depth of the small coils is undeniably a limiting factor, making it more appropriate to use 2D arrays of larger coils in cases where depth and high accelerations are required (36). SMASH and SENSE or other partially parallel imaging methods might also be used to reconstruct the less localized data further from the coils (37). This was not investigated in this study, as the coil was designed to provide the necessary localization as long as the imaging plane was close to the array. While the use of more sophisticated image reconstruction techniques may provide some degree of resolution enhancement and is certainly worth investigating, the reconstruction overhead may be formidable with a factor of 64 acceleration. Additional gains in the imaging depth could potentially be obtained by investigating designs such as the planar strip array (38) or pressing on the feasibility of using loops.

Finally, there are a number of practical issues that arise in the implementation of very large arrays. There comes a point as the number of elements in an array is increased where any SNR benefit is forfeited due to the increasing copper resistance. In this case, it was the acceleration technique that was under examination with less regard to SNR, but in general, implementing large numbers of coils will be influenced by this confounder. Typical methods to lower losses, including the use of cooled copper or high temperature superconducting materials could be investigated (39–41). Cabling is certainly an issue, and lower loss cabling would be worth investigating in future work in highly accelerated imaging. An obvious alternative may be the development of wireless links to eliminate the complexity and losses associated with large numbers of cables.

CONCLUSIONS

This paper has presented the first 64-channel array for MR imaging. The planar pair elements, with their inherently low coupling and constrained field sensitivity patterns, allowed us to use inexpensive 50-Ω preamplifiers without complex decoupling networks at the coils. The use of ultrasound cable bundles and varactors further streamlined what could have been a very complex system. The array was used to make images with true acceleration factors of 64 with only moderate loss in image quality. Finally, the localized field patterns of the planar pair elements allowed for an image to be formed from a single echo acquisition, without any phase encoding, by simply stacking the image strips from above each coil. The straightforward nature of the design of the large array demonstrated, coupled with the successful implementa-
tion of the new imaging technique that it enabled, portends new possibilities in MR imaging.

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