Scalable Multichannel MRI Data Acquisition System

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A scalable multichannel digital MRI receiver system was designed to achieve high bandwidth echo-planar imaging (EPI) acquisitions for applications such as BOLD-fMRI. The modular system design allows for easy extension to an arbitrary number of channels. A 16-channel receiver was developed and integrated with a General Electric (GE) Signa 3T VH/3 clinical scanner. Receiver performance was evaluated on phantoms and human volunteers using a custom-built 16-element receive-only brain surface coil array. At an output bandwidth of 1 MHz, a 100% acquisition duty cycle was achieved. Overall system noise figure and dynamic range were better than 0.85 dB and 84 dB, respectively. During repetitive EPI scanning on phantoms, the relative temporal standard deviation of the image intensity time-course was below 0.2%. As compared to the product birdcage head coil, 16-channel reception with the custom array yielded a nearly 6-fold SNR gain in the cerebral cortex and a 1.8-fold SNR gain in the center of the brain. The excellent system stability combined with the increased sensitivity and SENSE capabilities of 16-channel coils are expected to significantly benefit and enhance fMRI applications. Magn Reson Med 51:165–171, 2004. Published 2003 Wiley-Liss, Inc.†

Key words: MRI receiver; array coil; parallel imaging; SENSE; fMRI

Single RF surface receive coils allow a high sensitivity in a small region of the object under study, while their sensitivity to other areas is often inferior to that of a volume coil. Arrays of surface coils allow collection of signals from several object regions in parallel, and extend the area from which high sensitivity measurements can be obtained (1–3). With appropriately designed array coils, the signal-to-noise ratio (SNR) of volume coils can be matched or improved upon in any location in the object (3–5). Furthermore, surface coil arrays allow the use of parallel imaging techniques such as simultaneous acquisition of spatial harmonics (SMASH) and sensitivity encoding (SENSE) (6,7), leading to additional benefits such as improvement in image resolution and reduction of image artifacts (8–12). One of the limiting factors in practical application of array coils has been the small number of receiver channels of commercial MRI systems. Depending on the MRI instrument brand, only one to eight channels are available for high-bandwidth, independent signal reception. For most applications this number is too low to fully exploit the potential of coil arrays. In particular, at high magnetic fields substantial improvement in array performance is expected as the number of channels is increased beyond eight, even for the study of relatively small objects such as the human brain (4,5,13). These considerations invite a redesign of the MRI receive system. In the following, a design that provides a flexible choice in the number of receive channels is presented and initial results obtained with a 16-channel whole-head receive-only SENSE array coil at 3 T are shown.

MATERIALS AND METHODS

System Design

In addition to scalability of the number of receive channels, the design considerations included high bandwidth, high duty-cycle signal reception, and a cost of around $10,000 per channel. A 100% acquisition duty cycle was targeted at a 1 MHz output bandwidth in order to allow BOLD fMRI with high scan rates. Furthermore, the scalable MRI receiver was designed to perform the entire MRI acquisition process, including signal reception, digitization, data storage/transmission, and image reconstruction and display. The system was intended to be used with a 3 T GE (Milwaukee, WI) Signa VH/3 MRI scanner (3 T/90 cm, gradient strength 40 mT/m, slew rate 150 T/m/s, whole body detunable transmit/receive (T/R) RF coil). A schematic overview of the design of the 16-channel receiver is shown in Fig. 1. The system consisted of three major components that will be discussed below: analog electronics, digital electronics, and control software.

Analog Electronics

The analog electronics of the custom receiver included a 16-channel MRI receive coil, PIN-diode drivers for actively detuning the receive coils during activity of the RF transmission, preamplifiers (preamps), second gain stages, signal combiners, a third gain stage with computer-controlled attenuators, and anti-aliasing filters. The MRI receive coil, shown in Fig. 2, was based on a gapped-element design (5) for optimal sensitivity and SENSE performance. All analog components and the 16-element receive coil were built by Nova Medical (Wakefield, MA, USA). On a close-fitting asymmetric head coil former (5), 16 curved trapezoidal gapped elements were laid out in a single row using 6.25 mm wide, 50 μm thick copper tape. Each element was roughly 20 mm wide and gapped from next neighbors by 7 mm, resulting in an element width to-gap ratio of approximately 3 to 1. Each element was tuned to 127.8 MHz with multiple distributed capacitors and matched to 50 Ω coaxial cable with a lumped element bridge balun circuit. For patient safety and protection of the preamplifier cir-

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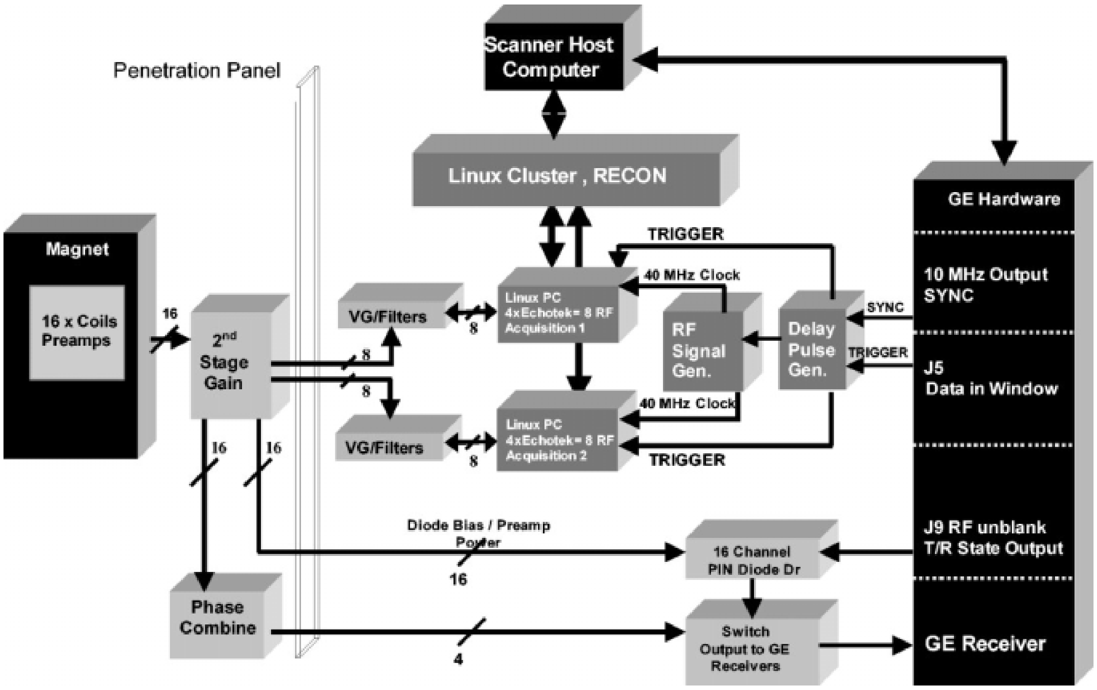

FIG. 1. An overview of the scalable 16-channel MRI data acquisition system. VG/Filters = variable gain stage (computer controlled) and filter bank; SYNC = synchronization signal from 10 MHz scanner master clock output; TRIGGER = scanner-generated TTL pulse, marking the time point for the start of data acquisition.

cuitry, pickup of RF transmit energy by the receive coils was minimized by PIN diodes (placed across the cable side of the bridge balun), which detuned each coil element during transmit. For additional patient safety, passive detuning traps were also included on each element and all cables had multiple cable traps to block common mode cable currents possibly induced during RF transmission. The low impedance preamplifiers, located in the magnet bore, provided less than 1 Ohm effective input impedance and 35 dB of gain at 127.8 MHz. These preamplifiers, in combination with capacitive isolating circuits between adjacent elements, provided greater than 20 dB isolation between all elements of the 16-channel coil (5,13). Outside the magnet bore, the MRI signal from the preamplifiers was amplified 13 dB by a second gain stage, after which the signal of each channel was split into two branches to allow simultaneous digitization by both the 4-channel GE receiver and the custom-built 16-channel receiver. The analog signals from adjacent groups of four coils were combined at fixed phase. The resulting four groups of combined signal from each quadrant of the 16-channel coil were fed into the four available GE receivers. These combined signals were used as GE scanner input to perform automated scan setting adjustments, such as RF flip angle optimization, shimming, and center frequency adjustment. The 16 signals going to the custom-built receiver were bandpass-filtered and additionally amplified with computer-controlled amplifications. The bandpass filter width was approximately 8 MHz and overall system gain was adjustable between 52 to 68 dB.

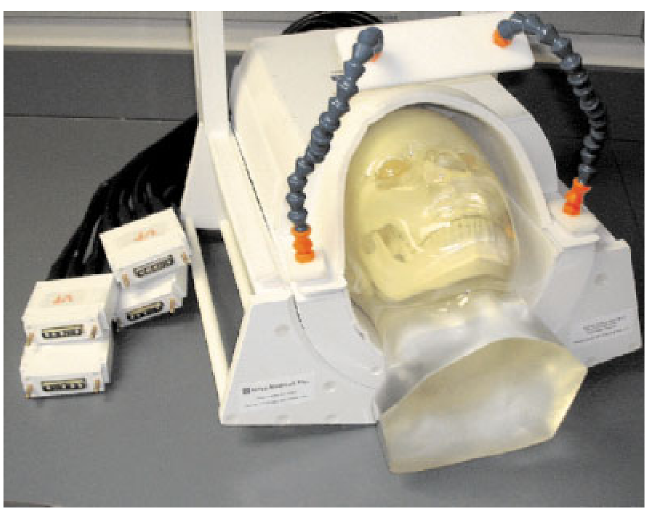

FIG. 2. The 16-element, gapped, SENSE-optimized, whole-brain surface array coil with the human head phantom inside.

Digital Electronics

The digital electronics included two acquisition computers and an image reconstruction Linux cluster. The reconstruction cluster included a master node plus 12 slave nodes, all rack-mounted dual Intel Xenon 2.2 GHz CPU personal computer (PC) workstations. Two rack-mounted, single CPU (AMD Athlon MP 1900+), PC-based Linux workstations served as acquisition computers. Each acquisition computer contained four wide-band digital-receiver PCI boards. Both acquisition computers and the Linux reconstruction cluster were connected through 1 Gbit/s Ethernet. Network connectivity with the scanner host

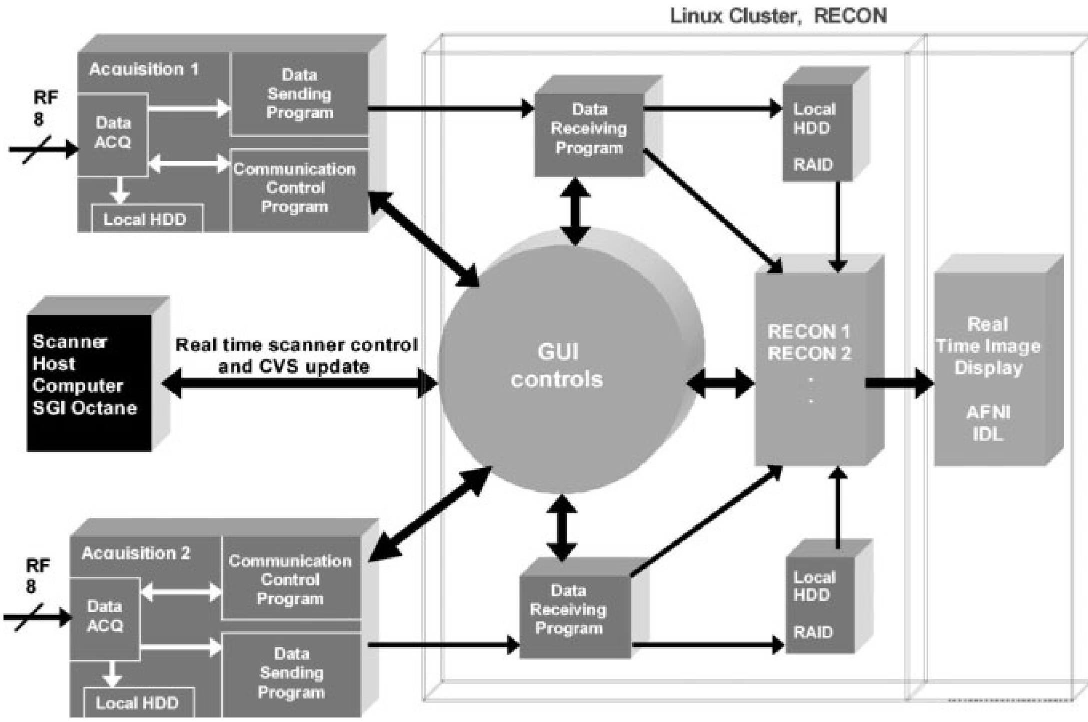

FIG. 3. Architecture of the receiver software, including data flow (thin arrows) and control commands (bold arrows). GUI = graphical user interface master control program; ACQ = Acquisition (on Echotek boards); HDD = hard disk drive(s) for raw data storage; RAID = redundant array of independent disks for raw data and image storage; RECON1, RECON2 = reconstruction algorithms/programs for different pulse sequences; AFNI = software for image analysis, manipulation, and visualization (16); IDL = The Interactive Data Language software; CVS = MRI scanner control variables.

computer (SGI Octane, Silicon Graphics, Mountain View, CA) was achieved via a standard 10/100 Mbit/s Ethernet connection.

The receiver was designed around a PCI-based digital receiver board (ECDR-GC214-PCI/TS, Echotek, Huntsville, AL), each with two analog input channels. The ECDR-GC214-PCI/TS has two wideband intermediate frequency (IF) analog inputs (+4 dBm, IF range 300kHz–200MHz). The board digitizes the analog inputs with two 14 bit analog-to-digital (A/D) converters (AD 6644AST, Analog Devices, Norwood, MA, A/D maximum conversion rate 70 MHz), and then down-converts and filters the digitized signals (Graychip GC4016 Quad Narrowband Digital Receiver, Texas Instruments). After down-conversion, the data are placed into a first-in-first-out (FIFO) memory buffer (128 kB/channel). The data could be read from the FIFO over PCI bus (through the on-board PCI interface chip, PLX PCI19080) or transferred by direct memory access (DMA) to PCI memory using one of the on-board DMA controllers.

The ECDR-GC214-PCI/TS board was used in "two-channel wide band" mode. This mode combines pairs of two of the four GC4016 digital down-conversion channels to make two independent wideband digital receiver channels (data alignment: 16 bit complex, split I/Q mode) (14,15). In split I/Q mode two GC4016 digital receiver channels work together to double the maximum output bandwidth of the downconverter. In this mode the real part of the 16 bit complex output data is processed in one channel and the imaginary part in the other. Programmable decimation ratio for this mode was from 16 to 16,384. An A/D conversion rate of 40 MHz was chosen (clock signal supplied by SML01 RF Signal Generator, Rhode & Schwartz, München, Germany), which resulted in available acquisition bandwidths (i.e., conversion rate divided by decimation ratio) ranging from 2.44 kHz to 2.5 MHz. In order to center the alias of the NMR signal (127.8 MHz) at DC, the tune frequency for digital down-conversion was set to 7.8 MHz ((127.8-6*(40/2)) MHz). Receiver operation was tested at acquisition bandwidths of 125 kHz, 250 kHz, 500 kHz, and 1 MHz (each with 80% pass-band width), corresponding to 8 μs, 4 μs, 2 μs, and 1 μs sampling times (dwell times), respectively.

For 16-channel acquisition, eight Echotek boards were started synchronously by a TTL pulse derived from the J5 ("Data in Window") output on the GE receiver chassis. Shape and timing of the TTL pulse was modified by a delayed pulse generator (DG 535, Stanford Research Systems, Sunnyvale, CA) to provide START and STOP signals for all receiver boards. To synchronize the DG 535 and the clock RF signal generator with the MRI scanner, the 10 MHz master clock from the scanner was used.

Control Software

A schematic overview of the control software for the receiver is shown in Fig. 3, which includes the major functions and data flows. To control the Echotek digital receiver boards a device driver (loadable kernel module for Linux kernel series 2.4.xx) was developed based on the

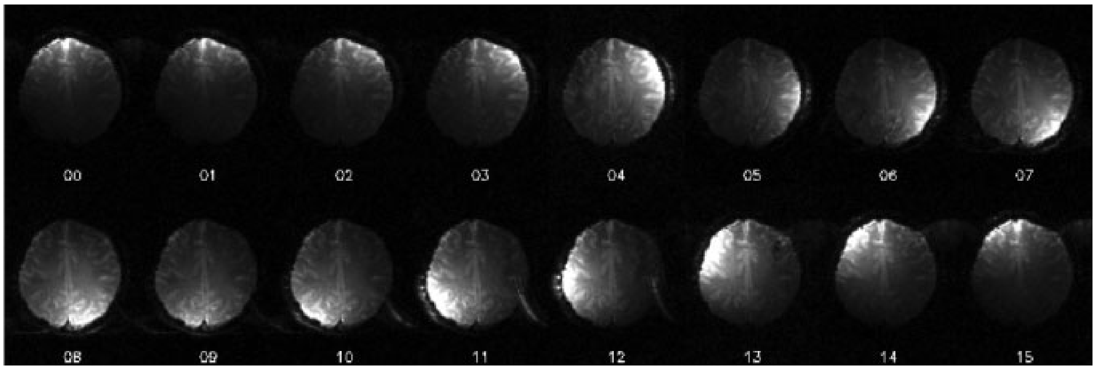

FIG. 4. Human brain images obtained with each of the 16 individual coil elements (single-shot full k-space gradient echo EPI, single axial slice).

software provided by the manufacturer. In addition, acquisition, communication, and data transmission programs were developed. All hardware control, data acquisition, and data handling programs were written in C. To control the entire system (both acquisition, reconstruction cluster, and scanner host computers) and provide real-time MRI scanner control, a master graphical user interface (GUI) was written in Perl with Tcl/Tk module extension.

The GUI master interface controlled, initialized, and communicated with both acquisition computers and the scanner host computer. The GUI also controlled the variable gain stage, data reception, data reconstruction, and the real-time display software running on the reconstruction cluster. Basic GUI operation is as follows. After scan prescription on the GE console, the GUI extracts the scan parameters from the scanner host computer. Based on this and optional input from the user, it launches and initializes the data acquisition and data transmission programs on both acquisition computers, as well as data receive and the image reconstruction software on the reconstruction cluster. After receiving "READY" signals from both acquisition computers it forces the scanner to download the current scan prescription to the GE acquisition computer, and starts the scan process on the GE acquisition computer. Raw data acquired with the custom receiver during scanning are incrementally saved on local hard drives on the two acquisition computers and simultaneously transferred to the Linux reconstruction cluster for storage and reconstruction. Data transfer between acquisition computers and Linux reconstruction cluster over 1Gbit/s Ethernet line was implemented using socket transmission. During the MRI scanning, the GUI displays data acquisition/sending progress in windows, sends information to the reconstruction program about incoming data, and allows the user to cancel scanning at any time.

System Evaluation

The system performance was evaluated by measuring the noise figure for the preamplifiers and measuring the overall dynamic range of the receiver, including the ADCs and digital filtering steps. The dynamic range was measured with a two-tone experiment (tone signal source: RF signal generator Rhode & Schwarz SL01), to check for range of system linearity as well as for possible intermodulation products. Imaging performance was evaluated on phantoms and human subjects. The human studies were performed after obtaining informed consent and were approved by the NINDS Internal Review Board (protocol 00-N-0082). All studies were performed using a gradient-echo single shot (rate-1, 2) SENSE echo-planar sequence

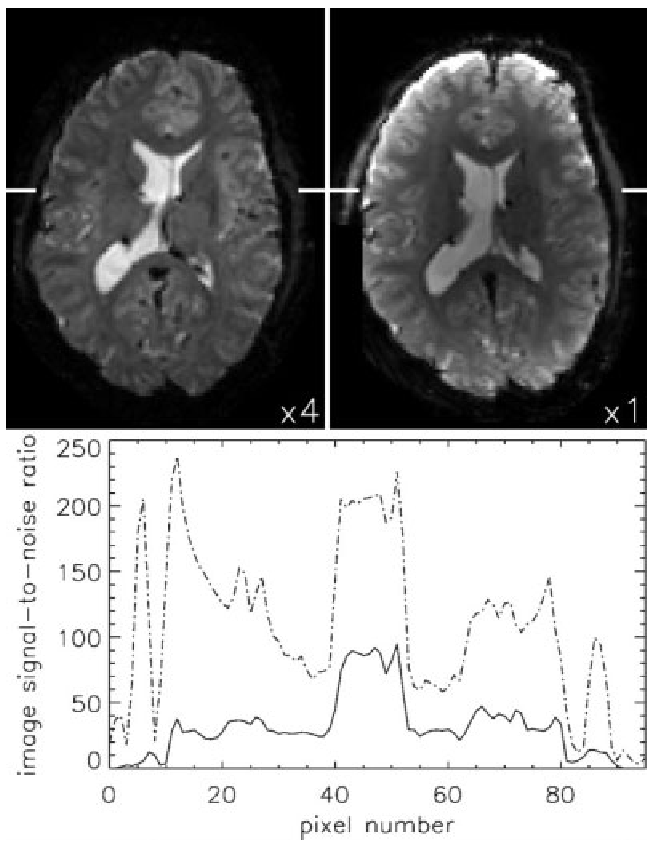

FIG. 5. The sensitivity-weighted combined image from the array coil (right) in comparison with the 28 cm I.D. GE transmit/receive birdcage head coil (left). Note 4 to 1 ratio in display level (windowing) for both images. Plot below: SNR vertical projection profiles from both images at the same location (vertical projection marked on both images with white tick marks). Solid line the birdcage coil, dashed line the array coil.

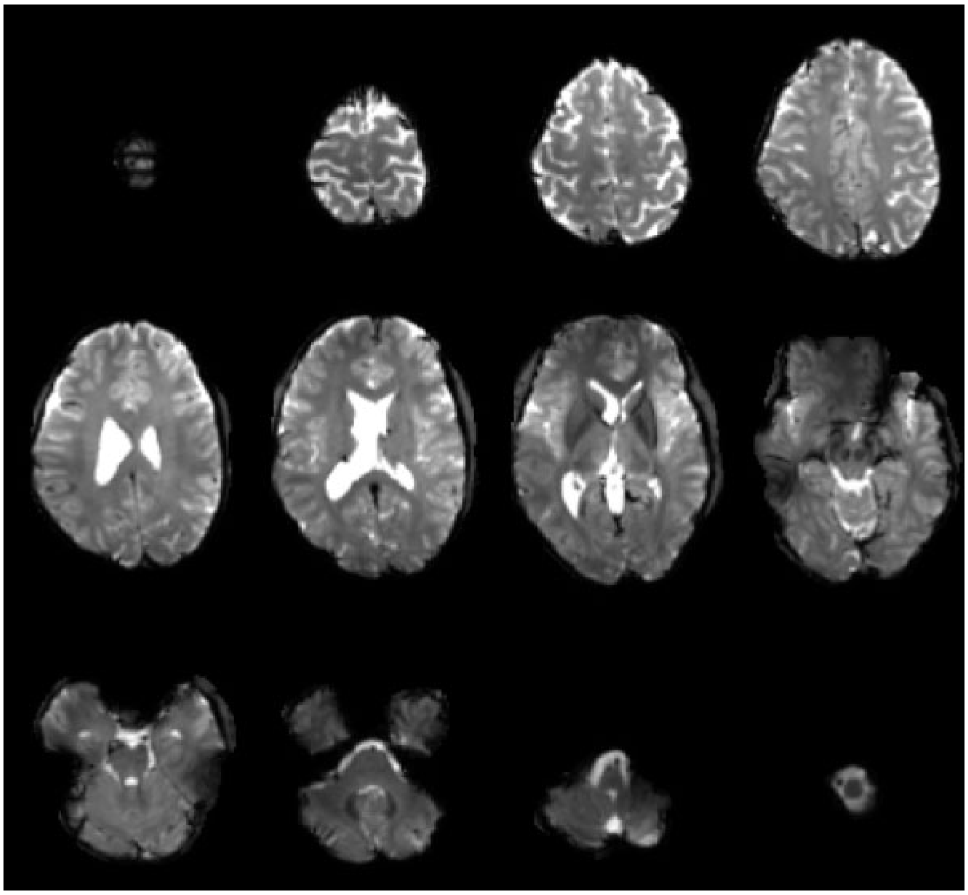

FIG. 6. Axial-oblique single shot full *k*-space gradient echo EPI of human brain using 16-channel reception. Multislice EPI was performed with matrix size 128 × 96, resulting in a nominal voxel volume of 1.875 × 1.875 × 3.5 mm^3.

(EPI) with internal phase reference (17). Data acquisition was performed using 500 kHz reception bandwidth (complex output data with 2 μs point spacing) and ramp sampling was performed on 50% of the readout gradient ramps. Two-fold oversampling was applied in the readout direction to avoid intensity roll-off at the band edge. Eight to 12 axial oblique slices were scanned with a field of view of 240 × 180 mm and a TR ranging from 1–3 s. The readout gradient was applied in the anterior–posterior direction. To evaluate system stability, low-resolution (high SNR) EPI was performed repetitively on phantoms with the following parameters: matrix size 64 × 48, slice thickness 3.5 mm, TE = 40 ms, echo train length 23 ms, TR = 1 s, 120 repetitions. To demonstrate EPI performance on human brain and evaluate SNR, EPI was performed with 128 × 96 matrix size, slice thickness 3.5 mm, TE = 45 ms, echo train length 75 ms. Finally, to demonstrate high resolution rate-2 SENSE-EPI on human brain, a 192 × 144 matrix was used, 2.0 mm thickness, TE = 55 ms, echo train length 72.5 ms.

All images were reconstructed offline with IDL software (Research Systems, Boulder, CO). For the array data, the root of summed squares combined images were calculated from the 16 individual coil images. The SENSE image reconstruction followed the algorithms published in Ref. 7, with coil sensitivity reference maps obtained from the array data itself, without body coil reference (9). For the stability evaluation, average signal and temporal standard deviation maps were calculated from the image intensity time course. The first 10 time-points were discarded to ensure steady state of the signal, and temporal SNR (TSNR) values were computed by dividing the time-averaged signal amplitude by the temporal standard deviation on a pixel-by-pixel basis.

RESULTS

The measured noise figure for the entire custom-built receiver chain was less than 0.85 dB as measured by a commercially available noise source (HP346A, Agilent Technologies, Palo Alto, CA). This compared favorably with the GE receive chain, which had a 1.5 dB noise figure for a 250 kHz EPI bandwidth. Channel-to-channel isolation was >39 dB with the custom receiver chain. The two-tone dynamic range measurements showed a 1 dB compression at an output of 80 dB above noise level and linear behavior (within 0.1 dB) up to 78 dB above noise level. No spurious signals or intermodulation products were observed up to 72 dB above noise level, at which point the dominant product rose with a slope of about 8. Note that the equivalent dynamic range is 84 dB, which is 6 dB above 78 dB range of linear behavior, as this was a two-tone measurement. A 100% duty-cycle with an output bandwidth of 1 MHz and data throughput 32 MB/s was realized within the maximum 132 MB/s data transfer (burst) rate of the PCI bus.

Region-of-interest (ROI) analysis of TSNR values on phantoms gives an average value of 571 with the array coil

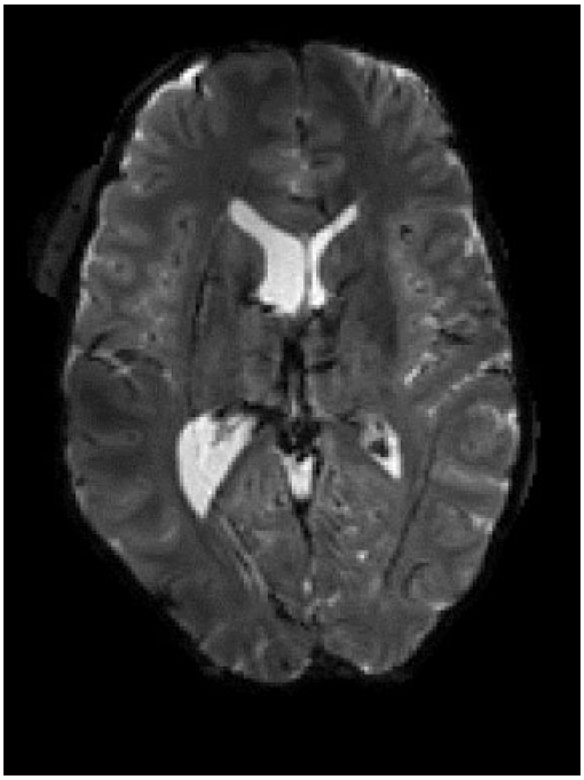

FIG. 7. Axial-oblique high-resolution rate-2 SENSE EPI of human brain using 16-channel reception. Rate-2 SENSE allowed an image matrix size of 192 × 144 (nominal resolution 1.25 × 1.25 × 2 mm^3) with relatively little EPI distortion. Note the fine anatomical detail available at this resolution.

and custom receiver, which was 1.7 times the value found with the product 28 cm birdcage T/R coil (operated in T/R mode). Single image SNR was about 800 and 500 for the array and the birdcage T/R coils, respectively. An example of single-shot gradient echo EPI human brain images from the 16 individual array channels is shown in Fig. 4. The distinct sensitivity profiles of each of the coil elements are clearly visible. Figure 5 shows an SNR comparison between the 16-channel receive coil and the product 28 cm birdcage coil. Peak cortical sensitivities were ~6 times that of the birdcage coil, while central brain SNR was improved 1.8 times. Representative multislice (oblique plane) single-shot full *k*-space high-resolution gradient echo EPI images from a normal volunteer are shown in Fig. 6. An example of multislice single-shot, high-resolution rate-2 SENSE gradient echo EPI images from the same volunteer and the same image volume are shown in Fig. 7. The SENSE *g*-factor (7) averaged over the entire brain was 1.02.

DISCUSSION

To achieve high-performance parallel MRI at 3 T, a high bandwidth receiver with 16 channels was designed and tested. The overall cost of the off-the-shelf components of the receiver was around $9,000 per channel. This included the 16-element array coil, eight Echotek boards, cables, connectors, two acquisition computers and one reconstruction (master) computer, a delay pulse generator, and an RF signal generator. This relatively low cost, combined with the modular system design, facilitates extension to more channels. Furthermore, the widespread availability and active development of the PC/Linux platform facilitates hardware upgrades and software modifications.

An acquisition bandwidth of 1 MHz, which is adequate for all brain applications with current gradient technology, could be used at 100% duty cycle. In the current configuration of 8 channels per acquisition computer, higher bandwidths, possibly needed for applications outside the brain, were also feasible, albeit at reduced acquisition duty cycle. 100% duty cycle at 2 MHz or higher sampling bandwidth could be readily achieved by increasing the number of acquisition computers and decreasing the number of RF channels (PCI cards) per computer.

The dynamic range measurement showed the receiver was linear over a range of 84 dB. This includes the effects of both the analog chain and the sampling and digital filtering. With a 14 bit ADC, a factor 40 of oversampling and an input noise level about 2 bits, the dynamic range appears to be limited almost entirely by the effects of the digitization.

With a data throughput of around 20 MB/s during a typical 16-channel fMRI experiment (1 s TR, 8 slices, 128 × 96 matrix size, 2 μs sample spacing), the amount of raw data generated during 40 min of fMRI acquisition exceeds 40 GB per subject. Therefore, raw data flow is large and data handling and storage is quite a demanding challenge.

Multichannel MRI of human brain with 16-channel whole-brain arrays offers significant SNR improvements over standard single-channel MRI with a whole-head birdcage coil. Combined with parallel MRI techniques such as SENSE, substantial improvements can be made in temporal and spatial resolution of BOLD fMRI. Excellent single-shot conventional (rate-1) and rate-2 SENSE EPI image quality was obtained at 3.0 T, showing fine anatomical detail across the entire brain.

CONCLUSION

A low-cost, scalable, 16-channel wide-band MR receiver was developed and built for a commercially available MRI scanner. Together with a 16-channel whole-brain surface coil array, it offers significant SNR improvements when compared to a standard single-channel birdcage coil. Availability of systems with a large number of high-bandwidth receive channels will greatly enhance the potential of array coils and parallel imaging techniques.

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