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Rapid Volumetric MRI Using Parallel Imaging With Order-of-Magnitude Accelerations and a 32-Element RF Coil Array: Feasibility and Implications¹

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Abstract

Rationale and Objectives— Many clinical applications of Magnetic Resonance Imaging are constrained by basic limits on imaging speed. Parallel MRI relaxes these limits by using the sensitivity patterns of arrays of radiofrequency receiver coils to encode spatial information in a manner complementary to traditional encoding with magnetic field gradients. Until now, parallel MRI has been used to achieve modest improvements in imaging speed; order-of-magnitude improvements have been elusive given fundamental losses in signal-to-noise ratio. The goal of this work was to demonstrate that, with appropriate hardware and careful SNR management, rapid volumetric imaging at high accelerations is in fact feasible.

Materials and Methods— Contrast-enhanced MRI with an axial 3D spoiled gradient echo imaging sequence was performed in healthy adult subjects using a 32-element RF coil array and a prototype 32-channel MR imaging system. Large imaging volumes were prescribed, in place of traditional limited slabs targeted only to suspect regions.

Results— As much as 16-fold net accelerations of imaging were achieved repeatably using this approach. The use of large 3D volumes allowed comprehensive anatomical coverage at clinically useful spatial and/or temporal resolution. The need for careful, time-consuming, and subject-specific scan prescription was also eliminated.

Conclusion— The highly parallel imaging approach presented here allows previously inaccessible volumetric coverage for time-sensitive MRI examinations such as contrast-enhanced MRA, and simultaneously provides a substantially simplified imaging paradigm. The resulting capability for rapid volumetric imaging promises to combine the strengths of MRI with some of the advantages of alternative imaging modalities such as multidetector CT.

Keywords

Rapid MRI; parallel MRI; volumetric imaging

Magnetic resonance imaging (MRI), as with its sister technique, multidetector X-ray computed tomography (CT), is a volumetric imaging modality. Unlike in CT, the use of magnetic field gradients for spatial encoding in MRI allows a free prescription of the orientation of acquired image planes or volumes. This same spatial encoding mechanism, however, places practical constraints on the extent of volumetric coverage for any given spatial resolution in MRI examinations. In the wake of radiofrequency (RF) pulses that excite magnetization in the imaged region, field gradients of varying amplitude, direction, or duration are applied and signal data are acquired in sequential readouts. However, the maximum rate of gradient switching is limited by the inductance of gradient coils and by the need to avoid neuromuscular stimulation from currents induced by the rapidly changing fields. Safety considerations for tissue heating also limit the rate of application of RF pulses. These physical and physiologic constraints on gradient switching rate and RF power deposition limit the rate at which MRI sequences may be played out and, consequently, the rate at which new image data may be acquired in the traditional paradigm. Meanwhile, the allowable window for data acquisition is generally limited—by the feasible breath-hold duration in abdominal and thoracic imaging, by the passage of contrast agents in vascular studies, by the dynamics of cardiac and respiratory motion in cardiac MRI, or by patient comfort and compliance. As a result of these constraints, MRI examinations are typically accomplished using multiple volumes tailored in orientation and extent to the application and anatomy of interest. Such a “tailored volume” approach, in combination with the inherent flexibility and variety of MRI pulse sequences, creates a large number of adjustable parameters and a need for careful patient-specific planning. This stands in sharp contrast to the comparatively simple acquisition paradigm of CT.

Parallel MRI can circumvent some of the basic constraints on MRI acquisition speed, and can thereby provide an alternative to the tailored-volume paradigm and its associated complexities. Parallel MRI supplements the field gradient-based encoding mechanism of traditional MRI by using the sensitivity patterns of RF coils arrayed around the imaging volume. Each coil's localized sensitivity pattern constitutes a distinct “view” of the imaged object, which may be combined with the spatial modulations produced by gradients to yield a set of projections. Because data are acquired simultaneously in all array elements, multiple projections are available in parallel, and the number of time-consuming gradient steps can be reduced while still preserving the full image information (1).

After several early suggestions for parallel data acquisition in MRI (2–5), practical parallel image reconstruction algorithms were introduced in the late 1990s (6,7). Since that time, parallel imaging techniques have been applied to achieve modest accelerations (eg, factors up to four or, in selected cases, six) in studies of a wide range of organ systems, resulting in improvements in efficiency or image quality for a variety of existing MRI applications. However, higher accelerations, at the level of an order of magnitude, would be required to enable rapid volumetric imaging in the style of multidetector CT (8), and such high accelerations have been elusive.

This article demonstrates the feasibility of highly accelerated volumetric parallel MRI using a prototype 32-channel MR system. After outlining some of the theoretical and technological challenges to highly accelerated imaging, we describe a specific volumetric imaging strategy and 32-element coil array design that addresses these challenges, and we present large-volume in vivo magnetic resonance angiographic data at acceleration factors as high as 16. Such high accelerations would be difficult to obtain for traditional limited slabs; however, by allowing heretofore inaccessible volumes to be acquired in practical imaging times, they promise to provide for routine clinical MRI some of the simplicity and generality of CT-like acquisitions while preserving the wide range of contrast options and biological and chemical selectivity associated with magnetic resonance.

MATERIALS AND METHODS

Signal-to-Noise Ratio and Challenges to High Acceleration

Increased speed in parallel MRI comes at a well-defined cost in signal-to-noise ratio (SNR) (7,9). Indeed, practical observations of SNR behavior with increasing acceleration might lead one to expect that order-of-magnitude accelerations would result in prohibitively large SNR losses for most useful MRI applications. There are two principal reasons for concern.

First, the maximum number of independent projections (and therefore the maximum acceleration factor) is set by the number of array elements, and highly accelerated parallel MRI thus requires large numbers of RF coils. As the number of adjacent coils lying on the surface of any fixed imaging volume increases, their individual size must decrease accordingly. Meanwhile, any given RF coil is known to provide optimal SNR at a depth in the vicinity of one coil diameter, with SNR falling off rapidly at greater depths. Consequently, subdivision of typical coverage areas into large numbers of coils might be expected to yield small individual elements with unacceptably low sensitivity to deep-lying structures.

The second concern arises from the observation that individual coil sensitivity profiles themselves are typically broad and overlapping, particularly at appreciable distances from the coils. Sensitivity functions at two different depths in a plane parallel to an exemplary four-element linear array demonstrate this property (Fig 1a, b). With such highly overlapping sensitivities, the matrix of projections which must be inverted in a parallel image reconstruction (1,7) becomes poorly conditioned, resulting in amplification of noise out of proportion to signal. The degradation in SNR resulting from this noise amplification is characterized by a spatially varying geometry factor, g (7):

$$\text{SNR}^{\text{accelerated}} = \frac{\text{SNR}^{\text{unaccelerated}}}{g \sqrt{R}} \quad (1)$$

(Here, R represents the acceleration factor, and the square-root dependence in Equation 1 reflects the reduced noise averaging resulting from a reduced number of acquired projections in an accelerated scan.) From the known coil sensitivities of our exemplary four-element array, one may calculate geometry factors, say for $R = 2$ (Fig 1c, d) or $R = 4$ (Fig 1e, f) along the principal axis of the array. As the resulting distributions demonstrate, the geometry factor (whose value varies appreciably across the image plane) increases in a nonlinear fashion both with increasing acceleration factor and with increasing depth. Thus high accelerations might in themselves be predicted to result in prohibitive SNR losses at useful depths. Such predictions appear to be reinforced in part by recent calculations of fundamental electrodynamic constraints on parallel imaging performance, independent of particular coil array design (10,11).

Numerical Simulations

Numerical simulations were used to demonstrate that sufficient freedom exists within the constraints of electrodynamics to achieve order-of-magnitude accelerations at clinically useful depths and signal levels. First, to examine the scaling of SNR with element size, we computed the mean relative SNR of reconstructed parallel images as a function of acceleration factor for simulated one-, two-, four-, and eight-element linear arrays with the same total extent but progressively smaller individual element size (Fig 2a). Next, using the methods described in (10), we computed the limiting value of SNR as a function of acceleration factor for large numbers of idealized coils surrounding a simulated imaging volume (Fig 2b). This value may

be shown to be the ultimate intrinsic SNR, or the maximum SNR that may be achieved using a coil array of arbitrary design (10).

Coil Array and Imaging System

To demonstrate the feasibility of order-of-magnitude accelerations in practice, we constructed a 32-element array and a supporting 32-receiver imaging system capable of receiving simultaneous data from all 32 array elements (12,13). The array (pictured in Fig 3a, b) consisted of a total of 32 loop-coil elements etched onto two separated “clamshell” formers, each containing 16 coils in a regular 4×4 grid. The individual coil size (10.5 cm superior-inferior by 8.1 cm left-right) and intercoil spacing (superior-inferior direction overlap of 1.8 cm and left-right direction gap of 1.4 cm) were designed to optimize net SNR for 10- to 16-fold accelerations in the central portion of a large simulated volume (12). The imaging system was constructed by integrating multiple sets of MRI system electronics (Fig 3c), including analog-to-digital converters and digital data pipelines, into a single clinical whole-body GE 1.5-Tesla TwinSpeed scanner (GE Healthcare Technologies, Waukesha, WI). All receivers were frequency and trigger locked to each other, and gradient and RF pulse sequences were adapted to make use of the synchronization mechanism.

Data Acquisition

Eleven healthy adult subjects were imaged using our many-channel system. Informed consent was obtained from all subjects in accordance with the guidelines of our Institutional Review Board. An axial three-dimensional spoiled gradient echo (SPGR) imaging sequence was used in all studies.

Four distinct imaging protocols were used:

1. Matrix size = $160 \times 256 \times 48$, field of view (FOV) = $40 \text{ cm} \times 40 \text{ cm} \times 19 \text{ cm}$, echo time (TE) = 0.9 ms, repetition time (TR) = 2.8 ms, flip angle = 12° , bandwidth = 125 kHz (Fig 3d).
2. Matrix size = $128 \times 128 \times 128$, FOV = $36 \text{ cm} \times 36 \text{ cm} \times 36 \text{ cm}$, TE = 1.3 ms, TR = 3.4 ms, flip angle = 25° , bandwidth = 62.5 kHz. An acceleration factor of four in the left-right phase-encoding direction and three in the superior-inferior phase-encoding direction was used. Twenty milliliters (0.1 mmol/kg body weight) of gadopentetate dimeglumine (Berlex Laboratories, Wayne, NJ) were injected into an antecubital vein of the right arm at 2 mL/second, followed by a 20-mL saline flush at 2 mL/sec, and dynamic imaging began 12 seconds after the start of contrast administration. As a result of the high temporal resolution, no timing bolus or bolus tracking preparation was required (Fig 4a–e).
3. Similar parameters as for protocol #2, but employing an acceleration factor of 16 (4×4) (Fig 4f–i).
4. Matrix size = $256 \times 256 \times 180$, FOV = $44 \text{ cm} \times 44 \text{ cm} \times 40 \text{ cm}$, TE = 1.9 ms, TR = 4.6 ms, flip angle = 25° , bandwidth = 62.5 kHz, fat saturation employed. Imaging was initiated after contrast administration (same bolus parameters as for protocol #2), after a delay based on the arterial transit time to the renal arteries determined using a test bolus (14) (Fig 4j–n).

In all cases, a low-resolution imaging volume was acquired as a sensitivity reference (6,7) before accelerated imaging and contrast administration. This calibration volume also served to verify that the desired volumetric coverage had been obtained. The FOVs selected were sufficient to contain a complete cross-section of every subject, thus no additional scout scans were required, and no subject-specific angulations of the imaging volume were required. (See

Discussion for additional comments regarding image plane prescription and selection of phase- and frequency-encoding directions.)

RESULTS

Figure 2 shows the results of SNR simulations, both for array performance as a function of element number and size (Fig 2a) and for ultimate intrinsic SNR as a function of acceleration (Fig 2b). In Fig 2a, SNR averaged along a long axis separated by the full width of the phantom from the coils is shown, scaled to unity at an acceleration factor of 1. The results of these computations reveal that the use of large arrays of small elements does not, in principle, pose an obstacle to high accelerations. Note that the net SNR is always highest for the largest number of elements, despite the fact that the smaller individual coils of the multielement arrays have markedly reduced individual SNR at the selected depth. This result is consistent with the predictions of superposition principles, which suggest that the loss of depth sensitivity by small individual elements will be compensated by an associated increase in the number of elements, provided that noise from individual coil circuits is kept under control (15–18). As an illustration of the superposition argument, imagine that a single rectangular coil is divided into two coils each with half the total area. By superposition, a unit current passed simultaneously through each of the two subcoils will generate the same net electric and magnetic fields as a unit current passed through the single larger coil, because the effects of oppositely directed current in the shared leg of the two adjacent subcoils will cancel. Applying the principle of reciprocity, both the signal (derived from the magnetic field) and the noise (derived from the electric field) received in the single large coil *at any depth* can be replicated by a simple sum of the signal and noise received in the two subcoils. However, an optimal linear combination of component coil data will by definition yield an SNR greater than or equal to this simple linear summation. In other words, the added freedom of combination provided by the two-coil arrangement allows higher SNR than the one-coil arrangement at arbitrary depth. This argument may be extended to larger numbers of elements by further subdividing each subcoil. Such a superposition argument holds equally well for parallel imaging reconstruction at any fixed acceleration factor.

It may, however, be noted from Fig 2a that SNR decreases markedly with increasing acceleration factor for any given number of coils, as expected because of the nonunitarity of the image reconstruction in combination with reduced noise averaging. Can the use of a sufficient number of array elements overcome this mechanism of SNR degradation? The dashed line in Fig 2b shows the ultimate intrinsic SNR at a typical clinical field strength of 1.5 T for a point at the center of an elliptical cylinder as a function of acceleration factor R along the short axis of the elliptical cross-section. If order-of-magnitude accelerations are desired, this curve is rather disheartening. At an acceleration factor of five, the maximum achievable SNR has dropped to approximately 20% of the unaccelerated value; by an acceleration factor of eight, the value has decreased to 2%. This rapid decrease in achievable SNR, above and beyond the baseline inverse square root dependence, is a consequence of the attempt to resolve finer and finer structures with slowly varying fields.

Such difficulties may be mitigated, however, by employing multidimensional accelerations in studies for which gradient phase encoding is applied along more than one direction (19). The solid line in Fig 2b shows the ultimate intrinsic SNR as a function of acceleration, with equal acceleration factors along both the long axis and the short axis of the elliptical object. Achievable SNR drops to 20% of its baseline value only at a combined acceleration factor of 12. At an acceleration factor of 24, a 5% relative SNR remains. This more encouraging behavior results from the spreading of the encoding burden into multiple dimensions: it is easier, for example, to distinguish the four corners of a square than four points along one side of the same square using encoding functions of a fixed breath.

These theoretical arguments and numerical results suggest that a suitably designed coil array—capable of efficient multidimensional encoding and possessing a sufficiently high baseline SNR—can indeed be used to achieve higher accelerations than have previously been thought possible. The results in Fig 3 and Fig 4 demonstrate such accelerations in practice using our 32-element array and 32-channel imaging system.

Figure 3d shows sample images from three-dimensional data sets at various acceleration factors from 1 to 24 in one subject. The 34-second acquisition time for the unaccelerated three-dimensional SPGR imaging sequence (an unacceptably long imaging time for most clinical breath-hold studies) was reduced to as little as 1.4 seconds at a factor of 24 acceleration. Substantial degradation in image quality from reduced SNR is evident at the highest accelerations. However, for acceleration factors as high as 12 to 16, reasonable image quality is preserved. We therefore selected acceleration factors in this range for subsequent contrast-enhanced studies.

Figure 4 demonstrates some of the ways in which the increased speed of highly accelerated parallel imaging studies may be put to use for in vivo imaging. In one such study, a 54-second three-dimensional SPGR acquisition in the abdomen and pelvis was reduced, by a factor of 12, to 4.5 seconds, and multiple datasets were acquired dynamically during inflow of the gadolinium-based contrast agent. Coronal maximum intensity projections of the resulting data sets clearly show passage of contrast from the arterial to the venous system (Fig 4a–d). (Contrast seen in the renal collecting system in the first frame, Fig 4a, is residual from a previous injection.) Similarly, in a study of the pulmonary vasculature for another subject, 58-second acquisitions were reduced by a factor of 16 to 3.6 seconds per dynamic volume, allowing effective capture of the comparatively brief pulmonary phase while preserving full volumetric coverage (Fig 4f–i).

These data sets (Fig 4a–i) were of relatively modest spatial resolution ($2.8 \text{ mm} \times 2.8 \text{ mm} \times 2.8 \text{ mm}$) by the standards of clinical practice. Angiographic studies were also performed at substantially higher spatial resolution ($1.7 \text{ mm} \times 1.7 \text{ mm} \times 2.2 \text{ mm}$; Fig 4j–n). For these studies, an impractical 4-minute, 24-second acquisition was reduced, again by a factor of 12, to a 22-second breath-hold. Various rotations show detailed views of the superior mesenteric artery arcade, the renal arteries, and the iliac and femoral arteries. Note the excellent detail of more peripheral vessels captured with a full sampling along the anteroposterior axis (Fig 4n).

DISCUSSION

The reductions in scan time and increases in volumetric coverage described in this feasibility study could have significant implications for the day-to-day practice of clinical MRI. For contrast-enhanced studies and body imaging studies, the acquisition of comprehensive volumes has traditionally been precluded by scan time constraints. For example, if an unaccelerated 1-minute acquisition had been used in the studies shown in Fig 4a–i, contrast would have been distributed in both arteries and veins, and the images would most likely have been corrupted by respiratory motion, because it is a rare patient who is capable of such a sustained breath-hold. Obviously, the need for a 4.5-minute breath-hold duration would have precluded unaccelerated imaging with the protocol used in Fig 4j–n.

In traditional magnetic resonance angiographic studies, anatomic coverage is typically truncated, or else spatial resolution in at least one plane is sacrificed, to bring scan time into an acceptable range. Typical slab dimensions for a conventional renal angiography study are indicated with dashed lines on a lateral view in Fig 4e. In the accelerated studies shown in Fig 4a–i, however, comprehensive volumetric coverage of the entire anatomy (including both soft tissues and vasculature) was obtained not just once, but in multiple distinct phases with high

temporal resolution. Alternatively, as shown in Fig 4i–n, high spatial resolution was obtained over a large volume in a single breath-hold.

It might be objected that anatomic coverage sufficient to answer particular clinical questions can already be obtained at high spatial and temporal resolution using the targeted volume paradigm; and it should be noted that the high levels of acceleration reported here may be more difficult to bring to bear on imaging sequences with more limited slab thickness. However, the rapid acquisition of large data volumes has a number of noteworthy advantages over the limited-slab paradigm. Using the traditional approach in the chest and abdomen, the process of scan planning is often iterative and time consuming, because the operator strives for sufficient coverage, maximum spatial resolution, and feasible breath-hold durations. With comprehensive volumes, scan prescription is dramatically simplified, allowing anatomic or functional scanning of any anatomic variant at the press of a button. The frequently encountered difficulty of missed coil or image plane placement is also avoided. Substantial time-savings in image prescription, as compared with the traditional limited-slab approach, were noted in this feasibility study, though detailed comparisons will be reserved for ongoing clinical studies using the 32-channel apparatus.

The use of otherwise unattainable imaging volumes also enables effective screening for incidental findings or evidence of diffuse disease processes. For applications such as pulmonary angiography, the temporally resolved volumetric approach offers a versatile set of tradeoffs to accommodate individuals with the most limited breath hold capacities. Although subsecond temporal resolution for the pulmonary arteries has been previously reported, this could only be achieved with a substantial sacrifice in through-plane resolution, such that off-axis maximum intensity projections were found not to be useful (20). Such a strategy has limited clinical utility as compared with a comprehensive volumetric approach.

For the axial studies described in this article, the comparatively compact anteroposterior direction, rather than the longer superior-inferior direction, was selected for frequency encoding. This somewhat unconventional choice is a feature of our particular coil array design, which is optimized for accelerations along left-right and superior-inferior phase-encoding directions. If compact imaging volumes more closely tailored to individual body dimensions were desired (with some investment of patient-specific scan planning), alternative designs, such as a circumferential 32-element design, would allow high accelerations with a more traditional choice of frequency-encoding direction. In fact, an encircling grid with a sufficient number of elements would allow arbitrary placement of encoding directions, because such an arrangement would provide distinct localized sensitivity patterns along all three dimensions.

It should also be noted that the imaging protocols reported here did not employ any of the existing repertoire of time- or data-sharing approaches commonly used to accelerate magnetic resonance angiography studies, such as Time-Resolved Imaging of Contrast Kinetics (21). Such approaches may be combined in a straightforward manner with highly parallel MRI, resulting in still higher imaging speeds. Moving table MRI approaches have also been proposed for situations in which extensive anatomic coverage is desired (eg, screening for atherosclerotic disease or tumor metastases) (22). Highly parallel MRI would greatly enhance the capabilities of such studies. In fact, if combined with moving table approaches, highly parallel volumetric MRI would begin to resemble multi-detector CT imaging, in which rapid volumetric data are routinely acquired over large portions of the body.

Contrast-enhanced MRI studies are prime candidates for highly accelerated parallel MRI, given the competing constraints of spatial and temporal resolution in these studies, in combination with their high intrinsic contrast-to-noise ratio. However, highly accelerated studies on our 32-channel system using three-dimensional imaging sequences without injected contrast agents

have also yielded high-quality images, for example for coronary artery imaging (23,24), brain imaging (25), or abdominal screening (26). This is in part because three-dimensional imaging sequences in general also offer a particular synergy with parallel imaging, not only because of the availability of multiple directions suitable for acceleration, but also because SNR in these sequences increases with the quantity of acquired data. Highly parallel MRI enables large volumetric acquisitions with otherwise prohibitive imaging times, and the resulting gains in baseline SNR serve to offset at least in part the SNR losses associated with parallel imaging. Studies are under way exploring the feasibility of highly accelerated three-dimensional coronary artery imaging and three-dimensional magnetic resonance urography with comprehensive anatomic and vascular coverage.

Two areas of ongoing research promise to improve the balance of SNR versus acceleration for highly parallel imaging. Implementation at field strengths higher than 1.5 T may be expected to improve both baseline SNR (because of increased nuclear spin polarization) and noise amplification factor g (from improved RF focusing capability at increased Larmor frequencies) (10,11), thereby allowing still higher accelerations. Use of polarization-enhancing contrast agents (27–29) will also increase achievable acceleration.

Meanwhile, the order-of-magnitude accelerations demonstrated here promise a change of paradigm for in vivo MRI. In addition to the well-known use of MRI for targeted examinations of suspect tissue regions, highly parallel MRI offers the capability to perform rapid time-resolved evaluations of large volumes, whether for clinical whole-body screening applications or for detailed spatio-temporal characterization of complex structures. Modern experience with multidetector CT has demonstrated some of the benefits of rapid volumetric imaging both for diagnostic information content and for clinical throughput. The value and clinical utilization of MRI—with its expanded contrast options and absence of radiation exposure—can be expected to increase as it begins to offer a similar capability for rapid and simple volumetric imaging.

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Figure 1.

Simulated coil sensitivities and spatially varying noise amplification factors for an image plane parallel to a four-element radio-frequency coil array at two distinct vertical distances from the array and at two distinct acceleration factors. Coil sensitivities at 40 mm (a) and at 70 mm (b) vertical distance, computed using BiotSavart field calculations, are plotted in arbitrary vertical units, with the color map reflecting vertical scale for ease of visualization. The conductor geometry of the array, whose total extent is 260 mm $\times$ 230 mm and in which adjacent array elements are overlapped to minimize inductive coupling, is shown above each coil sensitivity surface plot. The horizontal scale represents the image field of view of 300 mm $\times$ 220 mm. Noise amplification factors (g in Eq 1) for the same two vertical distances are plotted over the same field of view, for SENSitivity Encoding (7) image reconstructions with acceleration factors of $R = 2$ (c, d) and $R = 4$ (e, f) along the long axis of the array. The vertical scale on the noise amplification plots indicates the magnitude of the dimensionless multiplier g , with color again reflecting vertical scale.

Figure 2.

Predicted scaling of signal-to-noise ratio (SNR) as a function of acceleration factor. (a) Comparison of mean relative SNR for one-, two-, four-, and eight-element radiofrequency coil arrays with the same total extent but progressively smaller individual element size. Finite-difference time-domain (FDTD) simulations (30) were used to calculate coil sensitivities and noise characteristics for each of these arrays (shown at right), placed on a rectangular solid with dimensions and electrical properties matching those of an existing phantom used for SNR measurements (18). Pixel-by-pixel SNR was calculated for Cartesian SENSitivity Encoding (SENSE) image reconstructions at various accelerations. The mean value of SNR along a line bisecting the bottom face of the phantom is plotted on a vertical semilog scale, normalized to unity at an acceleration factor of 1. (b) Ultimate intrinsic or maximum achievable SNR at the center of an elliptic volume approximating the size and shape of a human torso (10), as a function of acceleration factor. For these calculations, coil sensitivities and noise correlation matrices for a large set of plane-wave electromagnetic fields were used in SENSE reconstructions at various accelerations, and the SNR was computed accordingly (10). Once again, SNR is plotted on a vertical semilog scale and normalized to unity at an acceleration factor of 1. Dashed line: limiting SNR for one-dimensional accelerations along the short axis of the elliptical cross-section; solid line: limiting SNR for equal two-dimensional accelerations along both the long and the short axes of the ellipse.

Figure 3.

Highly accelerated parallel imaging with a 32-element array. (a, b) Photographs of the 32-element array, which contains two separable "clamshell" formers each with 16 rectangular elements. (c) Photograph of four GE System electronics cabinets, which were synchronized to yield a 32-receiver system capable of accommodating the 32-element array. (d) Coronal gradient-echo images extracted from three-dimensional volumetric datasets covering the abdomen and thorax of a healthy adult subject. The unaccelerated volume was acquired in 34 seconds, using a three-dimensional spoiled gradient echo imaging sequence. Progressive two-dimensional accelerations allowed volumetric acquisitions in as little as 1.4 seconds. The acquisition time at each level of acceleration is shown above or below the corresponding image, with acceleration factors along the left-right and the superior-inferior directions indicated in parentheses. The total acceleration in each case is equal to the product of accelerations along each dimension (ie, a maximum acceleration factor of 24 is shown at the bottom right).

Figure 4.

Contrast-enhanced magnetic resonance (MR) angiograms acquired at an acceleration factor of 12 to 16 in three healthy adult subjects. (a–d) Coronal maximum intensity projections (MIPs) of multiple dynamic 4.5-second volumetric acquisitions, reduced in duration by a factor of 12 from 54-second acquisitions, are shown as successive time-resolved frames (identified by time delay following contrast bolus injection). Yellow dashed lines in (e) show typical restricted slab dimensions for a conventional unaccelerated renal MR angiography study, as compared with the full volumetric coverage (blue outline) of the accelerated study. (f–i) Various rotations of volume-rendered data from the pulmonary phase of another time-resolved angiographic study, accelerated by a factor of 16. These data were extracted from a full volumetric acquisition similar to that shown in (a–e). (j–n) Volume-rendered data with high spatial resolution shown in various rotated frames. Full volumetric coverage of the imaged structures was obtained in a 22-second breath-hold, reduced by a factor of 12 from an impractical 4 minutes, 24 seconds. The four images to the left (j–m) were obtained in the first postcontrast phase, timed for optimal arterial enhancement. The rightmost image (n), showing a zoomed segment of the full acquired volume, was generated from a second postcontrast acquisition obtained immediately after the first acquisition.