

Sensitivity Profiles From an Array of Coils for Encoding and Reconstruction in Parallel (SPACE RIP)

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A new parallel imaging technique was implemented which can result in reduced image acquisition times in MRI. MR data is acquired in parallel using an array of receiver coils and then reconstructed simultaneously with multiple processors. The method requires the initial estimation of the 2D sensitivity profile of each coil used in the receiver array. These sensitivity profiles are then used to partially encode the images of interest. A fraction of the total number of k -space lines is consequently acquired and used in a parallel reconstruction scheme, allowing for a substantial reduction in scanning and display times. This technique is in the family of parallel acquisition schemes such as simultaneous acquisition of spatial harmonics (SMASH) and sensitivity encoding (SENSE). It extends the use of the SMASH method to allow the placement of the receiver coil array around the object of interest, enabling imaging of any plane within the volume of interest. In addition, this technique permits the arbitrary choice of the set of k -space lines used in the reconstruction and lends itself to parallel reconstruction, hence allowing for real-time rendering. Simulated results with a 16-fold increase in temporal resolution are shown, as are experimental results with a 4-fold increase in temporal resolution. Magn Reson Med 44:301–308, 2000. © 2000 Wiley-Liss, Inc.

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Despite the many advances in ultrafast MRI, there is always a need for further increases in the speed of image acquisition. Dynamic imaging applications like cardiac and interventional imaging would be greatly served with an order of magnitude reduction in scan time without sacrificing spatial resolution and signal to noise ratio (SNR). Conventional multiecho imaging techniques such as RARE and EPI (1,2) are currently very fast. The new field of parallel imaging can be combined with these methods to further increase imaging speed.

In previous work on parallel imaging, Hutchinson and Raff (3) demonstrated the theoretical feasibility of fast data acquisitions using multiple detectors in MRI. In a subsequent work, Kwiat et al. (4) investigated methods to solve the inverse source problem on MR signals received in multiple RF receiver coils. Their technique required the use of a number of RF coils equal to the number of pixels in the image, as well as greatly increased receiver coil sensitivities. These requirements are quite impractical in conventional MR imaging, where the usual number of

pixels in an image is on the order of 256×256 , hence the technique was never successfully used in medical imaging.

A number of more promising parallel imaging techniques have been described in the literature (5–7) which use the sensitivity profiles of RF receiver coils for spatial encoding. Ra and Rim (5) described a method that uses sets of equally spaced k -space lines from multiple receiver coils and combines them with sensitivity profile information in order to remove the aliasing that occurs due to the undersampling. A 4-fold decrease in the image acquisition time of a water phantom was shown using an array of four coils, although no biological images were shown.

The SMASH method proposed by Sodickson and Manning (6) has proven more practical, yielding good results in volunteers with clinical implementations. SMASH is designed to enhance imaging speed by using multiple receiver RF coils. It is based on the computation of the sensitivity profiles of the coils in one direction. These profiles are then weighted appropriately and combined linearly in order to form sinusoidal harmonics which are used to generate the k -space lines that are missing due to undersampling. This technique showed an 8-fold increase in imaging speed. SMASH has some inflexibility in the choice of imaging planes due its restriction on the placement of receiver coils along one direction.

The SENSE method proposed by Pruessmann et al. (7) is another parallel imaging technique which relies on the use of 2D sensitivity profile information in order to reduce image acquisition times in MRI. Like SMASH, the cartesian version of SENSE requires the acquisition of equally spaced k -space lines in order to reconstruct sensitivity weighted, aliased versions of the image. The aliasing is then removed with the use of the sensitivity profile information at each pixel. This is done by resolving in the space domain the linear system of equations obeyed by the intensity of each pixel in the image.

A generalization of SENSE was introduced by Pruessmann et al. (8) which would allow for data to be sampled along arbitrary k -space trajectories. A high computational cost, however, accompanies the arbitrary k -space sampling in generalized SENSE methods, currently making reconstruction inconvenient. An elaborate description of the differences between SMASH and SENSE can be found in the SENSE manuscript (7).

In this article, we present a parallel imaging and reconstruction technique which attempts to generalize the SMASH approach by allowing for the arbitrary placement of RF receiver coils around the object to be imaged as well as for the use of any combination of k -space lines as opposed to regularly spaced ones. In addition, our reconstruction technique is completely parallel, allowing for

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real-time rendering possibilities with the use of multiple processors.

METHODS

Encoding Scheme

The concept of parallel imaging is based on using multiple receiver coils, with each providing independent information about the image.

The MR signal received in a coil having $W_k(x, y)$ as its complex 2D sensitivity profile, when neglecting all relaxation phenomena, can be written as:

$$s_k(G_y^g, t) = \iint r(x, y) W_k(x, y) e^{i\gamma(G_x x t + G_y^g y \tau)} dx dy, \quad [1]$$

where $r(x, y)$ denotes the proton density function, G_x represents the readout gradient amplitude applied in the x direction, G_y^g represents the phase encoding gradient applied during the g^{th} acquisition, x and y represent the x and y positions, respectively, and τ is the pulse width of the phase encoding gradient G_y^g .

In most conventional serial imaging sequences, the readout gradient is constant along one direction and the phase encoding is applied along an orthogonal direction. In addition, only one receiver coil is used to collect all the data required to reconstruct a digitized version of $r(x, y)$, with the tacit assumption that $W_k(x, y) = 1$. To achieve this, the phase encoding gradient G_y is varied so as to cover all of k -space with the desired resolution. For each value of G_y^g , an echo is acquired, making serial imaging a time-consuming procedure. In this technique, we use sensitivity profile information from a number of receiver coils in order to minimize the number of acquisitions needed to estimate and reconstruct $r(x, y)$. Taking the Fourier transform of Eq. [1] along the x direction when a phase encoding gradient G_y^g has been applied yields:

$$S_k(G_y^g, x) = \int r(x, y) W_k(x, y) e^{i\gamma(G_y^g y \tau)} dy, \quad [2]$$

which is the phase modulated projection of the sensitivity weighted image onto the x axis. For the purpose of dis-

cretization, we expand $r(x, y)$ and $W_k(x, y)$ along the y direction in terms of a spatially localized set of orthogonal sampling functions $\Omega_n(y)$ to obtain the following equations:

$$r(x, y) = \sum_{n=1}^N \rho(x, n) \Omega_n(y), \quad [3]$$

and,

$$W_k(x, y) e^{i\gamma(G_y^g y \tau)} = \sum_{n'=1}^N W_k(x, n') e^{i\gamma(G_y^g n' \tau)} \Omega_{n'}(y). \quad [4]$$

where N is the number of pixels in the y direction. Combining Eqs. [3] and [4], we get:

$$S_k(G_y^g, x) = \int \sum_{n=1}^N \rho(x, n) \Omega_n(y) \sum_{n'=1}^N W_k(x, n') e^{i\gamma(G_y^g n' \tau)} \Omega_{n'}(y) dy. \quad [5]$$

Rearranging the terms for simplification yields:

$$S_k(G_y^g, x) = \sum_{n=1}^N \rho(x, n) W_k(x, n) e^{i\gamma(G_y^g n \tau)} \int \Omega_n(y) \Omega_n(y) dy. \quad [6]$$

Since $\Omega(y)$ is orthonormal, we have:

$$\int \Omega_n(y) \Omega_{n'}(y) dy = \delta(n, n'). \quad [7]$$

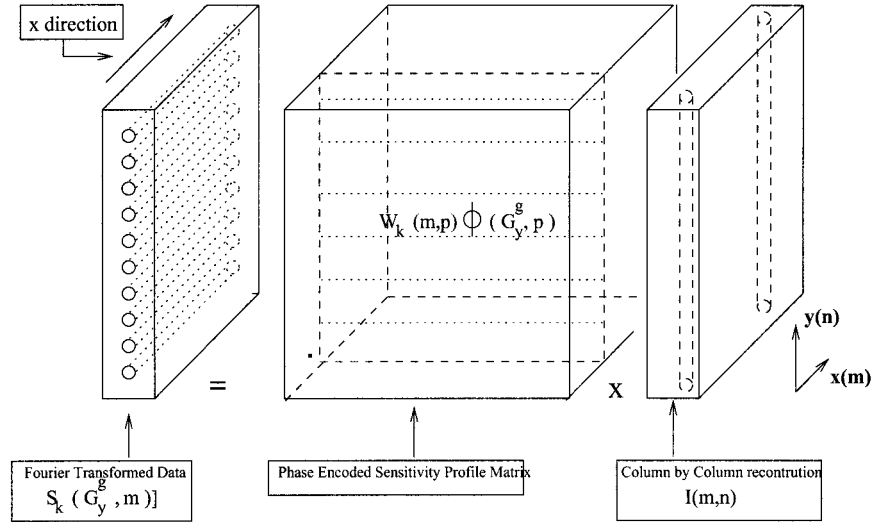
Therefore, Eq. [6] can be written as:

$$S_k(G_y^g, x) = \sum_{n=1}^N \rho(x, n) W_k(x, n) e^{i\gamma(G_y^g n \tau)}. \quad [8]$$

This expression can be converted to matrix form for each position x along the horizontal direction of the image, as follows:

$$\begin{pmatrix} S_1(G_y^1, x) \\ \vdots \\ S_1(G_y^F, x) \\ \vdots \\ S_k(G_y^1, x) \\ \vdots \\ S_k(G_y^F, x) \end{pmatrix} = \begin{pmatrix} W_1(x, 1) e^{i\gamma(G_y^1 \tau)} & \cdots & W_1(x, N) e^{i\gamma(G_y^1 N \tau)} \\ \vdots & \cdots & \vdots \\ W_1(x, 1) e^{i\gamma(G_y^F \tau)} & \cdots & W_1(x, N) e^{i\gamma(G_y^F N \tau)} \\ \vdots & \cdots & \vdots \\ W_2(x, 1) e^{i\gamma(G_y^1 \tau)} & \cdots & W_2(x, N) e^{i\gamma(G_y^1 N \tau)} \\ \vdots & \cdots & \vdots \\ W_2(x, 1) e^{i\gamma(G_y^F \tau)} & \cdots & W_2(x, N) e^{i\gamma(G_y^F N \tau)} \\ \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots \\ W_k(x, 1) e^{i\gamma(G_y^1 \tau)} & \cdots & W_k(x, N) e^{i\gamma(G_y^1 N \tau)} \\ \vdots & \cdots & \vdots \\ W_k(x, 1) e^{i\gamma(G_y^F \tau)} & \cdots & W_k(x, N) e^{i\gamma(G_y^F N \tau)} \end{pmatrix} \cdot \begin{pmatrix} \rho(x, 1) \\ \rho(x, 2) \\ \rho(x, 3) \\ \vdots \\ \vdots \\ \vdots \\ \rho(x, N) \end{pmatrix} \quad [9]$$

FIG. 1. Schematic representation of the parallel reconstruction scheme. The matrix to the left represents the 1DFT of the chosen k -space data, the block of matrices in the center is the 3D sensitivity array formed by stacking M 2D matrices such as the one expressed in Eq. [9]. The matrix on the right represents the image, which is reconstructed column by column by inverting each matrix in the sensitivity array.



where F is the number of phase encodes used in the experiment, and K is the number of coils.

Equation [9] is a matrix equation where the term on the left side of the equality is a $K \times F$ element vector containing the F phase encoded values for all K coils. The term on the far right is an N -element vector representing the “image” for one column. The middle term in Eq. [9] is a matrix with $K \times F$ rows and N columns which is constructed based on the sensitivity profiles and phase encodes used. Hence, this approach is not restricted to the case where $K \times F = N$. Solving Eq. [9] for each position along the x axis yields a column by column reconstruction of the image.

Figure 1 shows a schematic representation of the reconstruction process. As described above, each column in the image is reconstructed separately. In the case where the image matrix has N rows and M columns, a block of M matrices must be inverted to reconstruct the M columns of the image. The matrices are not necessarily square, so that a pseudoinverse must be computed for each column. The choice of the number of phase encodes F affects the quality of the reconstruction. Increasing F results in an increase of the rank of the matrices, yielding pseudoinverses that are better conditioned. There is a large computation load associated with this reconstruction; however, the potential for parallelization is obvious, since each column can be reconstructed separately. For each slice, the pseudoinverses have to be computed only once. Subsequent updates of the same slice can be reconstructed by simple matrix vector multiplication, reducing reconstruction times to real-time rates.

Conditioning and the Choice of the Phase Modulations

The reconstruction scheme outlined in the previous section is based on matrix inversion. In order to ensure a stable and robust reconstruction, the condition number of the inverted matrices, defined as the ratio of the largest eigenvalue to the lowest eigenvalue, should be minimized (9). Equation [9] shows that the condition number depends on the number of phase encodes F acquired per coil. It is also affected by the choice of the phase encodes used in

the acquisition, as well as on the specifications of the receiver coils, which include the sensitivity profiles and the RF penetration. In addition, the condition number is affected by the SNR of the sensitivity profile estimations.

Our results show (Fig. 6) that increasing the number of phase encodes used in the reconstruction would enhance the conditioning of the reconstruction matrices. To avoid errors due to numerical propagation, the pseudoinverse of each reconstruction matrix is computed after setting a minimum threshold to the eigenvalues. This effectively removes any noise amplification due to bad conditioning. For our reconstructions, we chose a cutoff threshold of 5% of the maximum eigenvalue whereby all eigenvalues below that threshold are set to zero and therefore do not contribute to the reconstruction. Our results also show that more RF penetration contributes to better conditioning. Finally, if both the choice of the k -space encodes and the coil penetration are set, the condition number would be expected to depend on the k -space characteristic of the sensitivity profiles of the receiver coils.

k -Space Coverage

In order to appropriately cover the k -space of the image $I(x, y)$, the choice of the phase modulations used in the inversion matrix should be determined by the frequency content of the sensitivity profile. In the spatial domain, the image received in a coil having a sensitivity profile $W_c(x, y)$ can be written as $I_c(x, y) = I(x, y)W_c(x, y)$. In the frequency domain, the k -space profile of $I_c(x, y)$ is the convolution of the k -space profile $I(k_x, k_y)$ of the image $I(x, y)$, with the k -space profile $W_c(k_x, k_y)$ of the sensitivity profile $W_c(x, y)$. This convolution amounts to a blurring of the k -space data $I(k_x, k_y)$ of the image. Since a different convolution is performed for each coil, a different blurring of $I(k_x, k_y)$ occurs at each coil. Subsampling the convolved k -space data received in different coils therefore results in different coverages of the k -space of the image $I(x, y)$. Hence, in order to get the best k -space coverage of the $I(x, y)$ for a given $W_c(x, y)$, it is necessary to optimally sample the k -space data from all the coils.

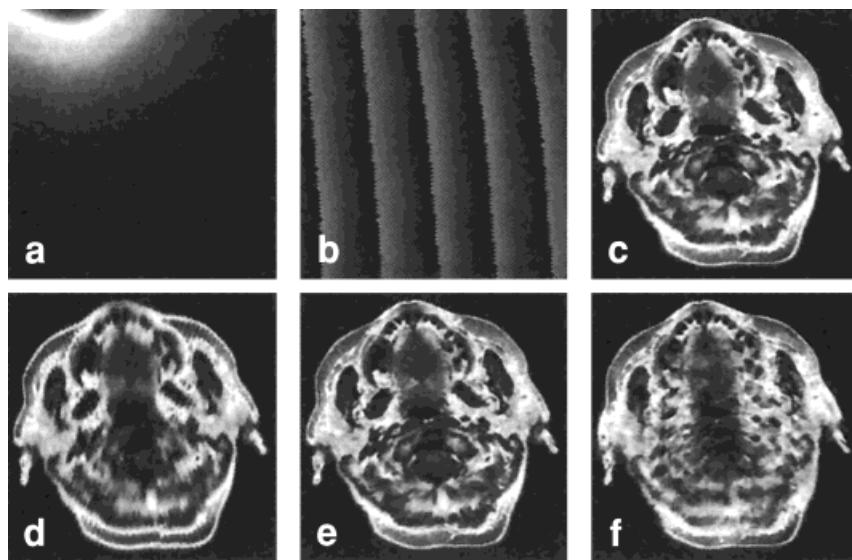


FIG. 2. **a:** Magnitude of one of the sensitivity profiles used for encoding. **b:** Phase profile used on all sensitivity profiles. **c:** Reference image reconstructed by 2DFT using all k -space lines. Sixteen lines of k -space were used in the simulation assuming 16 coils arranged every 22.5° around the head. Parts **d–f** were reconstructed by using the k -space lines acquired from skipping, respectively, 1, 8, and 16 lines.

In contrast to SMASH (6), which requires the use of equally spaced k -space lines, this technique is completely flexible as to that choice. In this work, we use equally spaced k -space lines to demonstrate the technique, and show how more optimal sets of k -space lines can be found to achieve better results. Finding the optimal set of k -space lines is, however, a subject beyond the scope of the current work.

Sensitivity Profile Calculation

Our method is based on using the sensitivity profiles of RF pickup coils in order to encode MR images. To calculate these profiles, a number of techniques described in the literature could be used (7,10–12). In this article, we use a technique that only requires comparison between body coil and surface coil images without any filtering or other numerical manipulations. It was chosen for its simplicity and adaptability to real-time applications. A baseline image of a homogeneous water phantom is acquired using an RF coil with a homogeneous sensitivity profile covering the whole image. This image can be written as $I(x, y)$. Subsequently, individual images of the same water phantom are acquired using each of the surface coils. The image acquired using the k^{th} coil can be represented as $I(x, y)W_k(x, y)$ where $W_k(x, y)$ is the sensitivity profile of the surface coil. Taking the point-by-point ratio of the two images yields the sensitivity profile $W_k(x, y)$.

The sensitivity profiles of the receiver coils depend on the loading. We assume that the variation incurred by these profiles amounts to a constant scaling between any two different loads. Therefore, in order to find the sensitivity profile of the receiver coils when loaded with an arbitrary body of interest, we perform a sensitivity profile estimation on that body as described above, then compare it to the sensitivity profiles computed on the homogeneous water phantom in order to extract the scaling factor. The sensitivity profiles calculated from the homogeneous water phantom are then multiplied by the scaling factor and used in the encoding scheme. This is done in order to get maximum coverage of the field of view by the sensitivity

profiles. Optimizations of the computed sensitivity profiles by smoothing and interpolation have been proposed in SENSE (7) and may well be used to refine the sensitivity profile estimations in our technique, leading to better conditioned reconstructions.

RESULTS

We show the results from a simulation of the method assuming a 16-element coil as well as the results from an experimental implementation of the method with a 4-element array coil. All experiments were performed on 1.5 T GE clinical MRI systems operating at either the 5.7 SIGNA or 8.2.5 LX hardware-software configurations.

Simulation Results

Noise-Free, Low-Frequency Profiles

In order to assess the feasibility of the technique under ideal conditions, simulations were performed. We acquired a homogeneous image of a brain, with a matrix size of 256×256 , using a head coil, then computed an ideal, noise-free sensitivity profile having a $1/r^2$ falloff and a linear phase profile. Its magnitude image is shown in Fig. 2a, and its phase profile is shown in Fig. 2b. Then, 16 rotated magnitude sensitivity profiles were computed with an angle of $2\pi/16$ between any adjacent two of them and given the same phase profile shown in Fig. 2b. Next, the computed sensitivity profiles were multiplied point by point with the brain magnitude image in order to get approximations of sensitivity weighted images from surface coils placed at different positions. The sensitivity-weighted images are then Fourier-transformed in two dimensions in order to get the k -space data. Subsequently, 16 lines of k -space data were taken from each matrix and used to reconstruct the head image. Image matrix size was chosen to be 256×256 .

As mentioned in the previous section, the choice of the k -space lines affects the image resolution. In Fig. 2d,e,f, we show three images reconstructed using different sets of

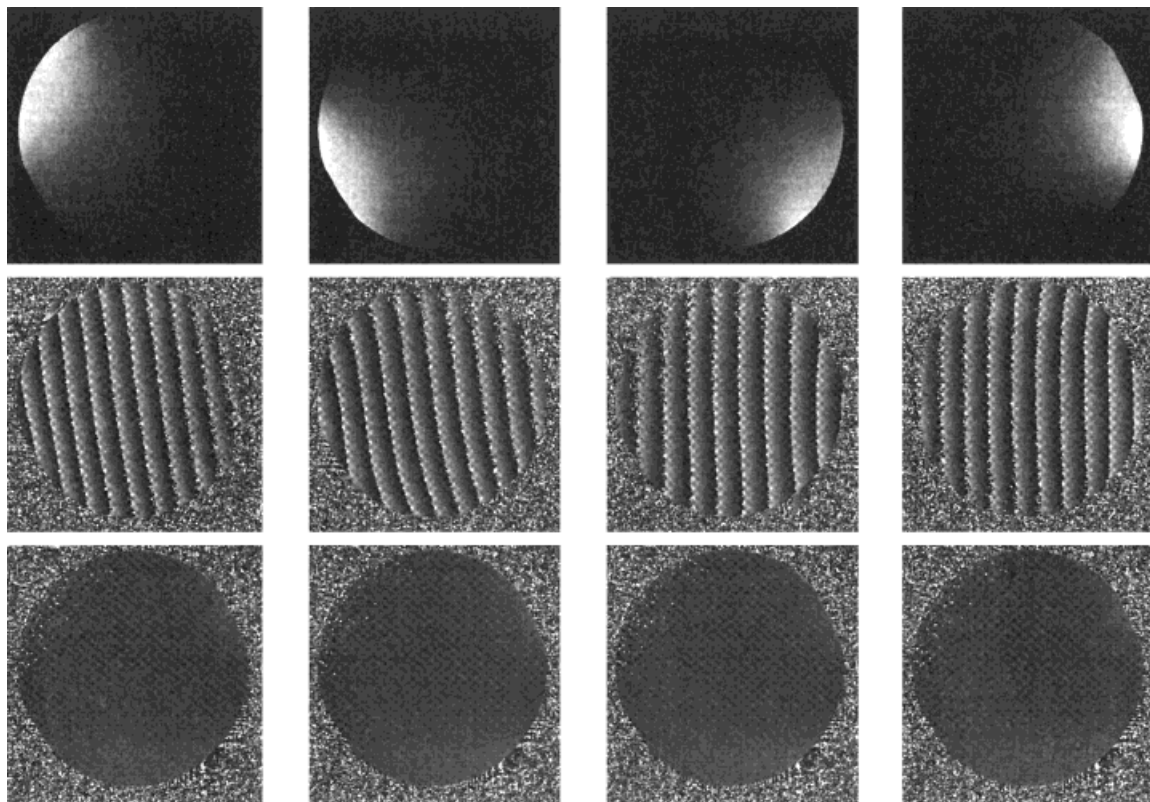


FIG. 3. Experiment results. The top row shows the magnitude images of the 2D sensitivity profiles as computed from a homogeneous water phantom from all four coils in the phased array. The middle row shows the phase images of the sensitivity profiles. The bottom row shows the adjusted phase profiles after linear shifting.

k -space data in the $K \times F = N$ case. These computations required the inversion of 256 matrices each of size 256×256 and were completed in less than 600 sec on a SUN UltraSparc machine with a processor speed of 266 MHz. These reconstructions can be performed in less than 1 sec if a processor is used for each column and with current processor speeds of over 700 MHz. This represents a clear advantage over the generalized SENSE reconstruction method, especially when the choice of the imaged slice is dynamically changed and a complete calculation of the pseudoinverses is needed. Reconstruction speed increases beyond real-time rates, however, when the same slice is being refreshed, and the calculation is reduced to a set of vector matrix multiplications.

Figure 2c shows the reference image reconstructed using a 2DFT on all k -space lines. Figure 2d shows a reconstruction using the 16 lines of k -space going between $k = -7$ and $k = 8$; Fig. 2e shows a reconstruction using the lines of k -space going between $k = -79$ to $k = 80$ and skipping 8 lines; finally, Fig. 2f shows a reconstruction using the 16 lines of k -space going between $k = -127$ to $k = 128$ and skipping 16 lines. It can be seen that Fig. 2e represents the best result of the three, as it shows better resolution than Fig. 2d and no artifact as in Fig. 2f. A certain deterioration of the image is observed in Fig. 2f as 16 lines are skipped before reconstruction. This suggests that the k -space coverage from such a subsampling is not appropriate.

Experimental Results

Two imaging experiments were done in order to test the performance of the technique in practical situations.

Imaging With the Cardiac Coil

In the first experiment we imaged a human head using a phased array cardiac coil with 4 elements of size 10×10 cm fixed around the FOV. Image matrix size was chosen to be 128×128 . First, a baseline FSE T_2 -weighted image was acquired from a homogeneous water phantom using the body coil with TR = 2 s, TE = 102 ms, and ETL = 12. The phased array coil was then used to simultaneously collect four images, one from each coil. These images were used in conjunction with the baseline image to calculate the sensitivity profiles of the coils at different positions using the point-by-point ratio method as described above. The magnitude images of the sensitivity profiles are shown in Fig. 3 (top) and their respective phase images are shown below them in Fig. 3 (middle). Initial calibration was performed in order to remove a linear phase shift which exists between the different coils. This calibration is necessary to insure that the k -space data in all coils is identically centered and that no destructive interference would arise during reconstruction. This finding is common to all parallel imaging techniques (6,7) where phase fidelity is crucial for the stability of the reconstructions. The resulting phase images of the adjusted profiles are shown in Fig. 3 (bottom).

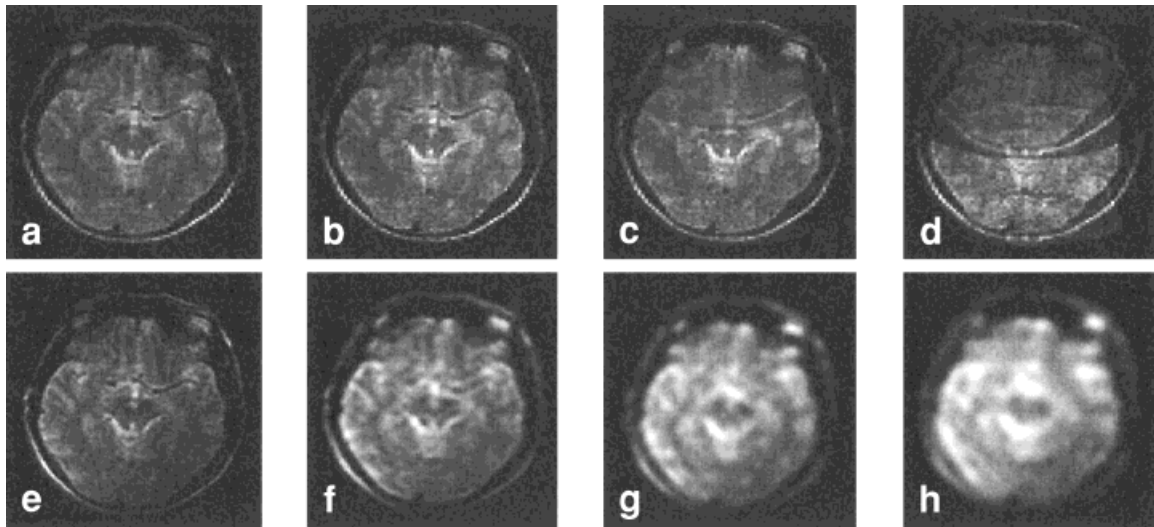


FIG. 4. Experiment results. The parts on the top show the reconstruction using our technique with the cardiac coil array, and choosing equally spaced k -space lines. The number of k -space lines used are, from left to right, respectively: 128, 64, 43, and 32. The parts shown on the bottom represent the images obtained by summing the magnitude of the images reconstructed using the middle lines of k -space in all four coils. The number of k -space lines used from left to right are equal to those of the corresponding top parts.

The calibrated profiles were subsequently used to encode a brain image as described previously. In a first reconstruction, we show in the top half of Fig. 4 the results of this technique, where equally spaced k -space lines were used. Figure 4a shows an image reconstructed using all 128 k -space lines. Figure 4b was reconstructed using 64 equally spaced and centered k -space lines; Fig. 4c was reconstructed using 43 k -space lines and Fig. 4d was reconstructed using 32 k -space lines. Each of these images is shown on top of an image reconstructed by using the same number of k -space lines in the keyhole mode (13). These images are shown, respectively, in Fig. 4e–h.

In a second reconstruction, we show the effect of choosing different sets of lines of k -space to enhance the rendering.

These results are shown in Fig. 5 where different sets of 32 lines of k -space are used to reconstruct a brain image. Below each image, we show the lines of k -space that were used to reconstruct it with our technique. The leftmost image is obtained by using 32 equally spaced lines in k -space whereby the consecutive images are reconstructed by using variations of denser coverages of the center of k -space. Reconstructions were performed on a Sun Ultrasparc station having a processor speed of 266 MHz. In all cases, reconstruction times were less than 160 sec.

Imaging With the Torso Coil

In order to test the effect of coil penetration on the reconstruction in practical situations, we repeated the previous

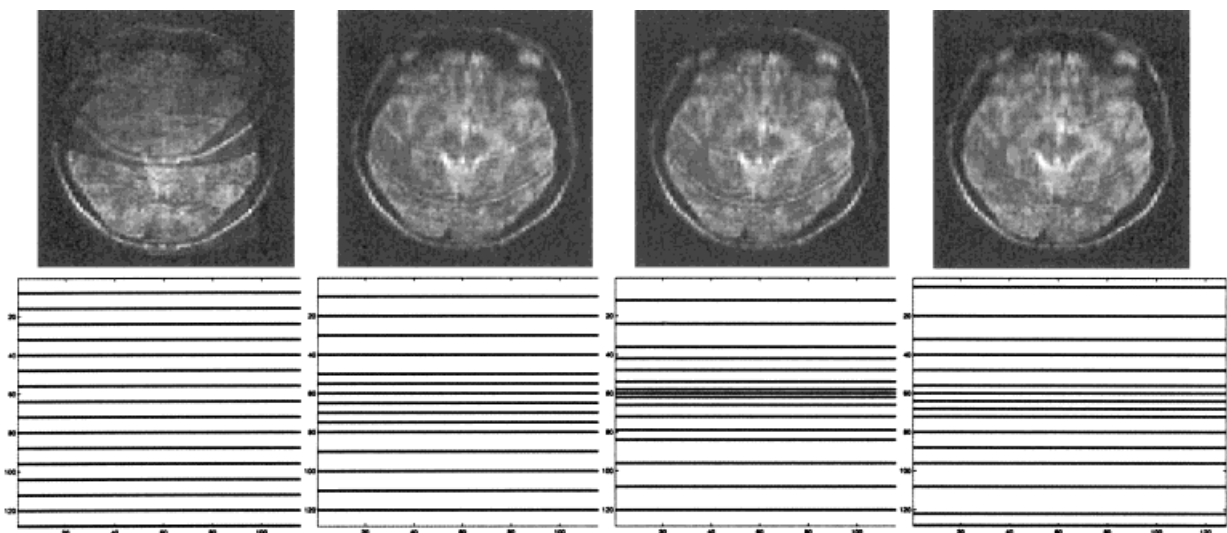


FIG. 5. Experiment results. The parts on the top show the reconstruction using different sets of 32 lines of k -space with our technique using the cardiac coil array. Below each part we show the corresponding k -space lines chosen.

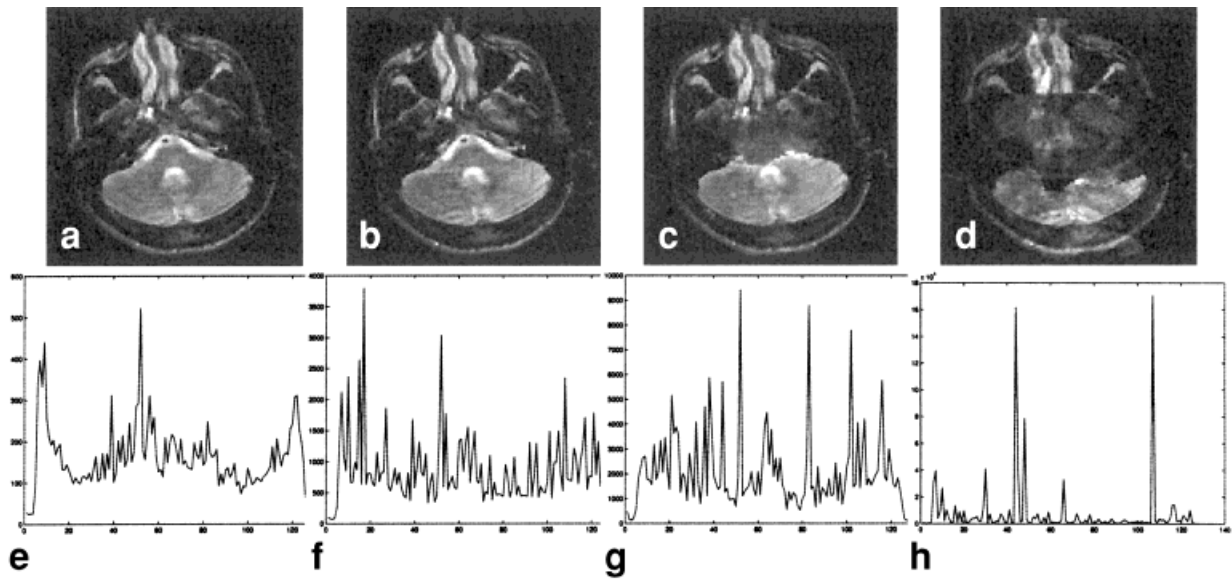


FIG. 6. Experiment results. The parts on the top show the reconstruction using our technique with the torso coil array and choosing equally spaced k -space lines. The number of k -space lines used are, from left to right, respectively: 128, 64, 43, and 32. The parts shown on the bottom represent the plot of the condition numbers of the reconstruction matrix for each column. The number of k -space lines used from left to right are equal to those of the corresponding top parts.

experiment (FSE, TR = 2 s, TE = 102 ms, ETL = 12) using the phased array torso coil with four elements of size (15 × 15 cm) fixed around the FOV. Image matrix size was chosen to be 128 × 128.

In a first reconstruction, we show in the top half of Fig. 6 the results of this technique, by using equally spaced k -space lines. Figure 6a shows an image reconstructed using all 128 k -space lines, Fig. 6b was reconstructed by using 64 k -space lines, Fig. 6c was reconstructed using 43 k -space lines, and Fig. 6d was reconstructed using 32 k -space lines. Each of these images is shown on top of a plot showing the magnitude of the condition number of the reconstruction matrix for every column in the image. It

can be seen that the condition number increases when fewer k -space lines are used, hence a slight overdetermination ($K \times F > N$) is recommended for better reconstructions. We also note that the reconstruction shown in Fig. 6c contains less artifact than Fig. 4c due to the larger coil elements.

In a second reconstruction, we show the effect of choosing different sets of lines of k -space to enhance the rendering. These results are shown in Fig. 7, where different sets of 32 lines of k -space are used to reconstruct a brain image. Below each image, we show the lines of k -space that were used to reconstruct it with our technique. The leftmost image is obtained by using 32 equally spaced

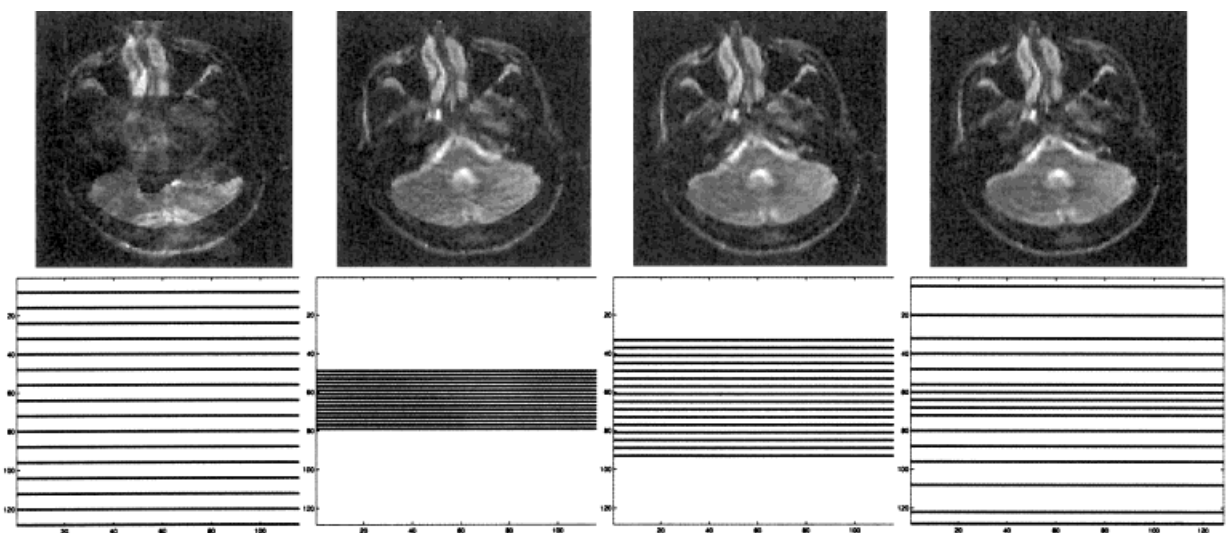


FIG. 7. Experiment results. The parts on the top show the reconstruction using different sets of 32 lines of k -space with our technique for the torso coil array. Below each part we show the corresponding k -space lines chosen.

lines in k -space whereby the consecutive images are reconstructed by using variations of denser coverages of the center of k -space. An overall improvement in image quality is seen compared to the images in Fig. 5, which we attribute to the greater penetration depth of the sensitivity profiles due to the larger coil elements.

DISCUSSION

We have designed and implemented a parallel imaging and reconstruction method intended to complement the SMASH and SENSE techniques. This technique is based on using 2D sensitivity profile maps from an arbitrarily placed array of RF receiver coils, in order to encode and reconstruct MRI data in parallel, therefore reducing imaging time proportionately to the number of coils present in the array. It is adaptable to any imaging sequence and to arbitrary imaging planes and has the flexibility of being able to use any set of k -space lines for image reconstruction.

The placement of the RF pickup coils can be arbitrary, around the imaged field of view. This allows for the placement of more RF coils than certain parallel imaging techniques such as SMASH (6). Having more coils in the receiver array would enable faster imaging.

A clear reduction of imaging speed could be achieved using this technique. In this article, we show the simulated results of a 16-fold speed increase and experimental results with a 4-fold speed increase. Since there is a tradeoff between coil size and penetration, an optimal number of coils can be determined to yield the best results.

The reconstruction in this method is based on a matrix inversion algorithm. In order to ensure a good reconstruction, our results show that it is necessary to minimize the condition number of the inverted matrices. Better conditioning of the reconstruction matrices can be achieved with the use of more k -space data from each coil ($K \times F > N$), as shown in Fig. 6e–h. Once the receiver coil and the number F of acquired k -space lines are chosen, conditioning is achieved by setting a threshold to the eigenvalues in the reconstruction matrix. This is done to minimize numerical propagation errors.

For a given number F of k -space lines used in the reconstruction, our results (Figs. 5, 7) show that the choice of the phase encodes is crucial to ensure that image resolution is similar to images acquired using full k -space data, and that artefacts due to the regular and insufficient sampling of k -space are eliminated. For all the images shown in Figs. 5 and 7, we found the condition numbers of the reconstruction matrices to be similar; however, the choice of a different set of k -space lines results in dramatically reducing the artifacts.

A current practical limitation of this technique is the need for a large number of receivers. Most commercial scanners are limited to four receivers, and the high cost of additional receivers makes having as many as 16 practically prohibitive. This problem could be solved with the use of channel multiplexers (5), whereby multiple coil outputs could be multiplexed and fed into a single receiver and later demultiplexed to get the output of each coil.

Our technique is based on the use of coil sensitivity profiles to encode spatial dimensions; therefore, a good

estimation of these profiles is crucial to ensure good results. In this article, we applied our technique by considering that the sensitivity profiles are independent of the coil loading, allowing us to extrapolate the results of the sensitivity profiles computed from a water phantom to encode other images. In reality, however, loading affects the sensitivity profiles of the coil elements and a dynamic method of profile computation would be useful to account for the different objects imaged.

This technique is sensitive to phase differences between coils, which should be accounted for in the reconstruction. Care should be taken to calibrate the data acquisition before the beginning of every imaging experiment.

The reconstruction algorithm presented in this technique allows for the independent computation of each column in the image. This technique lends itself, therefore, to parallel reconstruction, which can result in realistic image computation times while keeping the flexibility of choosing arbitrary cartesian k -space acquisition trajectories.

We believe that the limitations mentioned are surmountable, and the technique has great potential to overcome the problem of relatively long acquisition times in MRI. The parallel reconstruction algorithm should allow for real-time acquisition as well as reconstruction of MR data and may be very well suited for applications such as cardiac, functional, and interventional imaging.

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