Spatial Encoding Using Multiple rf Coils: SMASH Imaging and Parallel MRI

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1 INTRODUCTION

The speed of MR image acquisition has increased dramatically since the 1980s through a combination of technological and methodological advances. Nevertheless, many clinical applications of MRI continue to require motion compensation in some form. In body regions containing moving structures such as the heart or diaphragm, serious artifacts can arise if scan times exceed the characteristic time scales of physiologic motion. Accurate tracking of dynamic processes, such as cardiac contraction or the arrival and uptake of intravenous contrast agents, may require high temporal resolutions without undue sacrifices in spatial resolution. Meanwhile, the fastest imaging sequences and the most state-of-the-art scanners are approaching certain basic limits of imaging speed. These limits, which have both a technological and a physiological component, are related to the maximum switching rates of magnetic field gradients and rf pulses. Most of the fast imaging sequences now in use – echo planar imaging (EPI), fast low angle shot (FLASH), turbo spin echo (TSE), spiral, or BURST, for example – achieve their high speeds by optimizing the strengths, the switching rates, and the patterns of applied gradients and pulses. Beyond a certain threshold, however, rapidly switched field gradients are known to produce neuromuscular stimulation, while excessively dense rf pulse trains can lead to unacceptable levels of rf energy deposition and heating of tissue. One common feature of fast imaging sequences is that they acquire data in a sequential fashion. Regardless of the particular sequence the acquisition follows, the MR signal is always acquired one point and one line at a time, with each separate line of data requiring a separate application of field gradients and/or rf pulses. Therefore, imaging speed is generally limited by the maximum switching rates compatible with scanner technology and patient safety.

Recently, a new paradigm of parallel MRI has been used to increase imaging speeds beyond the basic limits just described. The term ‘parallel MRI’ may be used to describe any MRI strategy in which multiple MR signal data points are acquired simultaneously, rather than one after the other. Parallel imaging strategies in general require the use of multiple distinct detectors, with each detector providing some component of distinct spatial information to the image. (A many-detector CCD camera is a familiar optical example of a parallel imaging device while a FAX machine is an example of a sequential line-scanning device.) For MRI in particular, some of the burden of spatial encoding traditionally accomplished by field gradients may be shifted instead to arrays of rf coils. Recent work has shown that coil arrays may be used, in combination with appropriate image reconstruction strategies, to encode and detect multiple MR signal or image components simultaneously and thereby to multiply the speed of existing imaging sequences without increasing gradient switching rate or rf power deposition.

2 THE HISTORY OF PARALLEL MRI

Radiofrequency coil arrays were first developed for use in MRI in the late 1980s. Prior to that time, single volume or surface coils were typically used for signal detection, or else pairs of coils were arranged in quadrature to increase signal-to-noise ratio (SNR). Work by Roemer and co-workers demonstrated that arrays of suitably decoupled surface coils could be used to achieve further substantial SNR increases (see Whole Body Machines: NMR Phased Array Coil Systems). When used in combination with traditional gradient-encoding sequences, coil arrays increased the achievable SNR for any given field of view (FOV) in any given imaging time, but ultimate imaging speed was still governed by the imaging sequence and the gradient hardware.

The fact that spatial information from coil arrays might also be used directly for the encoding and decoding of images was realized in principle relatively early in the development of MRI-compatible arrays. The simplest manifestation of the parallel imaging principle was the use of spatially separated coils to image distant and nonoverlapping body regions simultaneously. More challenging was the prospect of imaging a continuous FOV rapidly using detectors overlapping in space and/or sensitivity. An early theoretical proposal by Carlson suggested that the Fourier coefficients of signal voltages in multiple coils disposed around a cylinder could be used to calculate multiple $k$-space lines for magnetization within that cylinder. Hutchinson and Raff suggested in 1988 that a large array of narrow loops surrounding an object could, in principle, be used to acquire an image without the use of any phase-encoding gradients at all. Other proposals followed, and these proposals fall into two general categories: massively parallel and partially parallel strategies.

In massively parallel strategies, the number of detectors approaches the number of data points or lines. These techniques aim to replace gradient encoding entirely, with resulting dramatic improvements in imaging speed. The original massively parallel proposal of Hutchinson and Raff was intended to show theoretical feasibility, and no direct practical implementation was suggested. A subsequent proposal for massively parallel imaging was made by Kwiat, Einav, and Navon in 1991. Their study included a more detailed investigation of practical issues, though no images were presented. A preliminary array design based on the general principles of this proposal was presented in 1995. In partially parallel imaging strategies, the number of detectors is significantly smaller than the number of data points. Spatial encoding using multiple rf coils is used to supplement the spatial encoding normally accomplished using magnetic field gradients. Kelton, Magin, and Wright, and later Ra and Rim, described a ‘subencoding’ approach by which aliased
component coil images acquired rapidly with reduced phase encoding may be ‘unaliased’ using information about component coil sensitivities. Carlson and Minemura proposed using nested volume coils with differing sensitivity patterns to approximate multiple k-space lines from a series expansion. In 1997, Simultaneous Acquisition of Spatial Harmonics (SMASH) was introduced, and the first accelerated in vivo MR images using a parallel imaging strategy were obtained. The SMASH technique uses linear combinations of component coil signals from a surface coil array to replace time-consuming gradient steps directly. Following the introduction of SMASH, the subencoding principle was revisited and refined in the SENSitivity Encoding technique (SENSE) of Prüssmann, Weiger, and co-workers, which has also been used recently to obtain accelerated in vivo images. At the present time, various other research groups both in academia and in industry have also begun to investigate parallel imaging strategies.

Table 1 lists the various proposed parallel MRI techniques, further categorized by coil sensitivity calibration and image reconstruction method. The principal message of this table is that there are numerous ways to extract spatial information from an array of rf detectors. We will now discuss some of these methods in greater detail, emphasizing approaches for which in vivo results have already been published, and which are likely to be well suited for clinical MRI. We will begin with the SMASH technique, for which the broadest range of in vivo results have been achieved to date. We will then discuss the image-domain subencoding reconstruction techniques.

3 SMASH IMAGING

3.1 Theory

The SMASH technique exploits sensitivity variations in a surface coil array to substitute for spatial modulations normally produced by phase-encoding gradients. This use of coil encoding in place of gradient encoding allows the whole of k-space to be traversed using a reduced number of phase-encoding gradient steps, thereby reducing image acquisition times.

The function of phase-encoding gradients is to impose sinusoidal modulations of magnetization across the image plane. The MR signal integrated against these sinusoids then corresponds to spatial Fourier components of the image, or the familiar k-space lines. Figure 1 illustrates schematically this well-known effect. In the figure, sinusoidal modulations of varying spatial frequency, resulting from spin evolution in phase-encoding gradients of varying strength, are shown next to their associated k-space lines. In the SMASH technique, some of these sinusoidal modulations, or ‘spatial harmonics’, are generated by manipulations of component coil sensitivities, rather than by gradient-induced modulations of magnetization.

An array of rf coils contains spatial information in the form of its component coil sensitivities (Figure 2). In a linear surface coil array with adjacent components, each coil j has a distinct but overlapping sensitivity Cj(x, y). By forming appropriate linear combinations of component coil signals (Figure 3), we may generate composite sensitivity profiles Ctot which oscillate in

Table 1 Parallel MRI techniques

<table>
<thead>
<tr>
<th>Technique</th>
<th>Sensitivity calibration</th>
<th>Image reconstruction</th>
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<tbody>
<tr>
<td>Partially parallel</td>
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<tr>
<td>k-space techniques</td>
<td>Known volume coil sensitivities</td>
<td>k-space series expansion</td>
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<tr>
<td>Carlson and Minemura (1993)</td>
<td>Surface coil sensitivity reference</td>
<td>k-space linear combination (spatial harmonics)</td>
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<tr>
<td>SMASH (1996)</td>
<td>(phantom, in vivo, AUTO-SMASH)</td>
<td></td>
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<tr>
<td>Image domain techniques</td>
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<tr>
<td>SENSE (1997)</td>
<td>Pixel-by-pixel sensitivity extraction from full reference images</td>
<td>Pixel-by-pixel matrix inversion (subencoding)</td>
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<tr>
<td>Massively parallel</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hutchinson and Raff (1988)</td>
<td>Not discussed</td>
<td>Inverse source</td>
</tr>
<tr>
<td>Kwiat, Einav, and Navon (1991)</td>
<td>Point source references</td>
<td>Inverse source</td>
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SMASH, simultaneous acquisition of spatial harmonics; SENSE, sensitivity encoding technique.

aIn vivo results have been published for this technique.

For list of General Abbreviations see end-papers
much the same way as the gradient-induced modulations of Figure 1:

\[
C^{\text{tot}}(x, y) = \sum_j n_j C_j(x, y) \approx \exp(i m k_j y)
\]

where \( n_j \) are complex weight factors, \( m \) is an integer, and \( k_x = 2\pi/\text{FOV} \) is the minimum \( k \)-space interval corresponding to the desired FOV.

If a composite sensitivity profile generated in this way forms an accurate spatial harmonic pattern, the same \( k \)-space step is produced as would have resulted from a traditional gradient step. In other words, each combined signal \( S^{\text{tot}} \), generated from linear combinations of component coil signals \( S_j \) using the weights \( n_j \) from Equation (1), is shifted in \( k \)-space by an amount \((-m\Delta k_y)\):

\[
S_j(k_x, k_y) = \int \int d x \, d y C_j(x, y) \rho(x, y) \exp\{-i k_x x - i k_y y\}
\]

\[
S^{\text{tot}}(k_x, k_y) = \sum_j n_j S_j(k_x, k_y)
\]

\[
= \int \int d x \, d y \sum_j n_j C_j(x, y) \rho(x, y) \exp\{-i k_x x - i k_y y\}
\]

\[
= \int \int d x \, d y C^{\text{tot}}(x, y) \rho(x, y) \exp\{-i k_x x - i k_y y\}
\]

\[
\approx \int \int d x \, d y \rho(x, y) \exp\{-i k_x x - i (k_y - m\Delta k_y) y\}
\]

\[
\approx \tilde{\rho}(k_x, k_y - m\Delta k_y)
\]
called AUTO-SMASH, may be used for imaging in regions of heterogeneous spin density such as the thorax, where large differences in signal between the heart and lungs preclude straightforward estimation of rf sensitivity. In AUTO-SMASH, a small number of reference k-space lines are added to the acquisition, and the relation between these reference lines and the usual MR signal data lines are used to ‘train’ SMASH reconstructions directly in k-space.

Following sensitivity calibration, the sensitivity profiles are fitted to the desired spatial harmonic functions, using a numerical optimization algorithm with the complex weight factors $n_j$ as fitting parameters. For favorable array geometries, the resulting fits can, in practice, be quite as good as the schematic fits shown in Figure 3.

The remaining steps in the SMASH reconstruction procedure are summarized in Figure 5. The left-hand side of Figure 5 shows a k-space schematic, and the right-hand side shows image data from a water phantom at each of the corresponding stages of reconstruction. With the necessary weights in hand, MR signal data are acquired simultaneously in the coils of the array. A fraction $1/M$ of the usual number of phase-encoding steps are applied, with $M$ times the usual spacing in k-space (Figure 5A, left). The component coil signals acquired in this way correspond to images with a fraction $1/M$ of the desired FOV (Figure 5A, right). With $1/M$ times fewer phase-encoding steps, only a fraction $1/M$ of the time usually required for this FOV is spent on data collection.

Next, the appropriate $M$ linear combinations of the component coil signals are formed to produce $M$ shifted composite signal data sets (Figure 5B). The composite signals are then interleaved to yield the full k-space matrix (Figure 5C, left), which is Fourier transformed to give the reconstructed image (Figure 5C, right).

The schematic summary in Figure 5 shows a SMASH reconstruction with acceleration factor $M=2$ using a 3-element rf coil array. Substantially larger factors are possible, however, when coil arrays with larger numbers of elements are used. In fact, for favorable image plane and coil array geometries, the maximum achievable SMASH acceleration factor $M$ is equal to the number of independent component coils in the array, since a maximum of $M$ distinct harmonics may be generated using a total of $M$ independent coils. Since generation of spatial harmonics does not depend upon how the gradient-encoded k-space lines were generated, the SMASH reconstruction is to a large extent sequence independent. Nearly all existing rapid imaging sequences may be accelerated in this manner, and, to date, SMASH has been successfully tested with a wide range of sequence types. Both two-dimensional and three-dimensional acquisitions are amenable to acceleration using SMASH, provided distinct coil sensitivity information is available along one or more phase-encoding directions.

3.3 In Vivo Results

The improvements in imaging efficiency afforded by a parallel imaging strategy may be put to use in a number of ways. The following examples demonstrate some of the applications for which SMASH imaging has been used to increase imaging speed and improve image quality.

Reductions in breath-hold duration for scans requiring breath-holding can be achieved, which increases patient comfort and compliance and allows scans free of respiratory motion artifact in patients incapable of prolonged breath-holds. Figure 6 demonstrates a twofold reduction in breath-hold duration for abdominal MR imaging.

Improvements in spatial resolution can be achieved in any given imaging time: images of increased spatial resolution may
be generated in a given acquisition time by carrying the faster SMASH acquisition farther out in k-space. Figure 7 shows the use of SMASH for spatial resolution enhancement in a cardiac scan. Additional resolution benefits can result from the use of SMASH or other partially parallel techniques in single-shot imaging sequences such as HASTE (half-Fourier single-shot turbo spin echo), EPI, or BURST. Indeed, single-shot images with both reduced acquisition time and increased spatial resolution compared with the corresponding reference images have been obtained. This is possible because a reduced acquisition time also entails reduced relaxation, and hence reduced attenuation of high spatial frequencies in single-shot sequences.

Improvements in temporal resolution (i.e., reductions in image-acquisition interval for gated or ungated scans) minimize undesired effects of physiologic motion while allowing accurate tracking of time-dependent phenomena. Figure 8 illustrates a twofold and a fourfold reduction in acquisition window at fixed spatial resolution in cardiac MR images. These progressive increases in temporal resolution result in progressively reduced motion-related blurring of the right coronary artery and other cardiac structures in the SMASH images. SMASH has also been used to increase true frame rate in real-time cardiac MR scans up to and beyond current two-dimensional echocardiographic frame rates.

Reductions in the overall duration of long MR scans increase patient comfort and compliance and also increase the throughput of clinical MR scanners and the cost-effectiveness of MR diagnosis. Noncontrast MR coronary angiograms obtained using SMASH are shown in Figure 9. Reduction in overall acquisition time in these navigator-gated scans has the further advantage of reducing long-term diaphragmatic drift over the course of the scans.

4 IMAGE-DOMAIN SUBENCODING TECHNIQUES

4.1 Theory

A subencoding image reconstruction begins at the same starting point as a SMASH reconstruction – namely, with a set of component coil signals acquired using a reduced number of phase-encoding gradient steps. Fourier transformation of these signal sets results in aliased component coil images like those shown in Figure 5A. From that point on, the subencoding reconstruction operates entirely in the image domain.

The basis of the technique lies in the fact that each pixel in an aliased image is in fact a superposition of multiple pixels.

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from a corresponding full unaliased image (Figure 10). In other words, as a result of Nyquist aliasing, an $M$-times aliased image $I^{\text{fold}}$ is related to the full image $I^{\text{full}}$ as follows:

$$I^{\text{fold}}(x, y) = I^{\text{full}}(x, y) + I^{\text{full}}(x, y + \Delta y) + I^{\text{full}}(x, y + 2\Delta y) + \ldots = \sum_{m=0}^{M-1} I^{\text{full}}(x, y + m\Delta y)$$

(3)

When $I^{\text{fold}}$ is acquired using a single coil, this superposition cannot be ‘unfolded’ without a priori knowledge of the full image.

The situation changes when an array of coils is used. The full image $I^{\text{full}}_j$ in each coil $j$ is actually made up of two pieces: the spin density $\rho$, and the coil sensitivity function $C_j$:

$$I^{\text{full}}_j(x, y) = C_j(x, y)\rho(x, y)$$

(4)

and in an array, each component coil $j$ has a different sensitivity $C_j$. Therefore, we now have multiple ‘views’ of the aliasing that can be used to deduce just how much of each aliased pixel belongs at any position in the full image. Substituting Equation (4) into Equation (3) gives

$$I^{\text{fold}}_j(x, y) = \sum_{m=0}^{M-1} I^{\text{full}}_j(x, y + m\Delta y)$$

(5)

For any particular aliased pixel $(x,y)$, this may be written as follows:

$$I^{\text{fold}}_j = \sum_{m=0}^{M-1} I^{\text{full}}_{jm} = \sum_{m=0}^{M-1} C_{jm}\rho_m$$

(6)

where $I^{\text{full}}_{jm} \equiv I^{\text{full}}_j(x,y + m\Delta y)$, $C_{jm} \equiv C_j(x,y + m\Delta y)$, and $\rho_m \equiv \rho(x,y + m\Delta y)$.

Let us study the particular example (illustrated in Figure 11) in which a four-coil array is used with a factor of three aliasing. We may write

For list of General Abbreviations see end-papers
This equation may be rewritten in matrix form as

\[
\begin{bmatrix}
I_{1}^{\text{fold}}
I_{2}^{\text{fold}}
I_{3}^{\text{fold}}
I_{4}^{\text{fold}}
\end{bmatrix} =
\begin{bmatrix}
C_{11} & C_{12} & C_{13}
C_{21} & C_{22} & C_{23}
C_{31} & C_{32} & C_{33}
C_{41} & C_{42} & C_{43}
\end{bmatrix}
\begin{bmatrix}
\rho_{1}
\rho_{2}
\rho_{3}
\end{bmatrix}
\]

or, in other words,

\[I^{\text{fold}} = C\rho\]  \hspace{1cm} (7)

As long as the number of coils \(N_{c}\) is greater than or equal to the aliasing factor \(M\) (as in our exemplary case for which \(N_{c} = 4, M = 3\)), Equation (9) may be inverted:

\[
\rho = C^{-1}I^{\text{fold}}
\]

and the unaliased spin density over the full FOV may be determined. This ‘unaliasing’ approach bears some kinship with the principle of computed tomography, in the sense that multiple different ‘views’ or projections are used to extract full two-dimensional image information.

4.2 Implementation

Subencoding implementations differ from SMASH implementations primarily in their approaches to sensitivity calibration and in the nature and numerical stability of their image reconstruction algorithms. Sensitivity estimates at each pixel of the full FOV are generally required to perform the pixel-by-pixel subencoding matrix inversion of Equation (10). Ra and Rim describe the use of an in vivo sensitivity reference in the form of full-FOV component coil images of the target image plane, manipulated in the reconstruction in such a way that the spin density divides out. The SENSE technique incorporates a different in vivo sensitivity calibration method in which full-FOV component coil images are divided by an additional full-FOV body coil image, and the quotient images are then subjected to several stages of interpolation, filtering, and thresholding. Phantom sensitivity references can also, in principle, be used for subencoding reconstructions.

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Whereas the quality of SMASH reconstructions is governed by the ‘goodness’ of spatial harmonic fits, the principal algorithmic concern of subencoding reconstructions is the numerical stability of the inverse $C^{-1}$. For $N_c > M$, the inverse may be implemented, for example, as a Moore–Penrose pseudoinverse, resulting in a least squares solution to the overdetermined problem for each pixel. For $N_c = M$, a standard matrix inverse may be used. The use of a pixel-by-pixel inversion affords the advantage of fine regional control over the reconstruction: it does not, for example, require a global spatial harmonic fit. A disadvantage of the pixel-by-pixel approach, however, is that in regions of low actual or apparent coil sensitivity, the matrix $C$ may be poorly conditioned, and error propagation through the inverse may amplify the effects both of noise and of sensitivity miscalibrations.

4.3 In Vivo Results

Image-domain subencoding techniques can be used for many of the same applications as were demonstrated with SMASH in Section 3.3. The resulting images will resemble accelerated SMASH images, with some particular differences in image artifacts and noise distribution resulting from the different approaches to sensitivity calibration and image reconstruction. In vivo SENSE images in the brain and the heart with two- to threefold accelerations have been presented.\(^\text{12,13}\) An in vivo sensitivity calibration method similar to that suggested by Ra and Rim has also been used by the SMASH group to obtain accelerated subencoding images for direct comparison with SMASH reconstructions of the same data sets. Figures 12 and 13 compare reference, SMASH, and subencoding images from contrast-enhanced angiography and real-time cardiac MR imaging studies.

5 A RECIPE FOR PRACTICAL PARALLEL IMAGING

The following is a list of essential elements needed for effective implementations of partially parallel imaging strategies such as SMASH or subencoding.

1. \textit{rf coil array}. Arrays should be designed with overall dimensions appropriate to the desired FOV for imaging applications of interest. Inductive decoupling of array elements is helpful for improved SNR and for increased spatial selectivity of the sensitivity profiles, although some degree of coupling between array elements can generally be tolerated. The imaging system must be equipped with a sufficient number of receiver channels to allow simultaneous data acquisition in the various component coils of the array. Alternative data-reception schemes such as time-domain multiplexing may also be used.\(^\text{16,17}\)

2. \textit{Sensitivity calibration}. Accurate coil sensitivity calibration is crucial for all parallel imaging techniques. A great deal of effort has been devoted over the years to the understanding and calibration of gradient waveforms in MR scanners. Since parallel imaging strategies use rf coils in place of gradients for spatial encoding, it is not surprising that some
degree of analogous effort must go into the calibration of coil sensitivities.

3. Scan planning. After a target image plane and FOV have been selected, all that is required on most MR scanners to plan a SMASH or subencoding scan is a simple FOV reduction in the phase-encode direction. The need to perform spatial encoding with the rf coil array places some limitations on the choice of image plane position and orientation. For example, phase encoding must be performed along a direction in which the sensitivities of different array elements are sufficiently distinct. Substantial angulations between the image plane and the coil array are often possible (see, for example, the double-oblique images in Figures 9 and 13), though some penalty in SNR may be incurred.

4. Image reconstruction. Signal and image processing requirements are modest, but some capabilities for variable component coil signal combinations are required, whether in the MR scanner itself or in offline processors with access to the raw MR signal data. Details of software and/or hardware implementations will depend upon which parallel reconstruction strategy is elected.

Occurring as they do on either side of the Fourier transform, the k-space and the image-domain reconstructions bear a theoretical kinship, but their practical specifications differ. The choice of reconstruction strategy may be influenced by a number of factors:

(a) Ease of implementation. SMASH uses a small number of weight factors (a minimum of one per component coil per spatial harmonic), and operates in k-space prior to Fourier transformation. Consequently, it is particularly amenable to in-line implementations in which the coil-encoded k-space data are generated as soon as each gradient-encoded data point is read out. By contrast, the large numbers of weight factors in subencoding (one per component coil per image pixel) provide finer pixel-by-pixel control at the expense of increased calibration and reconstruction time. In the future, compromises between these two extremes may be anticipated, using various extrapolation schemes.

(b) Image artifacts. In SMASH, residual aliasing artifacts result from errors in sensitivity calibration or spatial harmonic fitting. In subencoding images, localized pixel-by-pixel artifacts are more commonly seen as a result of pixel-by-pixel errors in the coil sensitivity references. (These local artifacts may be accompanied by global aliasing artifacts when there are systematic errors in sensitivity calibration.) In some circumstances, however, both techniques can actually lead to artifact reduction. Not only can accelerated acquisitions reduce motion artifacts, but they have also been shown to reduce geometrical distortions in single-shot SMASH EPI images through reduced

![Figure 9](image1.png)

**Figure 9** Noncontrast MR coronary angiograms obtained in reduced total scan time using SMASH (same basic sequence, coil array, and MR scanner as in Figure 8, with an acquisition window of 70 ms and an in-plane resolution of 0.7 mm × 1.0 mm). Double oblique image planes were planned along the major axis of the right coronary artery (RCA). (A) A 7.5 cm length of the native RCA in a healthy adult volunteer is seen in this high-resolution image. The 3D data set from which this image was taken was obtained during free breathing in a total of 11 min using a SMASH acceleration factor of two. (B) The RCA is interrupted by a region of signal dropout from a coronary stent in this local maximum intensity projection of a twofold-accelerated 3D SMASH image set (11 min free-breathing acquisition) in a patient with coronary artery disease.

![Figure 10](image2.png)

**Figure 10** Superposition of pixels (white squares) in an aliased image
accumulation of phase discrepancies in the shortened acquisition time.

(c) SNR. The SNR in a partially parallel image reconstruction is a balance between coil-specific, reconstruction-specific, and sequence-specific effects. Theoretical and experimental studies of SNR in SMASH imaging have been reported.\textsuperscript{18} Generally speaking, both SMASH and subencoding techniques are bound by the well-known limit that SNR scales as the square root of the acquisition time for any given coil and sequence. With either class of technique, then, there is some expected loss of SNR compared with optimal combinations\textsuperscript{1} of traditionally acquired full data sets in the same array, though parallel acquisitions may still show improved SNR when compared with sequential acquisitions using single surface coils spanning the same FOV. For certain imaging sequences, furthermore, some additional SNR beyond the square root limit may be recovered through reduced relaxation in the shortened acquisition times.

6 THE FUTURE OF PARALLEL MRI

Since the initial demonstration in 1997 of twofold-accelerated in vivo images using SMASH, new in vivo work has shown up to fourfold increases in acquisition speed, and as much as eightfold improvements have been achieved in phantoms using specialized rf hardware.\textsuperscript{17} What future progress may be expected, and what are the key technical and methodological hurdles to further development?

Coil array design and rf sensitivity calibration top the list of developments that will be crucial for future improvements in parallel imaging, since array geometry and sensitivity characteristics determine the degree of spatial encoding which may be accomplished in any particular region of interest. The design of tailored arrays and the availability of increased numbers of receiver channels will greatly facilitate the optimization of image quality and SNR, and the maximization of achievable acceleration factors. In-line reconstruction hardware will be of benefit for real-time imaging applications. When the necessary weight factors are known in advance, the use of onboard analog signal combiners prior to digitization may even allow large ‘smart’ arrays to be interfaced to existing MR scanners with limited numbers of receivers. Methodological advances further in the future may also allow a return to the principle of massively parallel imaging. The potential gains of massively parallel imaging approaches are dramatic (they include the possibility of acquiring an entire image in a single echo, for example), but serious technological and theoretical hurdles remain to be overcome in areas such as electric decoupling of large arrays, massively parallel data reception, and SNR.

For list of General Abbreviations see end-papers
In the meantime, clinical implementation of partially parallel imaging is possible now using existing arrays and receiver systems. Indeed, clinical studies in selected patient populations are now being initiated using parallel acquisition techniques. Particularly for applications with stringent requirements on imaging speed, parallel imaging can be a useful tool to enhance image quality, to improve imaging efficiency, and, in general, to overcome the acquisition speed limit in magnetic resonance imaging.

7 RELATED ARTICLES

Image Formation Methods; Radiofrequency Systems and Coils for MRI and MRS; Spin Warp Data Acquisition; Surface and Other Local Coils for In Vivo Studies; Whole Body Machines: NMR Phased Array Coil Systems.

8 REFERENCES


Biographical Sketch

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