This method of tissue-signal nulling can also be applied with trueFISP to demonstrate the late enhancement of the infarcted myocardium [15.55]. The advantage of trueFISP over turboFLASH is that the additional signal contributions due to refocusing and balancing (spin-echo components) allow for faster image acquisition (~450 ms for a total image). Moreover, trueFISP exhibits significantly lower sensitivity to flow and motion artifacts compared to turboFLASH, thus enabling virtually artifact-free images.

The introduction of short T_R gradient-echo acquisition schemes made 3D imaging feasible. Applying an inversion pulse prior to a 3D acquisition scheme would not represent a promising approach because the preparation of the longitudinal magnetization would rapidly diminish during the relatively long measurement time and on account of the significant number of low angle excitation pulses. A feasible alternative is to repeat the preparation of the longitudinal magnetization either in the partition-encoding loop or in the phase-encoding loop. Placing the preparation scheme in the partition-encoding loop eliminates the above-mentioned TFL artifact associated with the over and under representation of *k*-space lines. For a selected phase-encoding gradient amplitude the inversion pulse is applied followed by the rapid execution of the partition-encoding loop – during which the amount of longitudinal magnetization will change according to the course of the T_1 relaxation, with recovery being influenced by the low angle excitation pulses. This procedure is repeated for the next phase-encoding line. The amount of signal within each partition is identical for all phase-encoding steps and the *k*-space (in-plane) is again symmetric – resulting in artifact-free images. This technique has been introduced as magnetization prepared rapid gradient echo (MP RAGE) [15.56].

The MP RAGE sequence allows for seamless coverage of the entire brain in less than 6 minutes. However, susceptibility gradients (especially at the base of the skull) cause a geometrically distorted representation of the anatomy or even signal voids. Furthermore, MP RAGE does not always permit the reproduction of the commonly observed increase in signal intensity in the case of tumors upon the uptake of a paramagnetic contrast agent [15.57]. It thus follows that this sequence cannot yet replace standard T_1-weighted SE sequences despite its fast image acquisition time.

Diffusion-weighted EPI

As has been discussed in section 7.2.8, spin diffusion is one further parameter that can be used to characterize biological tissue. Diffusion-weighted and diffusion-tensor imaging represent additional forms of magnetization preparation. These will be discussed in greater detail in a later section. Magnetization is prepared prior to an EPI sequence using a 90° and a 180° pulse (or three 90° pulses) with diffusion gradients before and after the refocusing pulse. The echo is read out using an EPI gradient switching scheme. Diffusion-weighted imaging enables one to evaluate the extent of cerebral ischemia due to infarction (see section 15.2.5.2, fig. 15.96).

15.2.1.5 Hybride methods

Following the success of fast spin echo imaging, a variant of the technique was introduced so as to enable the acquisition of multiple phase-encoded gradient echoes in a

spin-echo envelope. This technique is referred to the term gradient and spin-echo imaging (GRASE [15.58]) or turbo gradient and spin-echo imaging (TGSE). The potential reduction in measurement time scales with the number of gradient echoes used per spin-echo envelope, although the echo train length increases due to the additional read-out modules, and thus prolongs the slice-loop time or the minimum T_R. The spacing of the RF refocusing pulses is usually adjusted such that the echoes of fat oscillating due to homonuclear coupling (J-coupling) appear at intensities similar to a SE sequence. The introduction of gradient echoes has led one to anticipate an increase in sensitivity to susceptibility gradients and thus to an increase in sensitivity to hemorrhagic lesions [15.59].

15.2.2 Parallel imaging techniques (PAT)

Parallel imaging techniques (PAT) emerged as a method of parallelizing the sequential character of conventional MR image acquisition schemes in order to shorten imaging times. Despite the introduction of high performance gradients, RF electronics and sophisticated pulse sequences for many applications, scan time has remained limited by the time consuming sequential (line-by-line) acquisition of individual Fourier lines. The PAT concept involves the partial parallelization of the phase encoding process in order to substantially shorten conventional pulse sequences.

In addition to shorter scan times, the benefits of PAT include the ability to significantly increase image quality for certain applications. For instance, PAT may be applied to shorten the echo train length of EPI protocols, resulting in a reduction in susceptibility-induced distortions. This example also demonstrates the significance of the technique in general: it is hard to imagine that a similar improvement could be achieved by deploying more powerful gradients. Apart from the technical constraints (a factor of two in readout speed would require twice the gradient amplitude and a fourfold gradient slew rate!), patient peripheral nerve stimulation imposes a physiological limit on the applied gradient power.

PAT is not restricted to special sequence types. In general, it can be applied to any imaging protocol, a property whose significance should not be underestimated when it comes to clinical application given the fact that the contrast behavior of established protocols is preserved.

15.2.2.1 Basics of parallel imaging

Although the principles of PAT were first outlined in the 1980s [15.60-15.65], the technology was not implemented for clinical imaging until the late 90s. The basic idea behind PAT is to make use of the spatial information contained in the sensitivity profiles of local receiver coils. Array coils were originally introduced as dedicated signal reception devices designed to improve *SNR* in MRI [15.7]. In the case of PAT, they are incorporated into the spatial encoding process, which is conventionally based solely on RF excitation and gradient pulse schemes.

It is an intrinsic property of local coils that their sensitivity profiles vary in space, both in magnitude and phase. Hence, the detected signal is modulated with a spatially vary-

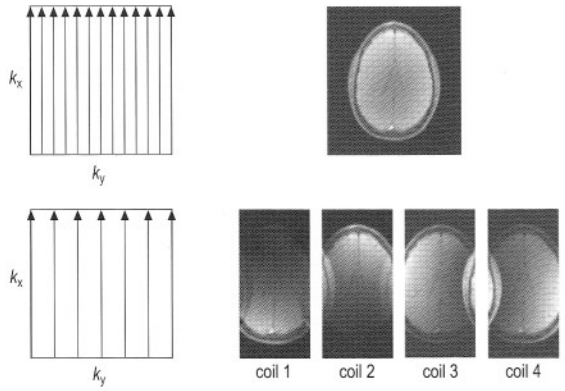


Figure 15.54
Typical parallel acquisition scheme. **Upper row:** non-accelerated, fully sampled *k*-space **(left)** and reconstructed image **(right)**. **Lower row:** accelerated acquisition via reduced phase encoding **(left)** leads to folded raw images **(right)**, in this example received by four individual coil elements.

ing function determined by the coil geometry. If a sample is covered by multiple, independently receiving coils, the spatial information contained in each receiver channel can provide part of the spatial encoding required for image reconstruction.

A typical parallel imaging scan will periodically leave out phase encoding steps, for instance, by skipping every other phase encoding line in *k*-space (fig. 15.54). These missing lines are calculated during image reconstruction with the use of the information contained in the scanned Fourier lines of the individual receiver channels.

Despite their sheer number, most of the parallel acquisition and reconstruction techniques that have been proposed [15.66-15.72] can be broken down into *k*-space-related methods and image-domain-related methods. These techniques are commonly referred to by the acronyms SMASH and SENSE that were coined by the authors of the first publications showing clinical results. In what follows, we offer a brief description of the techniques.

15.2.2.2 The SMASH technique

Introduced in 1997, the SMASH technique (simultaneous acquisition of spatial harmonics) was the first to incorporate processes of parallel acquisition into clinical MR imaging.

SMASH incorporates the coil sensitivity patterns $\sigma_l(x,y)$ into the spatial encoding function. As has been shown (section 7.3.1), the signal intensity in *k*-space $s_l(k_x,k_y)$ from the coil element *l* can be written as

$$s_l(k_x,k_y) = \iint \rho(x,y) \cdot \sigma_l(x,y) \cdot e^{i(k_x x + k_y y)} dx dy \qquad (15.41)$$

$\rho(x,y)$: Spin density, $\sigma_l(x,y)$: Sensitivity profile for coil *l*

These signals are taken in linear combinations with weighting factors c_{lm} selected so as to yield the simulation of a magnetization pattern generated by phase encoding gradients (fig. 15.55). This requires combining the sensitivity profiles of the coils to form spatially varying harmonics:

$$\sum_l c_{lm}(x,y)\sigma_l = e^{i \cdot m \cdot \Delta k_y \cdot y} \qquad (15.42)$$

Consequently, the signal combination

$$s_m(k_x,k_y) = \sum_l c_{lm} s_l(k_x,k_y) = \iint \rho(x,y) \cdot e^{i(k_x x + (k_y + m \cdot \Delta k_y) y)} dx dy \qquad (15.43)$$

produces a new Fourier line at $k_y + m \cdot \Delta k_y$ in addition to the line at k_y that is generated by the gradient G_y. The gaps in *k*-space between acquired Fourier lines can thus be filled with *n* synthetic lines. In terms of image acquisition speed, this feature can be converted into an *n*-fold acceleration of image acquisition time compared to the conventional, purely gradient-based acquisition scheme. The theoretical maximum acceleration factor *n* is determined by the number of receive coils.

The original SMASH technique faces two major problems. The first is the dependence on the coil profiles. If spatial harmonics cannot be properly assembled in phase encoding direction, image reconstruction will result in residual fold-in artifacts. Furthermore, while the coil profiles should ideally remain invariant in the readout direction (i.e. orthogonal to the phase direction) experience shows that they do vary, a problem that may be solved by deploying a segmentation of the reconstruction process along readout direction, using different sets of coefficients for each segment. A second disadvantage of SMASH is found in its inferior signal-to-noise (*SNR*) performance [15.73]. It has been shown that a linear combination of signals from a coil array

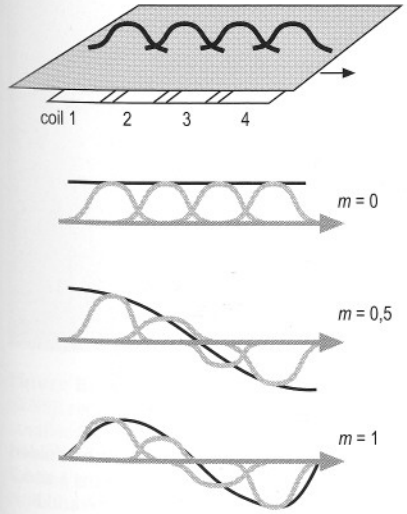


Figure 15.55
Generation of coil sensitivity patterns in SMASH. Different coil profile weightings are used to approximate different spatial harmonics.

(as is applied in SMASH) compares unfavorably in terms of *SNR* to the quadratic combination routinely used in standard imaging.

Methods related to the original SMASH approach have also been presented [15.67-15.69]. The aim of these methods is to refine the algorithms deployed so as to achieve robust, artifact-free and *SNR*-optimized reconstructions.

15.2.2.3 The SENSE technique

While the SENSE (sensitivity encoding) technique resembles SMASH in that it uses the sensitivity patterns of local coils as a substitute for gradient phase encoding steps, its mathematical approach to image reconstruction is rather different. Instead of generating phase encoding lines directly in *k*-space, reconstruction is performed in the image domain (i.e. after the Fourier transform).

For instance, if every other gradient phase encoding step in *k*-space is skipped for an acceleration factor of two, the result is equivalent to an image with half the field of view (FOV) in phase encoding direction. Those regions of the scanned object that are located outside the reduced FOV will fold over (aliase) as shown in fig. 15.54 (see also section 7.3.2). In other words, the PAT accelerated acquisition scheme (i.e. periodically skipping lines in *k*-space) translates into an unfolding problem in the image domain.

SENSE image reconstruction is a matter of restoring the full FOV by using each coil channel's tendency to display pixel intensities differently due to its sensitivity profile.

The concept of unfolding can be demonstrated using the following example (fig. 15.56): We assume that spin densities $\rho(P_1)$ and $\rho(P_2)$ at two locations P_1 and P_2 superimpose in the aliased images reconstructed from the signals of two coils that have different sensitivity profiles σ_A and σ_B. The aliased pixel intensity for the two coils can thus be expressed as

$$\begin{aligned} I_A &= \sigma_A(P_1)\rho(P_1) + \sigma_A(P_2)\rho(P_2) \\ I_B &= \sigma_B(P_1)\rho(P_1) + \sigma_B(P_2)\rho(P_2) \end{aligned} \qquad (15.44)$$

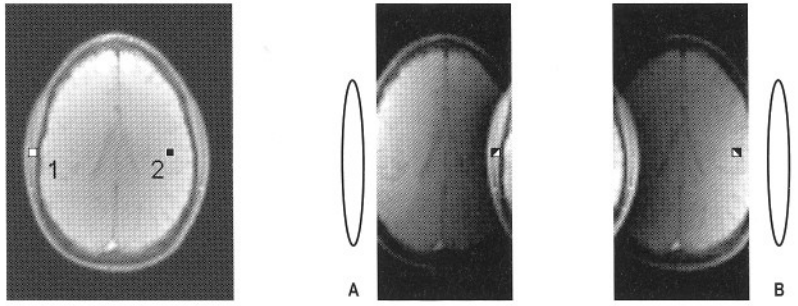


Figure 15.56
Unfolding aliased pixels with SENSE. **Left:** fully sampled, unfolded image. **Middle/right:** folded SENSE images prior to reconstruction, as received by coil A and coil B; folded image points 1 and 2 are no longer separate, but appear with different intensity values in both coil channels A and B.

Obviously, this set of equations can be solved for $\rho(P_1)$ and $\rho(P_2)$ if the sensitivities $\sigma_A(P_1)$ and $\sigma_A(P_2)$ are known. The noise characteristics of the receiving coil channels have to be taken into account for an *SNR*-optimized reconstruction. A detailed treatment of the SENSE reconstruction is given in [15.70]. Due to its pixel-by-pixel reconstruction, the SENSE technique is less demanding than the SMASH technique in terms of dedicated coil sensitivity profiles.

However, SENSE is subject to a tedious limitation in that the unfolding will fail whenever the image is pre-aliased, i.e. whenever the object extends beyond the chosen FOV in phase encoding direction [15.74] (fig. 15.57). In this case a two-fold ambiguity occurs: pixels beyond the borders of the FOV are folded back onto other parts of the image and, if phase encoding is reduced by PAT, these parts of the image fold in a second

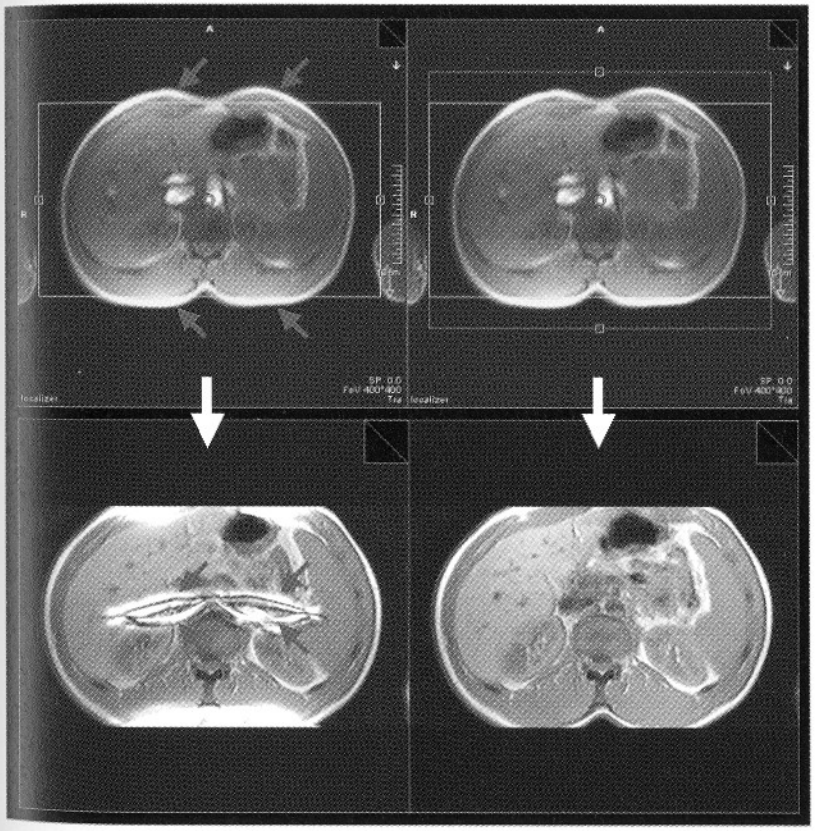


Figure 15.57
SENSE reconstruction of aliased images. **Left:** the selected FOV in phase encoding direction is smaller than the subject size (**upper row:** scout image with planned slice); the resulting aliasing cannot be handled by SENSE reconstruction (**lower row**), leading to severe artifacts. **Right:** folding is avoided by a few percent of phase over-sampling; the SENSE image is free of artifacts. Note that the additional time required for phase over-sampling reduces the net acceleration.

time. This problem does not occur in SMASH because the signal combination from the individual receiver channels takes place before the Fourier transform.

15.2.2.4 General technical aspects

The fact that PAT links the phase encoding process to the receiver coil geometry implies some noteworthy peculiarities. The most significant restriction is that PAT cannot be applied with the same flexibility as conventional, purely gradient-based imaging. The quality of the reconstructed image is strongly dependent on the encoding ability of the coil array. This implies a dependence on the selected phase encoding direction. Even coil arrays designed specifically for PAT allow only limited acceleration factors in certain phase encoding directions.

An inherent *SNR* reduction in parallel imaging is related to the applied coil array. Apart from the loss caused by the reduced number of Fourier line acquisitions, which leads to a relative *SNR* reduction by the square root of the acceleration factor, the data combination process may add a considerable amount of noise to the final image. Insufficient sensitivity variation or inappropriately located coils can result in excessive noise amplification. The so-called *g*-factor, which was originally introduced along with the SENSE technique [15.70], describes the amount of additional local *SNR* degradation (eq. 15.45):

$$SNR_{\text{accel.}}(x,y) = \frac{SNR_{\text{non-accel.}}(x,y)}{g(x,y) \cdot \sqrt{F}} \qquad (15.45)$$

g: *g*-factor describing local *SNR* degradation, *F*: Acceleration factor

Note that the *g*-factor is not a constant in space. This leads to the phenomenon of intra-image *SNR* variation (sometimes even with sharp edges) commonly observed in parallel imaging. Simulated or measured *g*-factor maps are a common tool for investigating the parallel encoding capacity of coil arrays.

The relationship between coil setup and geometric scan parameters (phase encoding direction, size and position of the field of view) on the one hand and the resulting image quality on the other is confusing for most users because it is neither intuitive nor commonplace in conventional imaging. Moreover, even small changes in slice orientation (and hence the phase encoding direction) and FOV size can have a considerably adverse effect on image quality.

15.2.2.5 Coil calibration

The fact that PAT is based on the local sensitivity profiles of the receiving coils makes sensitivity calibration a crucial part of image reconstruction. PAT reconstruction requires precise information about the sensitivity vector (magnitude and phase) of each coil for each pixel. This causes substantial practical problems.

It warrants mention that it is not sufficient in most practical settings to work with a time-invariant set of coil maps that have been calibrated once. This simple approach does not work for the following reasons:

- Coils that are used routinely may be bent and moved to fit the patient.
- Coils always show a certain degree of mutual coupling, which varies with different settings.
- The actual sensitivity profile of a coil array will vary depending on its electrical environment (including the dielectric properties of the human body).

A method is therefore required that includes coil profile calibration upon each individual examination. In what follows, we discuss two basic approaches. The first involves acquiring a volume scan immediately prior to examination to serve as a basis for coil profile extraction (prescan method). The second approach involves incorporating the acquisition of calibration data into each examination scan (auto calibration method).

One would normally assume that sensitivity profiles change rather smoothly in space and that coil maps would thus not need to be measured at high resolution. However, this assumption is not valid if the location of a coil is too close to the body: both magnitude and phase are subject to strong local variations near the coil wires. This makes it necessary in practice to find a reasonable tradeoff between resolution and scan time for the prescan.

The prescan method

A calibration prescan is ideally performed only once per examination. However, any displacement of the patient or the coil after the calibration scan will lead to a mismatch when the image is reconstructed.

A 3D-scan is typically performed to cover the full imaging volume. Such a calibration scan still consists of a superimposition of the coil sensitivity profiles and the anatomically-induced signal variations. A common method of canceling the anatomical caused signal variation is to acquire the coil map twice: for one scan the local coil array is used for signal reception, for the second scan the large whole-body coil with its homogenous receive profile is used. Dividing each local coil image by the homogeneous body coil image results in a pure sensitivity map of the coil array.

Further processing may be necessary (e.g. interpolation to handle anatomical structures exhibiting very low signal strength such as lungs and bones.

The auto calibration method

The auto calibration method incorporates a few additional phase encoding lines into the acquisition process for each image. These lines are typically placed around the center of *k*-space, which would otherwise be generated by the PAT scheme (dotted lines in fig. 15.58). One thereby obtains a fully sampled region in the *k*-space center, which represents a low resolution image that is free of aliasing for each coil. The required individual coil sensitivity maps can be obtained by normalizing to the sum of the single images.

The effective image acceleration factor is reduced in proportion to the number of added calibration lines. In the case of high resolution scans (e.g. full 256-matrix size or

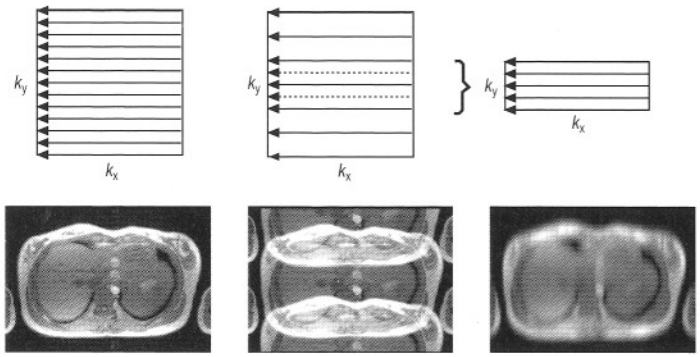


Figure 15.58
Coil map auto calibration. **Left:** non-accelerated acquisition. **Middle:** accelerated acquisition with a few lines added for auto calibration (dotted lines). **Right:** fully sampled central region of the accelerated scan provides a low resolution, but unfolded image that can be used to extract the coil sensitivity information.

higher), 12 additional lines are typically well tolerated while this number of additional lines might counteract the scan-time benefit of fast low-resolution scans. Given that each slice carries its own calibration data, the autocalibration method is hardly affected by potential changes in local coil profiles (e.g. due to patient movement).

PAT schemes based on k-space, in particular, benefit from the autocalibration approach. The explicit extraction of sensitivity calibration maps for each receiving coil can be skipped. As has been shown in the case of AUTOSMASH [15.67, 15.68] and as elaborated upon in further investigations, a procedure involving the direct fitting of the calibration lines to other measured lines provides the weighting coefficients for the signal combinations in advance. Interpolation problems within regions of low or vanishing signal strength do not occur.

Some *k*-space acquisition techniques even provide inherent calibration information. Radial scanning, where the *k*-space center is sampled much more densely than required by the sampling theorem, represents an example. As with AUTOSMASH, information can be generated from these samples to fill in for the missing values in the case of high *k*-values. Protocols for dynamic studies involving the continuous scanning of the same region represent another example [15.75]. For instance, each of the scans may be accelerated by a factor of two. If scans 1, 3, 5, ... skip every even line in *k*-space, and scans 2, 4, 6, ... skip the odd lines, then each set of two successive measurements will provide a fully sampled image that can be used to extract the PAT weighting factors. The assumption that the coil sensitivity variation between two successive measurements can be neglected is valid.

15.2.2.6 Applications-related aspects

PAT was originally introduced with a focus on imaging speed. Indeed, applications such as cardiac imaging and breath-hold abdominal imaging (including angiography)

benefit from either shorter scan times or increased resolution. However, PAT provides a still wider range of applications. The two main areas center on improvements in image quality and reductions in RF power.

Reducing unwanted relaxation effects

Shorter acquisition times following the excitation pulse reduce signal loss due to T_2 and T_2^* decay, the most obvious example being single-shot imaging [15.76]. Single-shot, turbo-spin echo imaging is restricted in terms of resolution (T_2 signal decay along the echo train). Echo planar imaging (EPI) suffers from susceptibility-induced distortions due to T_2^* dephasing, leaving this method impractical for many applications. In both cases, image quality can be greatly improved by applying PAT for a reduction in the echo train length.

Furthermore, the contrast properties of certain sequences can be improved by shortening the readout period following magnetization preparation (e.g. after inversion or saturation pulses).

SAR reduction

The application of many RF pulses within a short time may lead to an unwanted heating of the subject within the scanner. For high field strengths, in particular 3 T and above, this effect becomes a limiting factor for many clinical imaging protocols. This is because the RF power per RF pulse is proportional to the square of the field strength. Given that it reduces the number of RF pulses, PAT can be deployed to keep imaging protocols within specific absorption rate (SAR) limits.

Drawbacks

Despite all of the benefits mentioned, PAT is also associated with a number of drawbacks:

- Gain in imaging speed and/or image quality is counteracted by a loss in *SNR*.
- Specific requirements related to the coil setup complicate application. This represents a potential source of image degradation.
- Coil calibration adds to total scan time.

PAT significantly increases the complexity of both software and hardware in clinical environments. A robust and flexible PAT implementation requires scanners that are equipped with a large number of independent receiver channels and appropriate coil arrays (PAT has introduced a completely new paradigm for RF-coil design). A large number of simultaneously receiving channels alone represents a demanding task in terms of data rate, storage space and image reconstruction speed. User dependence on the data acquisition scheme and local coil setup introduces questions such as: what is the maximum acceleration factor for a given phase encoding direction? Dedicated user interfaces (UI) must provide convenient support to facilitate the avoidance of unfavorable settings.

15.2.2.7 Conclusion

Parallel imaging has opened new horizons for MR imaging in terms of scan speed and improved applications. In turn, it has added a new level of complexity to MRI concerning all system components, ranging from RF electronics, data acquisition and image reconstruction to pulse sequence design and (last but not least) user interfaces.

15.2.3 Whole-body imaging

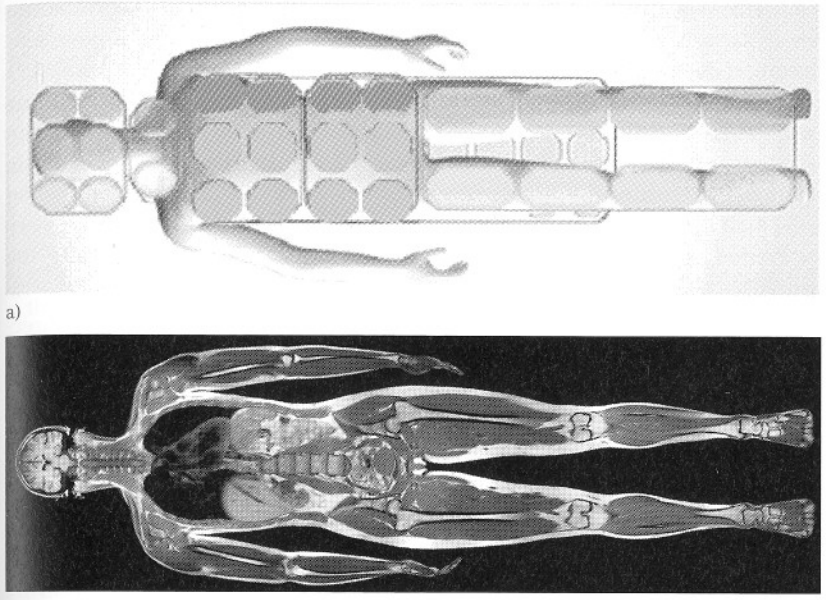
Some MRI examination types (e.g. spine imaging) require a field of view (FOV) along the magnet bore (z-direction) that is larger than the magnetic field's region of homogeneity (i.e. the intrinsic FOV). The natural approach to extended FOV imaging is to break up the desired z-range into smaller parts (stations). This then requires one to shift the patient position between the stations relative to the system's homogeneous region (isocenter) and to provide adequate coverage of local RF receiver coils for the extended FOV. The individual images can then be used separately or they can be combined into a composite image via a process called composing.

The development of extended FOV imaging was facilitated by the introduction of advanced scanner hardware such as patient tables capable of automatic movement and local RF coil arrays covering larger sections of the body. Fig. 15.59 shows an integrated patient table and local-coil concept that provide for automatic translation over a z-range of more than 2 m and table-mounted connections for up to 76 coil elements. Also shown is a composite multi-station image acquired in one contiguous exam on a 1.5 T MRI system with simultaneous reception by up to 32 individual RF receiver channels.

While extended FOV and whole-body imaging were developed from special applications involving standard scanners with intrinsic FOVs of approximately 0.5 m, the concept was a prerequisite for the development of imaging using ultra-short bore systems with much shorter intrinsic FOVs.

There are two groups of applications in which the scope of extended FOV imaging may encompass the whole body: examinations of the vascular system to determine the presence of malformations and atherosclerotic disease (see also section 15.2.4.3 on MR angiography) and metastasis imaging for the detection, characterization and staging of tumors. MRI competes in both of these areas with other imaging modalities, including X-ray imaging, CT, scintigraphy and PET. In many cases, MRI delivers at least equivalent diagnostic value, and may even provide additional diagnostic information thanks to its unique contrast mechanisms.

Angiography relies primarily on fast gradient-echo imaging and paramagnetic contrast agents. While peripheral angiography from the aorta to the foot has achieved the status of an accepted technique, whole-body applications have only been demonstrated recently [15.77, 15.78]. Efforts are being made to include cardiac imaging so as to provide comprehensive cardiovascular protocols. The resulting long examination times can be substantially shortened using parallel imaging techniques [15.79, 15.80].



a)

b)

Figure 15.59
Comprehensive coil setup **(a)** and composite multi-station whole-body MR image **(b)**

MRI is already the modality of choice in many areas of tumor diagnosis. Various MR imaging sequences have been deployed for metastasis screening, in particular short tau inversion recovery or STIR. Extensions to whole-body screening have recently been introduced [15.81].

The quality of extended FOV images is highly dependent on the accurate removal of geometric distortions induced by gradient nonlinearities in the MRI system. These cause a three-dimensional deformation that can be corrected via interpolation techniques. Given that the warping characteristics are time-invariant, each MRI system can be calibrated accordingly. The composing software uses the corrected images and their known positions to derive a single extended FOV image. A degree of station overlap is necessary to match images with remaining distortions.

Move during scan

One recent development involves the acquisition of data while the patient table is deliberately moved, a procedure referred to as move during scan (MDS) or continuously moving table MRI [15.82-15.84]. MDS removes the need to interrupt data acquisition between stations and is thus especially advantageous when the number of stations is high. Image deformations are minimized in that the data are mainly acquired from the isocenter. The remaining system-related spatial nonlinearities are distributed among the image data and therefore do not cause discontinuities between stations.

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