Editorial

A Guide to Understanding Key Aspects of Fast Gradient-Echo Imaging [1]

THERE IS A PLETHORA of fast imaging methods today. Each has its special contrast behavior, its strengths and weaknesses. On what basis does a clinical or basic science researcher design studies for comparison of methods, not just in initial scientific investigations but later in clinical trials? The study designs are generally based on the radiologist's knowledge of the disease and how it manifests itself on magnetic resonance (MR) images. Understanding has not been made easy by the proliferation of acronyms from industry and researchers alike.

As an example of the types of details that are important, consider the following case in spine imaging, in which high cerebrospinal fluid (CSF)–cord or CSF-disk contrast is of interest. Gradient-echo imaging is supposed to be able to provide this contrast when low flip angles are used. It appears to create T2-weighted or myelogram-like spine images. It seems appropriate to perform a study to compare gradient-echo images with T2-weighted spin-echo images. If we were to do this, we would also carefully examine lesion-tissue contrast. Low-flip-angle gradient-echo imaging does in fact provide myelogram-like spine images, but they are not T2 weighted. The signal intensity on these images is proportional to the spin density of the tissue when short echo times are used. To expect to see good lesion-tissue contrast based on T2 contrast is inappropriate. Therefore, spending manpower, resources, and patient time studying this is also inappropriate. It is possible to use longer echo times, but at higher field strengths, this is rarely done because of local field inhomogeneities and T2* signal loss. This is a particularly severe problem in two-dimensional (2D) imaging and less of a problem in three-dimensional (3D) imaging.

Fast gradient-echo MR imaging can generate images with different types of contrast, including spin density, T1, T1/T2, T2, and T2*. When properly understood and applied, gradient-echo images can be made to resemble conventional T1- and T2-weighted spin-echo images and to demonstrate additional clinical potential due to their speed advantages. (For more detailed discussions of the points addressed herein, references 1–4 should serve as appropriate reading.) We believe that fast imaging is destined to become a dominant MR imaging method in clinical practice.

The time has come not only to test the new fast imaging methods in hundreds of clinical cases but to use them properly to extract the desired contrast. The purpose of this editorial is to review some concepts that are often misunderstood, to list some of the important concepts for easy reference and to give some general recommendations and rules, and to encourage radiologists to incorporate fast MR sequences into their standard protocols.

STEADY-STATE SEQUENCES

In principle, one can distinguish between three types of MR signals in repetitive radio-frequency (RF) pulse sequences with TR much less than both T1 and T2. These signals lead to different types of fast gradient-echo sequences and a variety of image contrasts. The most simple signal is the free induction decay, or FID, that may be excited by a single RF pulse. It decays with the effective spin-spin relaxation time T2*, which contains both T2 and inhomogeneity effects. The acquired form of the FID in a gradient-recalled imaging sequence differs distinctly from an RF-recalled spin-echo sequence in that spins dephased because of magnetic field inhomogeneities are not rephased and related signal losses are not removed from the resulting images.

Incoherent Steady-State Sequences

The simplest form of the FID sequence is commonly referred to as FLASH (fast low-angle shot), spoiled FLASH, spoiled FAST (Fourier-acquired steady state), or spoiled GRASS (gradient-recalled acquisition in the steady state). The physical characteristics of the incoherent FID-type imaging sequences are based on a steady state of the longitudinal magnetization only. Residual transverse magnetization (or "transverse coherence") that exists before the application of another RF pulse is excluded from contributing to subsequent gradient echoes and thus to the image. This may be achieved by means of additional spoiler gradients but is best accomplished through constantly incremented phase differences between successive RF pulses ("RF spoiling"). The resulting equilibrium signal amplitude is proportional to the spin density, weighted by T2* relaxation (regulated by TE) and T1 saturation (regulated by flip angle and TR).

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Coherent Steady-State Sequences

Two other types of MR signal are obtained with related imaging sequences under conditions in which a steady state for both the longitudinal and transverse magnetization prevails. The entire steady-state free precession (SSFP) signal is composed of an FID part after the RF pulses and an echo part preceding the RF pulses. Thus, SSFP-FID imaging sequences such as FAST, ROAST (resonance offset averaging in the steady state), and GRASS acquire a gradient-recalled echo of the FID part of the SSFP signal in the same way that FLASH-type sequences acquire a gradient echo of the conventional FID. The SSFP-FID sequences exhibit the same proportionality to spin density and T2* as FLASH-type sequences but depend on a complicated function of flip angle, T1, T2, and resonance offset that results predominantly in T1/T2 contrast (*not* T1/T2*).

Similar arguments hold for SSFP-echo sequences that acquire a gradient-recalled echo of the echo part of the SSFP signal. However, because the SSFP echo represents an overlap of a large number of RF-refocused spin echoes and stimulated echoes that occur at the center of the RF pulses, SSFP-echo imaging sequences such as CE-FAST (contrast-enhanced FAST) or PSIF (mirrored FISP [fast imaging with steady-state precession]) yield considerably enhanced T2-like contrasts.

Finally, information from both the FID and echo portions is contained in the fully balanced FISP (also called true FISP) sequences. Since true FISP images are not integrated over the resonance offset, they are more heavily T1/T2 weighted than SSFP-FID images.

SPIN-DENSITY CONTRAST

Spin-density contrast is achieved by removing T1 contrast from FLASH-type sequences or T1/T2 contrast from SSFP-FID sequences. This may be accomplished by reducing the flip angle or increasing TR (eg, 10° at 100 msec for T1 = 1 second). Lower flip angles will be needed for shorter TRs. In general, the angle must be much less than the Ernst angle to remove T1 effects. Under these circumstances, no T1 saturation is imposed and no transverse coherences are generated, so that any differences between spoiled and refocused gradient-echo images vanish. The SSFP-FID becomes a regular FID, and its image transforms into a FLASH image.

Spin density represents a wide range of signal intensity and excellent contrast potential. However, despite the fact that spin-density–weighted brain images demonstrate good contrast between CSF and gray/white matter, low-flip-angle gradient-echo images must not be confused with T2 (or T2*)–weighted images. In fact, a predominance of spin-density contrast requires the use of short gradient-echo times to minimize T2* effects. Typical values of 5–6 msec are short enough at 1.5 T to fulfill this condition, except for imaging marrow, hematomas, and metal implants and when contrast agents are used.

T1 CONTRAST (FID)

T1-weighted fast gradient-echo images represent the original domain of (spoiled) FLASH sequences. Respective pairs of TR and flip angle may vary from 15 msec/20° to 40 msec/40° or 100 msec/70° for cross-sectional 2D imaging. For FLASH sequences, the signal-to-noise ratio (S/N) varies as $\sqrt{TR/T1}$ at the optimum flip angle. Hence, a longer TR can be used for improved S/N and allows for more sections up to a TR approximately equal to T1, while multiple short TR acquisitions of a single section yield images with the same S/N but are clearly less efficient. Of course, if motion plays a role or if flow information is desired, only a short TR, single-section sequence provides images with significantly reduced motion artifacts and less inflow enhancement. In that case, sequential 2D images should be acquired. General preferences with regard to single- or multisection gradient-echo imaging may be affected by a clinical need to minimize study times. These choices are also motivated by the follow-up of real-time signal changes in dynamic contrast agent studies or the requirement to overcome motion problems in abdominal imaging by means of breath holding.

Three-dimensional images with optimized S/N are best obtained by increasing the number of 3D partitions (ie, the number of phase-encoding steps in the section-select direction), rather than by signal averaging with repetitive acquisitions. Hence, more sections can be covered, and the problem of aliasing, or losing outer-section information, is reduced while the advantage of a higher S/N is maintained. For example, in combination with RF spoiling, excellent T1-weighted 3D FLASH images with a data matrix of 128 × 256 × 256 voxels and a field of view of 160 × 250 × 250 mm may be obtained within a measuring time of 8 minutes, assuming a TR of 15 msec and flip angles of 15°–20°. In fact, the accessible T1 contrast in 3D images is more pronounced than in 2D images, in which integration over the nonuniform section excitation profile reduces the theoretical contrast expected on the basis of the chosen TR and flip angle. This particularly applies to brain studies and gray-white matter contrast, so that 3D studies are recommended unless otherwise precluded. Short TE 3D imaging can be used for high-resolution detailed studies of the pituitary gland, the knee, or any other relatively stationary area of interest.

T1/T2 CONTRAST (SSFP-FID)

To create a steady state in the transverse magnetization, two different conditions must be fulfilled simultaneously: (*a*) The tissue must have a long T2 relative to TR, and (*b*) the overall signal phase must be constant from one repetition cycle to the next. The latter condition not only requires a rewinding (balancing) of the variable phase-encoding gradient in each TR interval but also restricts the application of SSFP sequences to stationary tissues (unless properly compensated), since motion-induced phase errors serve as a natural phase spoiler (4,5).

If one of the two SSFP conditions is violated, then SSFP-FID sequences transform into FID sequences. In other words, refocused FLASH becomes (spoiled) FLASH in cases in which no transverse coherence remains to be spoiled or refocused. Typical examples of this are the experimental conditions (low flip angle, long TR) used for obtaining spin-density contrast; the short T2 values encountered in the musculoskeletal system, in joints (with the exception of synovial fluid), and in most abdominal organs; and the motion of blood and myocardial wall in cine heart studies. In fact, depending on the tissue properties (motion, T2 values), parts of an SSFP-FID image may exhibit T1/T2 contrast while others show T1 contrast. Although this ambiguity may be confusing in general, one useful application is in SSFP-FID 3D imaging of joints, where T1/T2 contrast highlights synovial fluid (5) while T1 contrast prevails in cartilage, muscle, and marrow (similar to that in a 3D FLASH image). For example, 3D imaging of the knee or spine with 64 1–2-mm-thick sections (partitions) can be performed in just 4 minutes with a TE of 5msec and a TR of 15 msec, for in-plane res-

olution of 1 mm or less. In fact, because S/N is proportional to $\sqrt{T2/T1}$ (as long as TR $\ll$ T2) for SSFP imaging at the optimal flip angle, the shortest possible TR should be used to maximize S/N.

ADDITIONAL T2 CONTRAST (SSFP-ECHO)

In most clinical applications, it is desirable to remove any ambiguities and to acquire images with the highest possible T2 contrast. Fast gradient-echo SSFP-echo images come closest to the characteristic features of "late" spin-echo images. Sequences with a TR of 15 msec or longer and a flip angle of about 30°–50° result in both a high S/N for normal tissues and high contrast for tissues with prolonged T2. This allows for very rapid 2D and 3D imaging, with times of 4 seconds (256 × 256) and 4 minutes (64 × 256 × 256), respectively.

On the basis of the steady-state considerations discussed in the previous section, it is not surprising to acknowledge the high sensitivity of SSFP-echo sequences to motion. Even slow flow of the CSF in the ventricles often generates enough phase variability during imaging to effectively distort steady-state transverse coherences (4) and to cause signal voids and motion ghosting on the images. Image quality may depend on the choice of image orientation and also varies from subject to subject. On the other hand, even abdominal SSFP-echo images may be possible when restricting imaging time to breath-hold intervals (eg, 10–15 seconds). If imaging is successful, strong lesion-tissue T2 contrast becomes available in kidney or liver studies.

MAGNETIC FIELD INHOMOGENEITIES, OR T2* CONTRAST

In many cases, the sensitivity of gradient-echo images to susceptibility differences or magnetic field inhomogeneities can be clinically useful. Examples include the finding of calcifications, the examination of hemorrhage and bone marrow, and the investigation of tissue perfusion with contrast agents.

If inhomogeneities are viewed as artifact producers that give rise to signal dropout near air-tissue interfaces (eg, in the vicinity of the pituitary or the sinuses), the use of short TEs will significantly reduce the effect. In addition, any reduction of voxel size will linearly decrease sensitivity to susceptibilities, rendering 3D imaging at short TEs particularly attractive (6).

Depending on the sequence structure, a form of zero TE imaging with a missing pulse approach can be applied (1,2) to completely prevent T2* effects. This could prove particularly useful at very high field strengths in small-bore animal imagers or in microimaging systems in which field strengths are near 10 T.

In general, T1-weighted 3D FLASH sequences in combination with T2-weighted 3D SSFP-echo sequences are, in most respects, superior to classic multisection spin-echo sequences. This includes advantages such as the unsurpassed spatial resolution due to contiguous sets of images with section thicknesses on the order of 1 mm and the unique potential for retrospective multiplanar reconstructions.

HIGH-SPEED GRADIENT-ECHO IMAGING

Recent applications of gradient-echo sequences in the subsecond imaging regime differ from fast MR imaging with acquisition times of several seconds in two important respects: (*a*) Image acquisition is not in a steady state but starts from equilibrium and proceeds during the approach to steady state (7), and (*b*) regardless of technical differences, all high-speed imaging studies sacrifice image quality in terms of spatial resolution and S/N.

Even though sequences with high flip angles and acquisition times close to 1 second may generate images that resemble conventional SSFP images, steady-state transverse coherences do not have enough time to develop in a time that is short relative to T1. For low flip angles, the first difference simplifies the sequence to a FLASH-type sequence. Because very short TR FLASH sequences excite only a limited number of FID signals (the signal is approaching equilibrium), much higher flip angles (10°–20°) may be chosen than in comparable steady-state (RF-spoiled) sequences with lower flip angles at the same TR (7). Thus, single-shot high-speed FLASH images exhibit a somewhat higher S/N than is expected from steady-state applications.

The second difference mentioned above is based on the fact that a short imaging time and a good S/N are not only achieved by reductions in TR (5 msec or less); they are largely due to a reduction of the data matrix (typically 64 × 128) and an increase in section thickness (10 mm). In fact, both echo-planar images and subsecond FLASH images usually have large voxel sizes of 20–80 mm^3 as opposed to 1–5 mm^3 for conventional fast gradient-echo or spin-echo images. It should be noted that decreasing the number of sampling points by a factor m in any direction but holding the field of view fixed results in a $\sqrt{m}$ increase in S/N.

Because data acquisition in high-speed FLASH imaging is effectively performed outside steady-state conditions (ie, without imposing T1 saturation), the resulting images have spin-density contrast. Moreover, the use of 2–3-msec TEs reduces phase distortions due to motion and susceptibilities and, hence, virtually eliminates related image artifacts. Strong spin-echo T2 contrast may be obtained by preparing the initial equilibrium magnetization with a 90°–TE/2–180°–TE/2–90° pulse sequence (8) or, more effectively, by a magnetization transfer pulse sequence (9).

Correspondingly, inversion-recovery (IR) T1 contrast is achieved with a 180° inversion pulse followed by a variable recovery delay before the high-speed imaging sequence (10). Longitudinal relaxation during data acquisition can be considerable. If there is rapid T1 relaxation during data acquisitions, the data are high-pass filtered and the images contain artifacts. This particularly applies to the IR-FLASH concept in which tissue nulling is obtained during image acquisition. To circumvent this problem, most contrast manipulations require segmentation of the total sequence into several steps, including waiting periods. For example, the IR-FLASH concept leads to tissue signal nulling during image acquisition and, therefore, may cause serious high-spatial-frequency image artifacts. Of course, the incorporation of a waiting period of about 0.5–1.5 seconds eliminates the high-speed advantage. Another danger of IR-FLASH contrast is the cancellation of signal from two overlapping tissues whose inversion times are such that one is still negative while the other is positive during data sampling. This will cause signal cancellation and may make an image appear as if it has better contrast when, in fact, the apparent contrast is an artifact.

Meaningful applications for high-speed gradient-echo imaging are still under development (11,12). The usefulness of segmented 2D sequences for imaging the central nervous system or even the abdomen seems rather questionable in view of the excellent quality of "conventional"

fast gradient-echo images obtained with state-of-the-art technology. On the other hand, recent applications of "paused" (segmented) IR-FLASH sequences for 3D imaging of cardiac (11,12) (electrocardiograph triggering) and/or thoracic (breath hold) anatomy promise to yield additional clinical information and may prove competitive with echo-planar imaging for volume coverage and image quality.

CONCLUSIONS

To fully exploit the potential of fast MR imaging, to choose which sequences to use, and to optimize clinical applications, it is necessary to understand the underlying contrast mechanisms and the strengths and weaknesses of each method. Major concerns include the choice of sequence type (FID, SSFP-FID, SSFP-echo), sequence parameters (TR, TE, flip angle, etc), imaging method (single-section, multisection, 3D, high-speed), and desired contrast (spin-density, T1, T1/T2, T2, T2*). Moreover—although outside the scope of this report—the access to strong inflow effects in fast gradient-echo imaging provides the basis for current MR angiography techniques that exploit either flow-related signal amplitudes or phase properties.

General recommendations for fast gradient-echo imaging involve the use of (*a*) short TEs of 5–6 msec or less for all sequences, (*b*) the (spoiled) FLASH sequence for spin-density– and T1-weighted images, (*c*) the true FISP sequence for images with the most T1/T2 weighting, (*d*) the CE-FAST sequence for more T2-weighted images,(*e*) 3D rather than 2D acquisitions where feasible (head, spine, musculoskeletal system), and (*f*) breath-hold imaging along with high-speed acquisitions where physiologic motion is slow enough (abdomen).

Ideally, fast gradient-echo imaging will replace spin-echo methods in a variety of clinical settings, although adequate clinical comparisons still must be performed to verify any particular proposed application. No doubt the proliferation of acronyms has made it difficult to follow the many new developments in fast MR imaging. It is the aim of this communication to encourage the widespread testing of these sequences in a manner that will effectively point the way for specific new clinical applications that will be best handled with the fast imaging methods described herein. Finally, these methods have applications not only at the most common field strengths for whole-body imaging but at specific field strengths ranging from 0.04 to 9.7 T, depending on the sequence. ●

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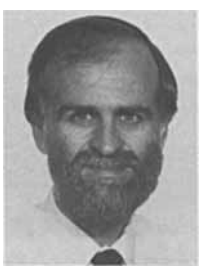

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