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Echoes—How to Generate, Recognize, Use or Avoid Them in MR-Imaging Sequences
Part II: Echoes in Imaging Sequences

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Part I of this sequence of two papers on spin echoes introduced the echo phenomenon and presented the extended-phase-graph algorithm for the description of signals arising in the multi-pulse sequences. In Part II, pulse sequences that are commonly used in MR imaging are discussed with the use of this algorithm. First, it is shown that phase graphs easily demonstrate how a proper multi-echo imaging experiment should be designed. Then, they are used to shed some light on the signal behavior of gradient-echo sequences. Finally, some hints are given about how to avoid or use spurious echoes in more general multi-pulse sequences.

MULTI-ECHO IMAGING SEQUENCES

Figure 1 is an idealized representation using a constant gradient and hard pulses throughout the experiment. It is valid for the description of an MR-imaging experiment only if the switched gradients are applied such that the phase graph at the relevant points, which are immediately before and after each refocusing pulse, coincides with that in Fig. 1. In addition, for the discussion of MR-imaging sequences, phase graphs should always consider all possible pathways for all values of α, because the shaped pulses used in MR can never be described by one particular value of α.

For example, the switching off of the read-out gradient during the pulses is permitted as long as the integral over this gradient in all refocusing periods remains constant and equal to two times the integral over the gradient between the excitation pulse and the first refocusing pulse. Such a phase graph for the read gradient of a possible multi-echo imaging sequence is shown in Fig. 2. If the latter integral does not fulfill this condition, the phase graph will appear as in Fig. 3. Multiple sets of echoes will evolve that occur at different times. The time difference between the proper echo and the spurious echoes will be given by the deviation of the gradient from its proper value, as displayed in Fig. 2. The resulting images will show artifacts

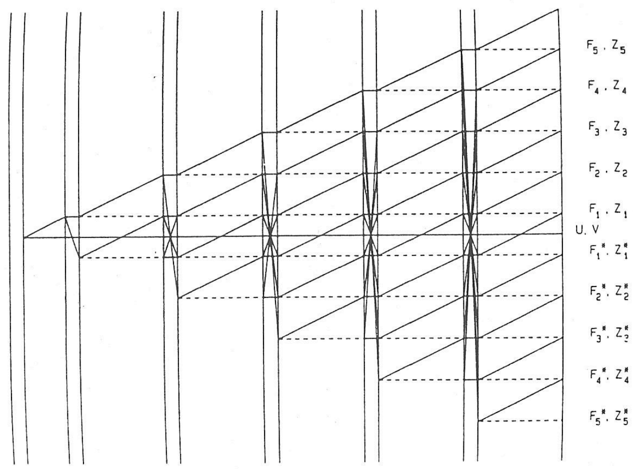

Figure 1. Extended-phase graph for a Carr-Purcell-Meiboom-Gill (CPMG) multi-echo sequence. Pulses are represented by two closely spaced vertical lines. The mixing between configurations F_n, F_n^*, Z_n, and Z_n^* with identical n is shown by the lines that connect these states. Time evolution of transverse magnetization is shown by solid lines between pulses; the dotted horizontal lines represent configurations of z magnetization.

created by interference of the two sets of signals. Although such interferences in similar pulse sequences can be put to practical use, like the measurement of flow velocities or magnetic field profiles (*1*), they are strictly unwelcome in a pure multi-echo imaging experiment. Avoiding this can be a considerable problem in whole-body imaging systems where eddy currents lead to additional time-dependent gradients. Only a few state-of-the-art MR-imaging systems are in fact capable of performing CPMG multi-echo imaging. The cheaper way is, of course, to apply the refocusing pulses in a nonperiodic time pattern or to use additional gradients to select the desired primary echoes. Because of the vast number of possible signals after a few pulses, this

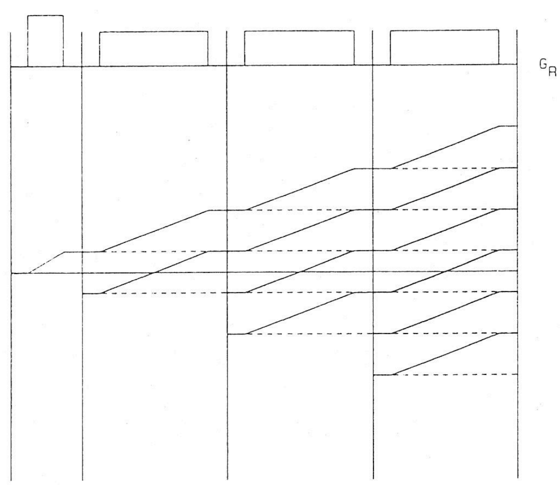

Figure 2. Extended-phase graph for a properly executed multi-echo sequence with a pulsed gradient. All essential features are identical to those shown in Fig. 1.

strategy works only for a very limited number of echoes. It also carries another penalty not attendant to a "real" multi-echo experiment. This is the erosion of the slice profile from one primary echo to the next as a consequence of the $\sin^2(\alpha/2)$ dependence of the refocused signal intensity. The stimulated echo contributions for the CPMG experiment avoid this problem. The second echo is in fact generally higher than the first echo because these stimulated echo contributions more than compensate for the loss of signal due to T_2 and imperfect primary echo formation (*2*). Both approaches are inadequate for T_2 calculations because the signal amplitude is in general a function of α and T_1, as well as of T_2. Refocusing pulses that affect a much thicker slice than the excitation pulse can be used to reach a better primary echo formation for all excited spins. However, these severely limit the possibilities of performing the experiment in the multi-slice mode.

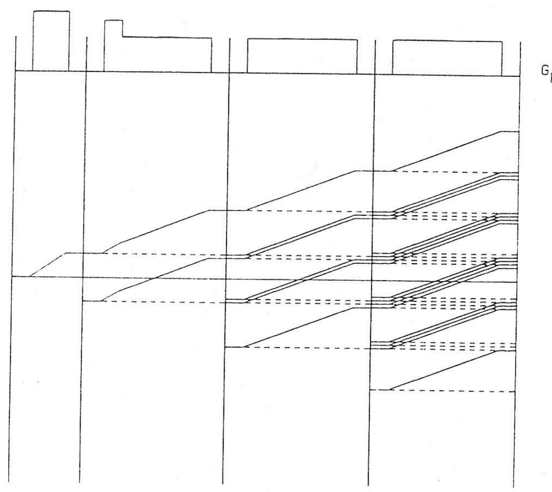

Figure 3. Extended-phase graph for a multi-echo experiment where the total gradient between the excitation pulse and the first refocusing pulse is not identical to ½ of the gradient between two refocusing pulses. Multiple refocusing pathways arise as a consequence of this disturbance.

Care also must be taken for the phase-encoding gradient. If phase encoding is done between the excitation pulse and the first refocusing pulse, this will again lead to two different sets of signals that interfere with one another. The result will be a ghost image that is mirror symmetrical in the phase-encoding direction to the proper image. This can be avoided if all phase encoding is done before each echo and refocused before the next refocusing pulse. Under this condition, Fig. 2 will be representative for the whole imaging experiment.

The necessity to undo the phase encoding before every pulse makes it possible to introduce a different phase encoding for each echo. The number of excitation steps necessary to sample all data for image reconstruction consequently can be reduced in the extreme case that a single multi-echo train is sufficient to acquire all data. The resulting RARE-fast (rapid acquisition with relaxation enhancement) imaging sequence has been used for several years as a fast-imaging technique (*3, 4*). The strong T_2 contrast of the resulting images makes this method extremely attractive for several clinical applications. The use of small refocusing flip angles to avoid potential rf deposition problems immediately follows from the above argument. Because the eddy current problems described for conventional multi-echo sequences apply equally to RARE, only recently has this method come into more widespread use (*5*). Following a basic law of MR imaging, it is to be expected that just as in gradient-echo imaging a few years earlier the same experiment will be rediscovered and rebaptized by each manufacturer of MR equipment. Already the acronym FSE (fast spin echo) has been cited for this method. Let the reader be assured that — slightly different from gradient-echo sequences we try to sort out in the next section — there is only one way to do fast imaging based on multi-echo sequences, and that is to use gradients such as those shown in Fig. 1.

GRADIENT-ECHO SEQUENCES

The possibility of generating a signal that is useful for MR imaging by reversing the read gradient rather than by a refocusing rf pulse was introduced in 1985 (*6, 7*). Because signal saturation due to T_1 is much lower than in spin-echo sequences, this principle can be used to acquire images with good signal-to-noise ratios in only a few seconds. The highly complicated contrast behavior of the resulting images has prevented gradient-echo imaging from supplanting the much longer conventional spin-echo techniques, except for applications such as imaging of the joints. There are, however, important applications, such as three-dimensional imaging, the observation of time-dependent signal intensities in contrast agent studies, and most of the methods currently used in MR angiography. These are indispensable in today's MR imaging, and they use gradient-echo sequences as their basis. Subsecond gradient-echo imaging (*8*) has recently opened new applications, such as the examination of tissue perfusion. During the first few months after its introduction, several papers were published on the signal intensity of gradient-echo images in dependence to T_1, T_2, and other parameters, such as chemical shift and susceptibility (*9-13*). It turned out that even if only T_1 and T_2 were taken into account, a highly different contrast can be expected for the same experimental conditions with respect to the flip angle and timing of the sequence if the gradients of the sequences are varied.

For the general understanding of this contrast behavior, it has turned out not to be very helpful that vast numbers of acronyms have been used for one and the same sequence. At the same time, significant differences between methods with different labels leading to different image contrasts have not generally been propagated. Some of this obscurity seems to be purposeful: A thorough discussion might reveal that sequence XXXX is identical to sequence YYYY. Meanwhile, everyone who wanted to become famous by reinventing gradient echoes has succeeded (or failed) in doing so, and several review articles have appeared that try to establish a common basis for discussion.

The formalism used is based on Carr's article on the steady-state behavior of magnetization in repetitive-pulse sequences (*14*). Although this approach leads to absolutely correct results — at least if it is applied correctly — the resulting formulas for the image contrast apparently have failed to communicate an adequate insight into the contrast behavior of such sequences, especially to those who are not initiated into the intricacies of magnetic resonance and who shy away from too many sine, cosine, and exponential functions. Practically everyone working in diagnostic MR belongs to this group, so a simpler approach might be welcome. In the following section, I will try to clear up the contrast behavior of gradient-echo sequences by treating them as echo sequences using the extended-phase-graph formalism described in Part I.

We will assume that a steady state with respect to M_z has been reached. If the phase graph is started at one particular pulse, the result will be as shown in Fig 4. It should be noted that no attempt has been made to include the phase of the pulses with the subsequent generation of different configurations. Even so, this rough sketch already reveals interesting information about gradient-echo sequences. It is obvious that the sequence is self-refocusing whenever the integral over the gradients between two pulses is constant. Every signal generated by the n^{th} rf pulse will then be fully refocused before the $(n + 2)^{nd}$ pulse. One part of this refocused magnetization will contribute to the signal after the $(n + 2)^{nd}$ pulse. The other part will be converted into z magnetization, which will generate a signal after the $(n + 3)^{rd}$ pulse. The proportion of both parts depends on the flip angle α and the relative phase $\Delta\psi$ of the refocused signal and the $(n + 2)^{nd}$ pulse.

The stimulated echo mechanism will lead to refocusing before the $(n + 3)^{rd}$ pulse. This refocused signal will again contribute to the next signal or be converted into z magnetization for further use.

The signal, which is observed in the steady state, will then be a sum of all possible components. It should be noted that long relaxation times T_1 will lead to a reduced signal intensity of the primary gradient echo but at the same time will increase the relative amount of the signal, which is "refocused" *via z* magnetization. Additional measures must be taken if the refocused portions of the signal are to be avoided. Some possibilities are discussed below.

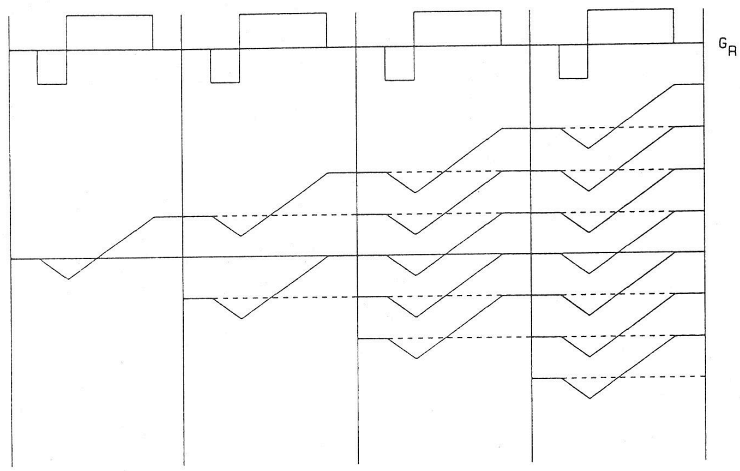

Figure 4. Extended-Phase Graph for a Gradient-Echo Sequence. All refocusing pathways generated by excitation from the first pulse are shown. Refocusing occurs after one "silent" repetition period in every subsequent period *via* multiple refocusing pathways.

Refocused magnetization alone, without contributions from the primary gradient echo, is generated if the gradient is time-reversed in the method known as CE-FAST (contrast enhanced Fourier acquired steady state) (*15, 16*). No primary echo is being formed. One gets a spin echo after the next pulse, however, a stimulated echo after the $(n + 2)^{nd}$ pulse, and so on (Fig. 5).

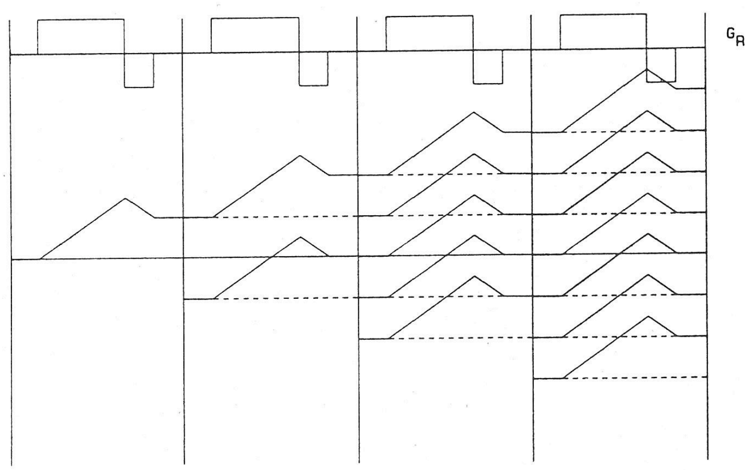

Figure 5. Same as Fig. 4 for a CE-FAST Experiment. No direct gradient echo is generated. Signal is generated from the second repetition period in every subsequent period.

The final signal will again be a sum of all these contributions; the signal intensity will again depend on T_1, T_2, and α. Because no primary gradient echo contributes to the final signal, the positive T_1 and T_2 components of the refocused magnetizations will be stronger than the negative T_1 contrast given by saturation in comparison with the refocused FLASH (fast low angle shot) sequence. Finally, attempts can be made to combine the signals of refocused FLASH and

CE-FAST by refocusing all gradients between two pulses. This special case has been published under the acronym FISP (fast imaging with steady precession) (*17*).

Any gradient-echo sequence falls into one of these four types with respect to the refocusing behavior. An attempt to place the more commonly used sequences into their respective categories and a qualitative assessment of the contrast is shown in Table 1. In the following sections, each category will be discussed in a little more detail.

TABLE 1
An Attempt to Correlate Gradient-Echo Sequences with Different Acronyms to Their Proper Generic Sequence

Sequence	Acronym	Remarks
Spoiled Gradient Echo	(spoiled) FLASH, FFE GRASS	Very often no clear distinction is being made between refocused and spoiled variants.
Refocused Gradient Echo	(refocused) FLASH, CE-FFE, FAST, GRASS, (FISP)	The FISP experiment as performed on many systems most probably belongs to this group.
Contrast Enhanced Gradient Echo	CE-FAST, SSFP	
Fully Refocused Gradient Echo	FISP	Very sensitive to field inhomogeneities.

Nonrefocused or Spoiled Gradient Echoes

The term nonrefocused FLASH implies that something must be done to generate refocused magnetization. It has already been pointed out that the opposite is true: A simple gradient-echo sequence leads automatically to refocusing. Additional measures must be taken to avoid this. It is obvious that spoiling the transverse magnetization by an additional gradient pulse in any direction is totally inefficient because its effect also will be refocused after two excitation periods. It has therefore been proposed that the transverse magnetization can be spoiled by a gradient that varies from one excitation period to the next. Because the voxel size across the slice is in general larger than that in the image plane, such a time-variable spoiling gradient is best applied in the slice selection direction. An efficient way to vary the gradient is to treat it just like a phase-encoding gradient.

To make spoiling by gradients efficient, some minimum time must be allowed for the spoiling gradient. Even with very fast and strong gradient systems, a minimum of at least 5 milliseconds (ms) should be allowed. This problem is nonexistent if rf pulses are used; they are good excitation pulses but inadequate refocusing pulses. One possible approach would be to use adiabatic pulses, which as yet have not been used in practice. Another approach is to use so-called prefocusing pulses (*19*). The idea is to use pulses that by themselves generate a strong linear phase dependence of the signal across the slice. If x is chosen as the slice direction, then the phase of the signal immediately after the rf pulse can be described by

$$\psi_1(x) = \psi_{10} + \Delta\psi(x) \qquad [1]$$

where $\Delta\psi(x)$ is the rf dependent phase variation across the slice, and ψ_{10} is the phase of the signal without this phase variation due to the effect of the slice-selection gradient. The term "prefocusing" arises from the fact that both terms can be made to cancel each other such that ψ_1 becomes zero even with the gradient on.

The phase effect of the next pulse will be given by

$$\psi_2(x) = \psi_{20} + \Delta\psi(x) \qquad [2]$$

if the same pulse-shape is used with another reference phase ψ_{20}.

The phase of the gradient echo formed by the first pulse alone will then be given by

$$\psi_{GE}(x) = \psi_1(x) + 90° \qquad [3]$$

The phase of the refocused signal will be given by

$$\begin{aligned}\psi_{SE}(x) &= 2 \cdot \psi_2(x) - \psi_1(x) \\ &= 2 \cdot [\psi_{20} + \Delta\psi(x)] - \psi_{GE} \qquad [4]\end{aligned}$$

If such pulses are applied using gradients designed to work with pulses without rf dependent phase terms, then the spins at the time of signal formation will be dephased, and the signal amplitude will be reduced. Because we have assumed $\Delta\psi(x)$ to be linear across the slice, this dephasing can be easily corrected by an additional gradient in the slice direction. It is now easy to see from Eqs. [2] through [4] that this correction can be performed either for the primary gradient echo or for the refocused signal, but not for both. Therefore, this gradient can be used to determine the mixture between direct and refocused signal contributions.

Radio frequency pulses that produce a linear phase dependence for off-resonance frequencies have been produced using the simulated annealing method (*18*, *19*). A more pedestrian approach useful for the limited amount of dephasing necessary for our purpose would be the use of the superposition principle (*20*), for which the slice-selection pulse could be composed of several components, each selecting only a small part of the total slice and each carrying a different phase ψ_1. Of course, other methods for the calculation of selective rf pulses can be used as well.

This particular variant of the gradient-echo sequence, which allows deliberate changing of the mixture between primary and refocused signal contributions, has not been described in the literature. It would therefore be tempting to join the crowd of gradient-echo sequence inventors and give it a new acronym, which could be FIFI (fast imaging using the FIFI sequence). On second thought, it would be a shame to waste a nice general purpose acronym on one particular pulse sequence.

Refocused Gradient Echoes

As has already been discussed, the refocused version is the most "natural" gradient-echo sequence. The fate of the refocused magnetization depends in principle on the phase of the rf pulses. If the magnetization is refocused *via* the spin-echo or stimulated-echo mechanism, such that it is in phase with the next rf pulse, then it will be unaffected by it. If it is orthogonal, it will be converted into z magnetization, which will then be excited by the next pulse. Because the T_1 of most tissues is long compared to the repetition time of gradient-echo sequences, this does not significantly change the contrast of the resulting images.

So far, discussion has been limited to refocusing with respect to gradients that are the same during each excitation period. This applies to the read gradient and the slice-selection gradient and, of course, to all magnetic-field inhomogeneities but not normally to the phase-encoding gradient. If no measures are taken to make the total effect of the gradient in the phase-encoding direction between two pulses constant throughout the experiment, then the refocused magnetization will carry different phase encoding and will produce a different echo. The simplest way to have the images formed by the primary gradient echo and the refocused echoes coincide is to refocus the phase-encoding gradient after data acquisition. Total refocusing of the gradient nulls its effect, which is not strictly required. It is sufficient to bring the total effect to any nonzero value that is constant for each excitation period.

Contrast-Enhanced Gradient Echoes

The CE-FAST method is distinguished from the refocused FLASH method by the fact that it uses only the refocused signal contributions and omits the primary gradient echo. The same arguments apply to the effect of the phase of the rf pulses or the refocusing of the phase-encoding gradient as apply to the refocused FLASH sequence.

Fully Refocused Gradient Echoes

The FISP sequence as originally published (*17*) uses fully refocused gradients for each excitation period. Because the phase of the signal is zero before each pulse, the resultant phase graph, including the effects of all gradients, consists of a simple zigzag line where no configurations with higher dephasing are being formed. Again, the phase of the rf pulse will lead to differences in the signal evolution. For constant pulse phase y throughout the experiment, the magnetization, which is refocused by gradient reversal onto the $(n + 1)^{st}$ pulse, will have phase y and will be converted into z magnetization, which will be excited by the following pulse. As a consequence, the refocused signal will oppose the primary gradient echo and the signal amplitude will be reduced. The use of an alternating pulse phase avoids this problem and leads to constructive superposition of signals.

The problem of the experiment is the fact that magnetic-field inhomogeneities cannot be refocused by gradient reversal and will consequently affect the different signals in a different way. Phase differences of the different signal contributions will arise that lead to positive or negative interference. Because the field inhomogeneity varies over the observed volume, the image contrast also will be inhomogeneous.

To discuss whether this problem is of practical importance, let us calculate the field inhomogeneity that will lead to a phase shift of 180° and consequently to positive or negative signal interference. For a typical repetition time of 50 ms, a difference in the Larmor frequency of 10 s^{-1} will bring the signal components into opposition. On a 1T magnet, such a frequency difference is caused by a field inhomogeneity of $10/42 \cdot$ ppm = 0.23 ppm (parts per million). Because whole-body magnets are specified to field inhomogeneities of 1-10 ppm, it is clear that a FISP experiment as proposed in Reference 17 cannot be performed on a clinical whole-body system. This is demonstrated in Fig. 6. A FISP experiment with true refocusing of all gradients

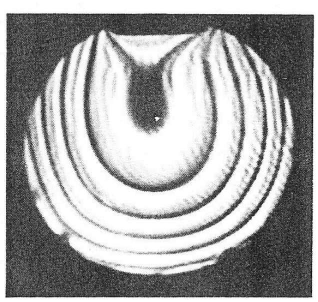

Figure 6. FISP Experiment Performed on a Water-filled Sphere. The stripe pattern is generated by destructive interference of refocused signals with different and space-dependent phases.

was attempted on a water-filled phantom used for proton spectroscopy. The inhomogeneous line-width over the whole phantom is about 10 Hz, corresponding to about 0.12 ppm on a 2T system. The dark lines are created by destructive interference between echoes carrying different field-dependent phases. The consequence of this discussion can only be that the FISP sequences installed on numerous whole-body systems are in fact refocused FLASH sequences, which produce nearly the same contrast as proper FISP but are insensitive to field inhomogeneities.

RESULTS AND COMPARISONS

The different signal contributions adding to the total signal of the four different ways to perform a gradient-echo experiment are diagrammed in Fig. 7, which demonstrates the general character of these sequences. Figure 8 shows results from a healthy volunteer, which demonstrate the different refocused signal contributions. No results from FISP are shown, for obvious reasons. Of course, explicit calculations of signal intensities can be done by applying the matrix formalism described in Part I. For the purpose of making the right decision for a particular diagnostic application, such formulas are, however, too overwhelming for most clinical MR users to be of any practical value. For misanthropic physicists, such calculations can be quite entertaining because they serve to point out the numerous errors in papers on the contrast of gradient-echo sequences.

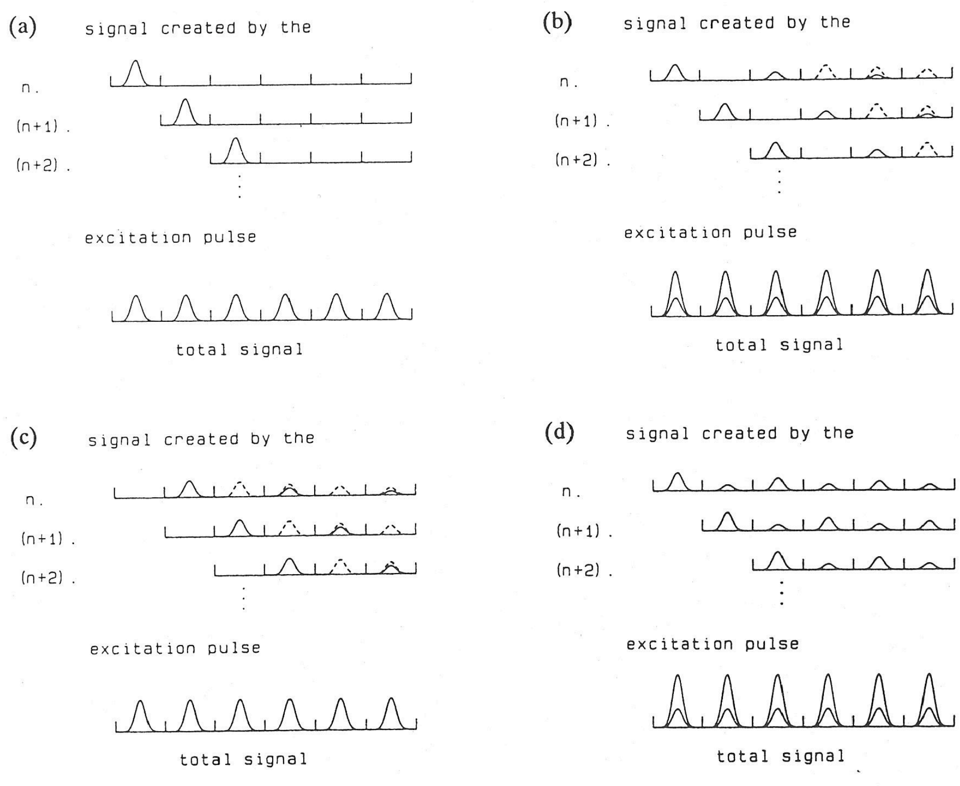

Figure 7. (a) Spoiled FLASH, (b) refocused FLASH, (c) CE-FAST, and (d) FISP as a superposition of refocused signals as shown in Figs. 4 and 5. The bottom line in each diagram gives the total signal acquired in a steady state. For the refocused FLASH and FISP, this signal is the sum of the direct gradient echo with refocused components, whereas spoiled FLASH and CE-FAST signals consist of pure direct gradient echoes or pure refocused signals, respectively. The direct echo contribution in (b) and (d) is shown as the smaller trace under each signal.

(a) (b) (c)

Figure 8. (a) Spoiled FLASH, (b) refocused FLASH, and (c) CE-FAST images of a transverse section through the brain of a healthy volunteer (5 mm slice thickness). Different experiment parameters were used to give representative results for each sequence. The spoiled FLASH image was acquired with $\alpha = 40°$, $t_r = 50$ ms, for (b) $\alpha = 70°$ and $t_r = 50$ ms was used, (c) was acquired with $\alpha = 70°$, $t_r = 28$ ms. For (a) and (b) no averaging was used; (c) was acquired with eight averages.

GENERAL CONSIDERATIONS

As yet we have discussed the two extremes of highly periodic pulse sequences, namely the multi-echo experiment and gradient-echo sequences. Other commonly used MR sequences are not quite as periodic as these two extreme cases. From the discussion of the partial saturation spin-echo sequence, I shall in the following try to derive some general rules about the occurrence of echoes.

Figure 7 gives the extended-phase graph for a repetitive spin-echo experiment. The experiment consists of an excitation period between the excitation pulse and the refocusing pulse and a signal generation period when the spin echo is acquired. The gradient has been assumed to be more realistic in that it is applied in discrete portions rather than continuously. The gradient after data acquisition can of course be chosen arbitrarily without any effect on the primary echo formation.

Some useful observations follow from this phase graph:

- As a consequence of the difference in the timing and the gradient effects between the excitation and the acquisition period, different echo formation pathways E_2, E_3, etc., will arise in addition to primary echo formation. One possible refocusing pathway for the formation of E_2 is given by the stimulated-echo mechanism of pulses 1-2-1, which is then refocused by pulse 2. Because the nominal value of 1 is optimal also for stimulated-echo formation, the signal contributions of this refocusing pathway alone will be quite considerable, even if reasonably good pulses with near rectangular profiles are used. The time difference between E_1 and E_2 will be given by the amount of the dephasing in the excitation period. If the gradient in the excitation period is strong enough that the signal is generated in the center of the acquisition time, then E_2 will fall onto the edge of the acquisition window. If the primary echo E_1 is acquired in the first half of the acquisition period, which is frequently done to minimize the echo time for a given gradient strength and switching time, then E_2 will even fall into the acquisition window and lead to image artifacts by interference. Because the stimulated-echo mechanism is insensitive to all attempts to use spoiling gradients after acquisition to delete this signal contribution, one is strongly advised to use a strong dephasing gradient during the excitation period to push E_2 as far outside of the acquisition window as possible.

- A second refocusing mechanism is given by the spin-echo formation, which refocuses all magnetization over two periods. This will lead to refocused signal contributions even if perfect 90° and 180° pulses are used. Just as in a refocused FLASH experiment, this refocused magnetization can be used to add some T_2 contrast to the images if each phase-encoding step is refocused within the same acquisition period. If short repetition times are used to get T_1-weighted images, then the effect of the refocused signal contributions will lead to a decreased image contrast because the (T_2) contrast of refocused magnetization is opposite the (T_1) contrast because of partial saturation in most tissues.

- The signal E_1 is reached by refocusing pathways other than primary echo formation (the reader is invited to follow some of those in Fig. 7). Because some of these proceed *via* stimulated-echo pathways and because others are subject to refocusing in every second excitation period, they are exceptionally stubborn with respect to spoiling gradients.

- A potentially important signal contribution arises from the free induction decay (FID) generated by the refocusing pulse. Although with a unipolar read gradient, the echo maximum will be outside the acquisition window, the signal intensity from the tail of this FID can be large enough to give rise to observable and disturbing artifacts.

All of these contributions of echoes other than the primary echo naturally become more important when short repetition times are used where more and more possible refocusing pathways fall within the lifetime of the signal, which is determined by T_2 and T_1. Such spin-echo images with short repetition times are conventionally used as pilot scans for patient and scan positioning.

This discussion can easily be extended to a general procedure for the design of arbitrary multi-pulse sequences: To get a grip on all relevant signal contributions, a phase graph over several excitation cycles of the sequence should be drawn. This need not follow the actual time course of the experiment; only the total effect of gradients between two pulses is relevant for the creation of different refocusing pathways. It might be advisable to create separate phase graphs for the read and slice-selection gradients. Whenever refocusing occurs in both graphs simultaneously, a significant spurious echo must be expected. Because this condition is met by the proper signal, it also will be true for practically all spurious signals.

The discussion of the simple spin-echo experiment has shown that spurious echoes will be generated in distinct patterns: They will occur at distances from the proper signal that are determined by the effect of the gradient between any two pulses or by the linear combinations thereof. Phase graphs like the one shown in Fig. 9 apply to any pair of pulses in a repetitive experiment. The amplitude of these echoes will depend on the flip angles of the pulses used. 90° pulses will favor stimulated-echo pathways; 180° pulses will refocus magnetization. But beware: For signals that are far enough "off-resonance," a 180° pulse is not a bad excitation pulse.

After selecting the most relevant signal contributions, it is possible to deal with them by the following procedures:

- The integral I_{nm} over the read-out gradient between any two adjacent pulses and any linear combination thereof should be large. If the effect of the read gradient over the acquisition window is smaller than any of these, then all spurious echoes will fall outside the acquisition window. No image artifacts occur. This method of totally removing artifacts looks promising on paper, but in practice, huge gradients are necessary to meet this innocent-looking condition. Consequently, this is not a practical procedure, especially for very fast sequences, where the time necessary for creating large gradient effects will increase the acquisition time significantly.

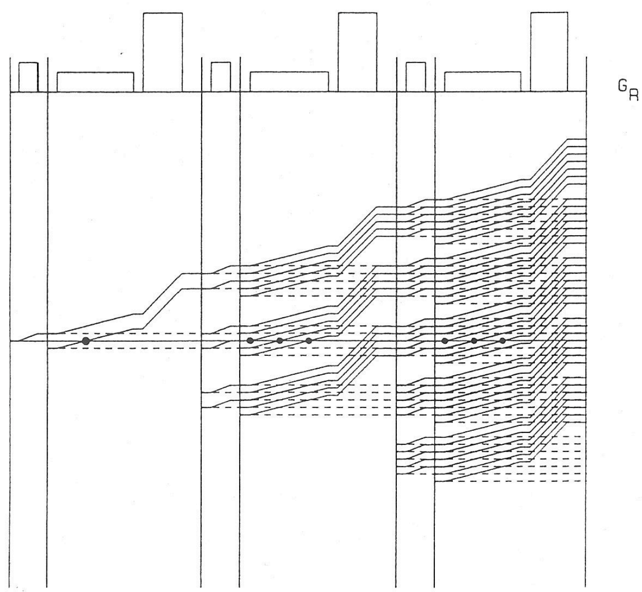

Figure 9. Extended-phase Graph for a Repetitive Spin-echo Sequence. Only pathways arising from the first pulse are shown; multiple spurious echoes (small circles) are generated after very few repetition periods apart from the proper signal (large circles at left).

- The slice-selection gradient can be used to spoil an unwanted signal, especially if it is applied as a nonperiodic spoiling gradient as described for the spoiled FLASH sequence.

- The phase-encoding gradient can be used to separate the signal from spurious echoes in the image plane. If the artifact is low, it can be smeared out all over the image using appropriate phase-encoding gradients. Another possibility is to give the spurious signals little or no phase encoding, such that the artifacts will be focused and can be wiped into an unused corner of the image either by choosing an appropriately periodic phase-encoding scheme or by appropriate phase cycling of the rf pulses.

- Phase-cycling schemes for the rf pulses can be chosen such that the spurious signals will cancel. A simple example is the use of alternating phases for the refocusing pulse in a spin-echo experiment. If two scans are averaged, then the FID produced by the 180° pulse will vanish. The formulas for the signal phase as a function of the pulse phase for excitation and refocusing are given by Eq. [3] and Eq. [4]; the phase of a stimulated echo is given by

$$\psi_{STE} = \psi_3 + \psi_2 - (\psi_1 + 90°) \qquad [5]$$

The phases of the spurious echoes can be calculated by repetitive application of these formulas.

A disadvantage of this procedure is, of course, that more than one acquisition per phase-encoding step is necessary. In addition, a phase cycle of two or four will not be sufficient to cancel all spurious signals. Therefore, such phase cycles are of restricted value for imaging experiments. They are, however, important for spectroscopy where many repetitions are used. This allows the application of quite satisfactory phase-cycling schemes, which can be used to remove spurious echoes by cancellation or by some other means.

- Last but not least, rf pulses that generate strong linear phases across the slice can be used to dephase all signals except the proper one, as shown for the spoiled FLASH sequence. This appears to be an extremely powerful concept, because it avoids additional gradients and delivers "clean" signals in a very elegant way. The use of this approach in gradient-echo imaging has already been mentioned. Its application in localized spectroscopy appears even more promising.

CONCLUSIONS

This two-part article has become more lengthy than originally expected, which proves either that the author is rather talkative or that the world of echoes is richer and more interesting than one would expect from waving one's arms around one's axis, as described in Part I. As soon as the principle and manifold ways of echo formation are understood, it is easy to recognize that echoes arise everywhere — at places expected and unexpected. Extended-phase graphs can be used to identify possible spurious echoes. In addition, they provide a simple means to discuss the T_1 and T_2 contrast of the proper signal, especially when more than one refocusing pathway leads to signal generation. The matrix formalism for the calculation of exact signal amplitudes allows the calculation of signal intensities even after hundreds of pulses.

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