Gradient Echo Imaging

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INTRODUCTION

The fundamental signal generation in magnetic resonance imaging (MRI) sequences is based on the principle of either spin echoes (1,2) or gradient echoes (3–7) or a combination of the two. This course elucidates concepts and basic properties of gradient echo methods with a special focus on fast gradient echo sequences.

Gradient echoes - also called gradient-recalled echoes (GRE), gradient-refocused echoes or field echoes – are the basis of many applications on modern MRI systems (5–7). Figure 1 shows the basic principle of such a GRE, where the free induction decay (FID) after radio-frequency (RF) excitation is measured.

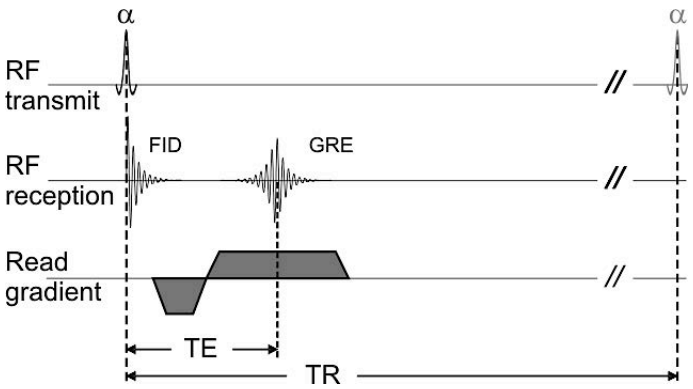

Fig. 1: *Simplified diagram for the basic principle of a* gradient echo*: after RF excitation with a flip angle α, a dephasing / rephasing reversal of the (read) gradient generates the so-called 'gradient echo' that is measured by the MRI system at the echo time TE. Subsequently, the RF excitation is repeated with the repetition time TR. For reasons of simplicity, further imaging gradients such as slice selection or phase encoding were neglected.*

The most striking differences in Fig. 1 compared to a conventional spin echo sequence are, generally, the lacking refocusing pulse, the typically low excitation flip angle of α<90°, and the read gradient reversal.

A GRE sequence as outlined in Fig. 1 represents a quite "simple" form, since the sketch shall imply a repetition time TR that is much longer than T_1 and T_2. Therefore, the transverse magnetization (M_{xy}) decays completely and the longitudinal magnetization (M_z) fully recovers until the next excitation pulse. Then, M_{xy} and, hence, the signal behavior and image contrast is dominated by the frequently introduced transverse decay time T_2*:

$$\frac{1}{T_2^*} = \frac{1}{T_2} + \frac{1}{T_2'} \quad . \qquad [1]$$

As defined in Eq. [1], T_2* is a combination of the tissue characteristic transverse relaxation time (T_2) and a relaxation term, T_2', which reflects signal decay due to static field inhomogeneity and susceptibility effects. Note that it is always T_2*<T_2.

The exhibited T_2* contrast of such a simple (and slow) GRE sequence is dependent on the echo time TE as follows:

$$SI(TE) = SI_0 \cdot \exp\left(-TE/T_2^*\right) \quad . \qquad [2]$$

It should be noted here that the introduction of T_2' and, thus, T_2* in Eq. [1] is an approximation which performs often well in properly describing the observed signal behavior depicted by Eq. [2].

It has always been a necessity in MRI to actually shorten the acquisition time TA that is directly proportional to TR, which leads under certain circumstances to a *dynamic magnetization equilibrium* or *steady state signal* due to the incomplete T_1 and T_2 relaxation until the next RF excitation pulse.

STEADY STATE FREE PRECESSION SEQUENCES

As originally introduced in 1958 by Carr (8) for NMR spectroscopy, a dynamic equilibrium or steady-state in the magnetization can be established by a train of RF excitation pulses interleaved by periods of 'free precession', commonly referred to as *steady state free precession* (SSFP) (see Fig. 2).

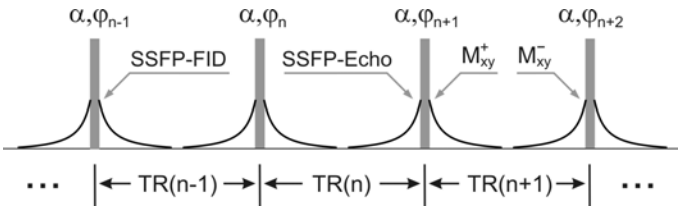

Fig. 2: *Train of equidistant RF excitation pulses with a repetition time TR << T_1,T_2, interleaved by free precession periods. Depending on the RF pulse properties and gradient switching patterns, different steady states can be established.*

The quick succession of RF pulses prevents the magnetization from returning to thermal equilibrium and each RF pulse therefore acts on both remaining transversal and longitudinal magnetization generating rather complex patterns of spatial magnetization distribution even after a few RF pulses. Nevertheless, the magnetization between consecutive excitation pulses can reach a dynamic equilibrium, i.e. a steady state, if the following conditions are fulfilled, see (9,10):

- The dephasing from gradients (G) within TR, TR itself and the flip angle (α) must be constant.
- The phases (Φ) of the RF pulse must satisfy the equation: $\Phi_n = a + b \cdot n + c \cdot n^2$.

Transition to steady state from thermal equilibrium (or any other magnetization prepared state) is completed after $5*T_1$, however, this is frequently not an acceptable waiting time for a fast imaging method. As a result, several preparation methods have been proposed to facilitate, enhance or smoothen this transition. In the following, we will assume that a steady state could have been established after sufficient RF pulses.

The class of fast GRE sequences that are based on the dynamic equilibrium effect of SSFP are commonly called SSFP sequences. Generally, the measured signal from such SSFP sequences will depend on relaxation ($T_{1,2}$) and on diffusion or flow effects, but also becomes a function of the repetition time (TR), the echo time (TE), the flip angle (α) and the RF pulse phase increment ($\phi_n - \phi_{n-1}$), and of the gradient switching pattern.

SSFP CLASSIFICATION

Overall, GRE sequences can be broadly classified as *incoherent* or *coherent* depending on whether or not any remnant transverse magnetization is spoiled prior to the next RF pulse (see Table 1). Since its introduction more than half a century ago, the use of the SSFP signal has become increasingly popular for imaging and a large number of SSFP imaging methods have been described so far under a sometimes confusing array of acronyms, e.g., see Refs. (5–7,11). For clearness, we will stick to the following generic SSFP nomenclature:

- The term '*SSFP'* embeds all steady state sequences and variants thereof since it just indicates the most basic SSFP principle as introduced by Carr.
- The term '*balanced*' is used to indicate that all gradient moments are fully rephased prior to the next excitation pulse.
- The term '*nonbalanced*' is used to indicate the presence of some crusher gradients, i.e. dephasing gradient moments, prior to the next excitation pulse.
- The term '*spoiled'* is used to indicate that the transverse magnetization components can be assumed to be zero before the next excitation pulse, i.e., there will be no contribution of transverse magnetization components to the signal.
- The term '*FID*' or '*Echo*' in combination with SSFP is used to indicate whether the signal refers to the transverse magnetization just after or before the RF pulse (see also Fig. 2).

For common sequence acronyms expressed in generic terms, see Table 1.

steady state free precession (SSFP)	GRE	Generic	Siemens	Philips	GE
	incoherent	spoiled SSFP	FLASH	T1-FFE	SPGR
	coherent	SSFP-FID	FISP	FFE	GRASS
		SSFP-Echo	PSIF	T2-FFE	--
		dual echo SSFP	DESS	--	--
		balanced SSFP	TrueFISP	balanced FFE	FIESTA
		constructive interference SSFP	CISS	--	FIESTA-C

Table 1*: Incoherent and coherent GRE methods with generic nomenclature and corresponding commercial acronyms for the major scanner manufacturers.*

SSFP SEQUENCES AND CONTRAST

For incoherent SSFP imaging, the contribution from any residual transverse magnetization prior to the next excitation pulse is assumed to be zero, or *spoiled*. As a result, spoiled sequences show a pure T_1 contrast, and the steady state signal immediately after the excitation pulse is given by

$$M_{xy} = M_0 \frac{1 - E_1}{1 - E_1 \cos\alpha} \sin\alpha \quad , \qquad [3]$$

where $E_1 := \exp(-TR/T_1)$. Equation [3] is also known as the "Ernst equation" (5–7,11). However, the tricky part is to actually get rid of all the transverse magnetization components prior to the next excitation pulse or to find a clever method to avoid any significant contribution in subsequent repetition periods.

The simplest way of efficient spoiling is a long enough waiting time. For repetition times of about three times T_2, the transverse magnetization has decayed sufficiently; however, in the light of a minimization of both TR and TA (c.f. beginning) this approach is not very practical and a different spoiling strategy has to be used.

An elegant and rather efficient approach for spoiling transverse magnetization is to adapt the phase of the RF excitation pulse Φ in every TR interval (c.f. Fig. 3a) to have a linear phase increment according to the formula

$$\phi_{\mathrm{n}} - \phi_{n-1} = \psi_0 + n \cdot \psi \quad . \qquad [4]$$

This method - commonly referred to as RF spoiling (10,12,13) - generates quadratically increasing phase offsets in the residual transverse magnetization. A proper choice of ψ (e.g., ψ = 50° or ψ = 117°) leads to RF phase cycling conditions that lead to a pure T_1-weighted signal, i.e., to a near complete destructive interference of all residual transverse magnetization components.

For SSFP, the dephasing from gradients must be constant and phase encoding gradients need to be rewinded - a possible sequence diagram for the (RF) spoiled SSFP is shown in Fig. 3a. Asides, RF spoiling always comes with crusher gradients - sometimes also (improperly) called spoiler gradients.

Figure 3 provides a general overview of four major types of SSFP sequences. In fact, all four major types (RF spoiled SSFP, SSFP-FID, SSFP-Echo, balanced SSFP) can be regarded as a variant from a generic SSFP sequence. Their sole difference is whether the particular type of SSFP employs RF spoiling and crusher gradients in each case (c.f. Table 1); however, the resulting signal behavior and image contrast is different.

Figure 3a displays the already discussed RF spoiled SSFP sequence (including a crusher gradient in the readout direction) that facilitates a dominant T_1 contrast. Omitting RF spoiling but keeping the application of crusher gradients prior to the next excitation pulse, leads to the gradient-spoiled (i.e., also nonbalanced) SSFP-FID sequence (Fig. 3b).
A time-reversal of the readout gradient as displayed in Fig. 3c leads to the (gradient-spoiled) SSFP-Echo sequence.

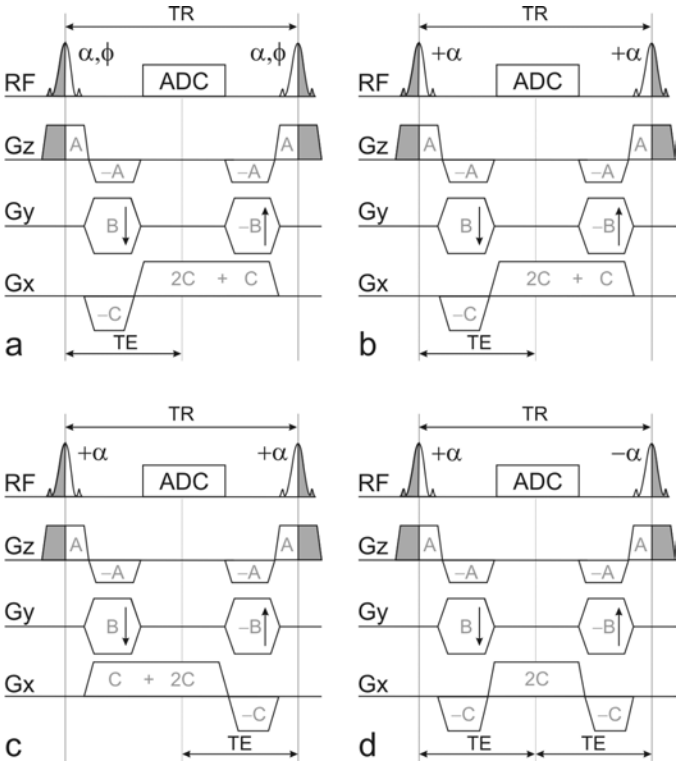

***Fig. 3**: Possible SSFP sequence diagrams for: (a) spoiled SSFP, using crusher gradients (along Gx) in combination with a quadratic RF phase increment (ϕ), (b) SSFP-FID, using a coherent RF pulse phase (ϕ = const.) in combination with a crusher gradient (along Gx), (c) SSFP-Echo, using a coherent RF pulse phase (ϕ = const.) in combination with a reversed crusher gradient, and (d) balanced SSFP, using a coherent RF pulse phase (ϕ = const.) in combination with a fully balanced gradient switching pattern (no spin dephasing within any TR!).*

Finally, balanced SSFP refers to an acquisition scheme as given in Fig. 3d, where all gradients have a zero net area within any TR (= balanced); and any residual phase accruals within any TR are therefore closely related to field inhomogeneities. As a result, balanced SSFP is prone to off-resonances that can lead to prominent signal voids or banding artifacts in regions of strong susceptibility variations and with poor shimming.

It is evident that except for the spoiled SSFP acquisition all other SSFP sequences show besides a T_1 contrast also some T_2 contrast due to the contribution from unspoiled transverse magnetization components. To leading order, the nonbalanced and balanced SSFP signal depends on T_1/T_2, which generally leads to a prominent contrast between tissues and fluids (Fig. 4).

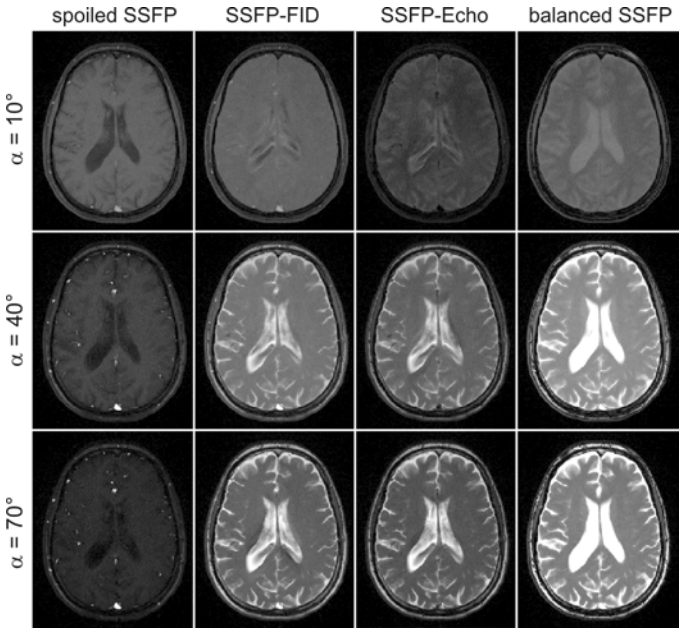

Fig. 4*: 2D axial brain images of a normal volunteer using spoiled, nonbalanced and balanced SSFP imaging methods for low, moderate, and high flip angle. For all sequences, a TR = 6ms and a TE = 3ms was used.*

MAGNETIZATION PREPARATION & APPLICATIONS

GRE sequences are often combined with magnetization preparation to obtain dedicated contrasts and to increase the variety of MRI applications. Furthermore, an important reason for using GRE (SSFP) based MRI sequences is the ability of acquiring 2D and 3D volumes in short experiment times of a few minutes down to less than a second. In the following, some important applications are selected.

Dynamic Imaging & CINE Imaging

Modern rapid GRE approaches allow time resolved measurements of dynamic processes. One example is a possible bolus tracking during contrast enhanced MRI based angiography (see also 'angiography' below).
Although MRI acquisitions are generally too slow to generate a "direct movie" of the beating human heart, e.g., a balanced SSFP sequence can be synchronized with the cardiac cycle via ECG triggering. Then, only segments or parts 1/N of k-space data are acquired at defined points of time within the cardiac cycle. After N cardiac cycles the data is measured completely and a sequence of images along the cardiac cycle can be reconstructed. The image data set can be viewed as a CINE loop ("movie") to reveal the dynamics during an averaged cardiac cycle. This approach is referred to as *CINE imaging* (14,15).

Inversion Recovery Preparation

For very fast SSFP approaches with short TR and low flip angles, the resulting signal intensity diminishes more and more and the inherent contrast becomes rather poor. A favorite approach to reinstate again stronger T_1 weighting in RF spoiled SSFP sequences is to use an initial 180° inversion pulse (inversion recovery preparation). Furthermore, in dependence of the inversion time TI, the signal intensity of certain tissues with a specific T_1 can be enhanced, suppressed, or even nulled. Particularly for T_1 weighted 3D imaging of the entire head such a magnetization prepared rapid gradient echo (MPRAGE) approach (16) has become popular in recent years.

Magnetization Transfer Preparation

Neglecting fat tissue for the moment, conventional clinical MRI sequences are only able to acquire signal of the "free water"; i.e., only from ^{1}H nuclei (protons) that are bound to water molecules that are free to move in a given tissue (17). Although they are excited, protons that are bound to macromolecules cannot be observed, because their T_2 is too short (c.f. NMR spectroscopy / MRS !). However, these protons are much more sensitive to off-resonant excitation, i.e., RF excitation pulses which are off the Larmor frequency. Excited macromolecular protons transfer their excitation to the "free protons". This effect is called *magnetization transfer* or *saturation transfer*. Hence, adding an off-resonant excitation pulse as a magnetization preparation to a GRE sequence will pre-excite tissues with a high macromolecular content. As a result, these tissues will display reduced signal intensity (-> MT effect).
MT preparations are usually combined with RF spoiled SSFP sequences.

Inflow Enhancement & Angiography

In GRE imaging the signal intensity of static tissue will be partially saturated in accordance with the steady state conditions. In contrast, blood that flows into the imaging slice or volume still carries "fresh", i.e., unsaturated magnetization – a signal enhancement occurs. Thus, the signal of blood is bright compared to the surrounding static tissue. This effect is exploited in time-of-flight (TOF) imaging (11,18), a non-invasive MR angiography (MRA) method without the administration of any external contrast agent. TOF imaging is usually based on fast RF spoiled SSFP sequences, since compared to SSFP-FID and balanced SSFP sequences they also suppress the bright signal from cerebrospinal and other fluids due to strong T_1 weighting.
RF spoiled SSFP sequences are also used for another family of MRA approaches: contrast enhanced MRA (CE-MRA, (11,19)). However, the mechanism of signal enhancement in the vessels differs. CE-MRA methods rely on externally administered contrast agents that

reduce the T_1 relaxation time in their vicinity. Thus, the RF spoiled SSFP sequences provide a means for fast T_1 weighted acquisitions.

Fat-Water-Imaging

An important consequence of the missing refocusing pulse in a GRE sequence is the chemical shift induced dephasing of magnetization that is not refocused anymore. Due to the natural Larmor frequency difference between water and fat of approx. 220Hz at 1.5T or 440Hz at 3T, their corresponding magnetization vectors experience a time-dependent phase difference. Particularly, at echo times TE = 4.4ms, 8.8ms, … water and fat signal are in-phase (1.5T), the GRE sequence acquires a so-called *in-phase image*. At echo times TE = 2.2ms, 6.6ms, … water and fat signal are opposed-phase (1.5T), the GRE sequence acquires an *out-of-phase* or *opposed phase image*. The latter images display dark rims due to signal voids in voxels with a similar content of fat and water.

Dedicated in-phase and out-of-phase images can be processed to obtain images that only display signal either from aqueous tissue ("water image") or from fatty tissue ("fat image"). This method is also known as "Dixon Imaging" (11,20).

KEYWORDS

Gradient echo (GRE), steady state, steady state free precession (SSFP), spoiling, nonbalanced, balanced.

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