Gradient echo or T2* weighted imaging: An introduction and some clinical applications

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By 1946 Bloch and Purcell and co-workers captured the essence of the behavior of an atom with a non-zero magnetic moment situated in a magnetic field. Their contributions to the field from these first papers was precocious and even 70 years later their work carries modern significance.

Although quantum physics was needed to formulate the experimental and theoretical results, the final impact rests on the very simple relationship given by the Larmor equation relating the frequency of rotation or precession frequency ($\omega$) and the local magnetic field (B) via $\omega = \gamma B$. 
Visualizing Quantum Mechanics

Thin slice T1 weighted post contrast 3D MRI of the brain
Basic MRI concepts

\[ \omega = -\gamma \ast B \]

- \( B \) the applied magnetic field
- \( \gamma \) the gyromagnetic ratio for the proton
- \( \omega \) the magnetic resonance frequency
A cylindrical object in a constant magnetic field.

Its bulk magnetization lies parallel to the main field.
Applying an oscillating rf field along the x-axis rotates $M_0$ from the z-axis into the transverse plane so that it lies along the y-axis.

At this time, the rf field is turned off and the transverse magnetization precesses about the main field $B_0$. 
The precessing transverse component induces a voltage in a coil which is placed so that it sees a changing magnetic flux.
GRADIENT ECHOES AND 1D IMAGING
The Larmor equation

$$\omega = -\gamma \times B$$

$$\phi = \omega t$$

$$\phi = -\gamma B t$$

For hydrogen,

$$\gamma = \gamma / 2\pi = 42.6 \text{ MHz/T}.$$
By purposely adding a gradient field $G$, we can spatially distinguish the spins by their frequency content since now

$$\omega(x) = \gamma (B_0 + Gx)$$
Applying a dephasing gradient followed by a rephasing gradient causes the spins to first run away from each other and then to turn around and run back creating a gradient echo.
The MR signal as a Fourier transform

\[ s(t) = \int dx \rho(x) \exp(-i \gamma G x t) \]

where \( \rho(x) \) is the spin density and using \( k = \frac{\gamma G t}{2\pi} \) as the spatial frequency we find

\[ s(k) = \int dx \rho(x) \exp(-i 2\pi k x) \]

\[ \rho(x) = \int dk s(k) \exp(i 2\pi k x) \]
Spatial encoding using a negative followed by a positive gradient to produce a gradient echo signal
Typical Gradient Echo sequence

RF

$G_z$

$G_x$

$G_y$

Acq.

TE

TR

2D imaging: K-space trajectory
Describes recovery of magnetization toward the equilibrium state

\[ M_z(t) = M_0(1 - e^{-t/T_1}) \]

where \( t \) is the time between the rf pulse (with \( \theta = \pi/2 \)) and the readout interval.
Spin-lattice interaction
Describes the regrowth rate of $M_z$ toward the equilibrium value of $M_o$
Definition: T1 is the time needed for $M_z$ to recover from 0 to $(1-e^{-1})$ of the equilibrium value $M_o$

<table>
<thead>
<tr>
<th>Tissue</th>
<th>$T_1$ (ms)</th>
<th>$T_2$ (ms)</th>
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<tbody>
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<td>gray matter (GM)</td>
<td>950</td>
<td>100</td>
</tr>
<tr>
<td>white matter (WM)</td>
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<td>80</td>
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<td>1200</td>
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T2 relaxation

- Free Induction Decay (FID)

\[ M_{xy}(t) = M_{xy}(0)e^{-t/T_2} \]
Spin-Spin interaction
Describes decaying rate of $M_{xy}$
Practically, $T_2^*$ decay, rather than $T_2$ decay, is observed in FID due to field inhomogeneity

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**T2* relaxation**

- \[ \frac{1}{T_{2}^*} = \frac{1}{T_2} + \frac{1}{T_{2}'} > \frac{1}{T_2} \]

- Combined external field inhomogeneity effects (T2') and thermodynamic effects (T2)

- T2' origins:
  - External: B₀ field, boundaries, partial volume effect
  - Internal: chemical shift, microscopic motion (diffusion, capillary blood)
  - Effective spatial scale: ~voxel size to molecular size
T2* relaxation

T2 decay (ideal)  T2* decay (actual)
T2* Weighted Contrast

- Directly collect the FID using long echo time TE:
  \[ M_{xy}(TE) = M_{xy}(0)e^{-\frac{TE}{T2^*}} \]

- Typical T2*W sequence:
  - Gradient Refocused Echo (GRE)
  - Echo Planar Imaging (EPI)
Dephase from field inhomogeneity (or background gradient) can be refocused by a $\pi$ rf pulse:

$$M_{xy}(TE) = M_{xy}(0)e^{-\frac{TE}{T2}}$$

(valid at $TE=2t_d$)

- **T2W sequence:**
  - Spin Echo (SE)
  - Fast Spin Echo (FSE)
GESFIDE SEQUENCE

Wei Feng, PhD
The GESFIDE Sequence\textsuperscript{1}

\begin{figure}
\centering
\includegraphics[width=\textwidth]{gesfide_sequence.png}
\caption{Signal decay over time for GESFIDE sequence.}
\end{figure}

\textsuperscript{1} Ma and Wehrli, 1996; Yablonski and Haacke, 1997
Signal magnitude at the first echo as a function of the flip angle.

\[ TR = 250 \text{ ms} \]
\[ TE = 4.6 \text{ ms} \]
\[ T1 = 719/1165/3300 \text{ ms} \ (WM/GM/CSF) \]
(a) T1W image at first echo; (b) SWI image at TE=21.04ms; (c) SWI image at TE=79.66ms (averaged over 5 echoes from 74.18ms to 85.14ms; (d) Same as (c) but generated with conventional multiecho SWI; (e) MRA at spin echo; (f), (g) and (h): R2, R2* and R2’ maps. Note (b), (c), (d) and (e) were generated with minimum intensity projection over 8 slices whose center slice was aligned with the slices shown in (a), (f), (g) and (h).
Approach to Steady State

\[ \hat{p}(\theta, T_E) = \rho_0 \sin\theta \frac{(1 - E_1)}{(1 - E_1 \cos\theta)} e^{-T_E/T_2^*} \]

\[ E_1 \equiv e^{-T_R/T_1} \]

\[ \theta = 10^\circ \]

\[ T_R = 40 \text{ ms} \]
\[ \rho^* (1 - E1) E2^* \] where \( E1 = \exp(-TR/T1) \) and \( E2 = \exp(-TE/T2^*) \)
Why choose 90° flip angle?

It limits the total T1 recovery and for small TR the signal only recovers to TR/T1. If TR = 10ms and T1 = 2000ms, the signal is only 0.5% of its maximum value.

Now if we use a smaller flip angle, then we find the signal becomes:

\[
\hat{\rho}(\theta, T_E) = \rho_0 \sin \theta \frac{(1 - E_1)}{(1 - E_1 \cos \theta)} e^{-T_E/T_2^*}
\]

At the optimal flip angle (also known as the Ernst angle), given by \(\cos \theta = E_1\), we find the peak signal is roughly 0.5sqrt(2TR/T1) or 5% of the total (that is 10 times bigger).
Signal behavior at 1.5T for a TR of 20msec

\[ \rho(\theta, T_E) = \rho_0 \sin \theta \frac{(1 - E_1)}{(1 - E_1 \cos \theta)} e^{-T_E/T_2^*} \]

\[ E_1 \equiv e^{-T_R/T_1} \]
Example data from a 3D gradient echo sequence with short TE and short TR = 20ms at 1.5T.

Small flip angles generate a spin density weighted image while large flip angles generate a T1 weighted image.
Structural Imaging:

- T1 3D MPRAGE
- 1.0 x 1.0 mm² in plane resolution
- 1 mm slice thickness
- 176 slices
- TE 5 ms
- TR(total) 2500 ms
- Flip angle: 12°
- Bandwidth 200 Hz/Px
- Scan time: 10:32 min
Effect of Bandwidth on MR image quality

Bandwidth 110 Hz/Px

Bandwidth 473 Hz/Px
Local field distortions at short TEs

**TE=10ms**

- 0.7x0.7x1.4mm³
- TR=14ms, Flip angle:12°
- Bandwidth=473 Hz/Px

**TE=2.5ms**

- 0.7x0.7x1.4mm³
- TR=12ms, Flip angle:12°
- Bandwidth=473 Hz/Px
Magnetic Resonance Imaging: Physical Principles and Sequence Design
How to choose the right resolution and echo time

\[ N_{\text{shift}} \Delta t = \left[ \frac{G'}{(G + G')} \right] \text{TE} \]

\[ \Delta x_{\text{nsl}} = \frac{12}{(\delta B_0 \text{TE}) \text{mm}} \]

with TE in ms and \( \delta \) is in ppm/mm

nsl = no signal loss
Sagittal images: 0.6mm x 0.6mm x 1.2mm

Original image  Sliding window filtered

Note this is not the happy campers structure but rather the Hippocampus
MRA

resolution:
0.5mm x
0.5mm x
1.0mm
1mm resolution vertically

SNR = a, CNR = b

0.25mm resolution vertically

SNR = a/2, CNR = 2b
High resolution MR angiography
Small arteries around 250 microns are beginning to become visible even without a contrast agent (0.5mm isotropic resolution).
Cadaver brain imaging of arteries using an injection technique
Thalamic arteries
Gradient echo imaging can be used for a variety of contrast types
It is fast with good SNR and high resolution
It can be used for SWI and MRA
It is perhaps the most robust method to use from field strength to field strength and manufacturer to manufacturer