TALK TITLE Gradient Echo Imaging

SPEAKER NAME Daniel B. Ennis, Ph.D. (ennis@ucla.edu), Department of Radiological Sciences, David Geffen School of Medicine, University of California, Los Angeles

TARGET AUDIENCE Clinicians, technologists, and scientists interested in the fundamental physics and clinical applications of gradient echo imaging.

HIGHLIGHTS:
- Provide a basic understanding of gradient echo (GRE) imaging.
- Explain proton-density, T_1-, and T_2*-weighted GRE images.
- Describe a range of clinically useful and emerging GRE applications.

OBJECTIVES – To define the basic physical principles of GRE imaging; provide the context for understanding image contrast in GRE imaging; and demonstrate current and emerging clinical applications of GRE imaging.

INTRODUCTION - Gradient echo (GRE) imaging has a wide range of clinically useful applications[1]. In particular, GRE imaging is frequently used in the clinic for anatomical/structural imaging and imaging dynamics for diseases that span all organ systems and numerous pathologies. In general, an *advantage* of GRE imaging is the fact that it is regarded as "fast" owing to the generally short repetition times (TRs). As a consequence, GRE imaging is especially useful for: proton density-weighted, T_1-weighted and T_2*-weighted imaging; capturing physiologic motion; perfusion imaging; angiography; real time imaging; and for true 3D imaging. A notable *disadvantage* is the fact that GRE imaging can't generally achieve T_2-weighted image contrast. This talk will also provide context by succinctly comparing and contrasting spin echo and GRE imaging.

THEORY - In general, a GRE pulse sequence consist of image contrast preparation pulses, spatial localization[2] (slice selection, phase encoding, and readout), and spoiling.

Image Contrast Preparation – Like many MRI sequences, GRE image contrast can be first manipulated through the use of inversion, saturation, T_2-prep, fat-saturation, and magnetization transfer pulses, which precede spatial localization (slice selection, phase encoding, and frequency encoding). Each of these techniques emphasizes a particular form of image contrast.

Slice Selection – The combination of an RF pulse and a slice-selection gradient excites, in general, a thin plane of spins 3-10mm thick. This is almost always followed by a slice-selection refocusing gradient to rephase spins in the through-plane direction, thereby significantly increase the overall echo amplitude. The correct choice of flip angle is important for achieving the desired image quality. For spoiled steady-state GRE imaging (i.e. FLASH, SPGR, or T1-FFE) there exists an optimal flip angle (α) for stationary tissues, called the Ernst angle, which can be calculated from the TR and T_1 of interest: $\alpha_{Ernst} = \cos^{-1}(\exp(-TR/T_1))$. If spins (e.g. blood) are flowing into the slice, then a higher α can be used to achieve higher signal values. This so-called "in-flow enhancement" makes blood appear bright and is a notable *advantage*. The inability to easily generate black-blood contrast is notable *disadvantage* of GRE imaging.

Phase Encoding – GRE imaging can be either 2D or 3D. In 2D imaging phase and frequency encoding along two orthogonal directions is used to encode image information within an excited slice. In 3D imaging the phase encoding is performed along two orthogonal axes (i.e. phase and slab/slice/partition) and frequency encoding is performed along a third mutually orthogonal axis. GRE imaging is particularly amendable to 3D imaging due to the typically short TRs, whereas spin echo imaging typically has much longer TRs. Compared to 2D imaging, 3D imaging can have an SNR *advantage*, which permits acquiring higher spatial resolution and/or thinner slices. Thinner slices are also possible because the excited 3D slab can be encoded with better slice resolution than is achievable with 2D slice-selective pulse. The *disadvantage* of 3D imaging is

the much longer scan time needed to acquire two phase encode directions.

Frequency Encoding – The frequency encoding gradient consists of a gradient pre-phasing lobe followed by a frequency encoding gradient. The pre-phasing lobe helps prepare the formation of the echo. The echo signal is generated and recorded during the application of the readout gradient and contains information (spatial location and contrast) about the object being imaged. Importantly, the magnitude of the readout gradient controls the speed of the k-space traversal. If the gradient has a large amplitude, then k-space is traversed quickly, therefore we need to sample more quickly (increase the bandwidth) to acquire the same number of k-space points. Because we acquired the data quickly (i.e. at high bandwidth) the data has a lower SNR. High bandwidth, however, also permits a shorter TE and TR because the frequency encoding gradient is shorter.

Spoiling –Gradient spoiling is a technique that uses a gradient pulse at the end of the TR to hasten dephasing of the transverse magnetization (M_{xy}) so that the net M_{xy} is very nearly zero at the end of the TR. This permits the use of very short TRs. Alterations of the slice-selective RF pulse's phase can also be used to promote signal spoiling, especially to enhance T_1-constrast.

METHODS – GRE imaging can be adjusted in innumerable ways to manipulate image contrast.

Image Contrast – GRE image acquisition can be adjusted to achieve proton density-weighting, T_1- weighting, or T_2*-weighting. This is easily accomplished by using a spoiled steady-state GRE sequence. Tissues with a higher proton density are brighter on proton density-weighted images, which can be obtained with a long TR (>100ms), short TE (<5ms), and small FA (<10°). Tissues with a short T_1 are bright in T_1-weighted images and are achieved with short TRs (<50ms), short TEs (<5ms), and a large FA (>30°). In distinction, tissues with a short T_2* are dark on T_2*-weighted images and are achieved with a long TR (>100ms), a long TE (>20ms), and a small FA (<10°). T_2* reflects both the intrinsic (irreversible) T_2 spin-spin relaxation and an additional reversible $T_2'(\propto 1/\Delta B_0)$ component that further hastens signal dephasing due to intravoxel field heterogeneities (B_0-field inhomogeneity and susceptibility variations). This means that T_2*-weighted GRE is more sensitive than spin echo imaging to the presence of hemorrhage due to the iron in blood products or to air-tissue interfaces (e.g. sinus or bowel air) both of which represent significant susceptibility differences compared to soft tissues. The reversible T_2' is refocused in a spin echo experiment by the refocusing pulse thereby regaining T_2-contrast.

APPLICATIONS – GRE imaging lends itself to many clinical applications due to its speed and flexibility.

Imaging Physiologic Motion – GRE imaging speed enables the imaging of transient and periodic physiologic motions. Transient motions are best imaged with real-time or sequential GRE images, whereas periodic events (especially cardiovascular events) can image the motion by synchronizing the image acquisition to the subject's ECG[3,4].

Perfusion Imaging – The speed of GRE imaging also lends itself to the qualitative and quantitative evaluation of tissue perfusion[5]. The off-label intravenous administration of a gadolinium based T_1 shortening contrast agent produces a transient "blush" of increased signal upon arrival of the agent to well perfused tissue. Poorly perfused tissue is conspicuous as a lack of transient signal enhancement. The characteristics of the apparent tissue enhancement can be used to estimate local tissue perfusion.

Magnetic Resonance Angiography – 3D GRE sequences can be sufficiently fast to permit the acquisition of high-resolution 3D images covering, for example, the chest in a single breath hold[6]. Such images are T_1-weighted and significantly benefit from the off-label intravenous administration of a gadolinium based T_1 shortening contrast agent, which enhances the ability to qualitatively and quantitatively evaluate cardiovascular anatomy.

Fat-Water Separated Imaging – Fat is chemically shifted by 3.5ppm compared to water (220Hz at 1.5T; 440Hz at 3.0T). This means that fat and water may be in-phase (pointing in the same direction) for one TE, but out-of-phase (pointing in opposite directions) for another TE. As a consequence manipulation of the TE can provide insight about the presence or absence of fat either qualitatively or quantitatively[7,8]. This can be useful for differentiating fatty-tissues from other bright tissues on T_1-weighted images.

Flow Sensitivity – A unique attribute of GRE imaging is the ability to encode the velocity of flowing spins in the phase of the MRI signal[9]. With proper correction for background off-resonance, eddy currents, Maxwell terms, and chemical shift effects phase contrast MRI (PC-MRI) becomes a useful quantitative imaging tool for measuring peak and average velocities flow rates, and total flows. To make a GRE imaging sequence quantitatively sensitive to flow we need to add flow-encoding gradients, which extended both the TE and TR. This, combined with the need to acquire two measurements, means the temporal resolution of PC-MRI is typically lower than other time-resolved GRE imaging sequences.

Quantitative Imaging – GRE imaging can also be used to perform quantitative T_1-mapping, T_2*-mapping, and fat-fraction mapping. These are emerging techniques that hold tremendous clinical promise for tissue characterization (i.e. parametric mapping) in a variety of pathologies[10].

DISCUSSION – As with many imaging sequence in MRI, GRE imaging is amenable to a great number of modifications, which provides both a unique strength and complexity. Importantly, GRE imaging is capable of proton density-weighted, T_1- weighted, or T_2*-weighted. The speed of GRE imaging (owing to the short TR) facilitates true 3D imaging and can be used for the most demanding high-speed imaging applications (e.g. real-time imaging).

CONCLUSION – GRE imaging is a cornerstone of clinical MRI because of its speed and flexibility.

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