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Improvements in MR imaging of solids through gradient waveform optimization

Marco L.H. Gruwel, Peter Latta, and Boguslaw Tomanek

Abstract: Magnetic resonance imaging (MRI) is known to provide a useful approach for the exploration of the chemistry and dynamics of a wide range of soft condensed materials. However, its application to solids has been limited to those materials with relatively narrow resonances. The time needed to obtain an image of a solid with a given resolution and signal-to-noise ratio (SNR) is directly proportional to the line width of the resonance. For MRI to become practical for the imaging of solids it will have to rely on the development and use of MR sequences that avoid the issues raised by line broadening of the resonance. In this paper we review the latest contributions towards MR imaging of solids from our laboratory, in particular, applications using optimized gradient waveforms. Acoustic noise reduction and SNR improvement obtained with modifications of the standard single-point imaging sequence are presented and discussed using examples.

Key words: MRI, solids, gradient shape, gradient ramp, sound pressure, signal-to-noise ratio.

Résumé : L'imagerie par résonance magnétique (IRM) est connue comme une approche utile pour explorer la chimie et la dynamique d'un grand nombre de matériaux condensés et mous. Toutefois, son application aux solides a été limitée aux matériaux avec des résonances relativement étroites. Le temps requis pour obtenir une image d'un solide avec une résolution et un rapport de signal à bruit (RSB) donnés est directement proportionnel à la largeur de la raie de résonance. Pour que l'IRM devienne pratique dans la production d'images de solides, il faudra développer et utiliser des séquences de résonance magnétique qui éviteront les problèmes liés à l'élargissement des raies de résonance. Dans ce travail, on présente une revue des plus récentes contributions de notre laboratoire à l'IRM des solides, en particulier aux applications utilisées pour optimiser le gradient des formes des ondes. On présente la réduction acoustique du bruit et l'amélioration du rapport signal à bruit obtenues par modifications de la séquence d'imagerie habituelle avec point localisé et on en discute à l'aide d'exemples.

Mots-clés : imagerie par résonance magnétique (IRM), solides, gradient de forme, gradient de forme, pression du son, rapport signal à bruit.

[Traduit par la Rédaction]

Introduction

Over the last decades magnetic resonance imaging (MRI) has become a major diagnostic tool in medicine and biomedical research. Nearly all applications involve the study of ^1H magnetization in soft water-containing tissues. Within these soft tissues, water mobility is more or less unrestricted, resulting in a typically narrow ^1H resonance and a corresponding long transverse relaxation time (T_2). Thus, the Hamiltonian best describing these spin systems consists of one major contribution, the Zeeman interaction.¹ Perturbations to the Hamiltonian usually do not have to be considered unless measurements are performed near interfaces, where

susceptibility effects from metal parts or air can be large.² Because of this high degree of water mobility, spatial information can be obtained using frequency encoding techniques.³ However, in solid materials the line width of the resonance is perturbed by large anisotropic interactions (approximately in the order of a few kHz), therefore, frequency encoding techniques result in images with poor resolution and sensitivity.

Although many different experimental approaches have been used to reduce unwanted line broadening, standard NMR line-narrowing techniques such as CPMAS, multi-pulse sequences, etc.,^{4,5} require additional attention to spatially dependent gradient interactions modulating image

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This article is part of a Special Issue dedicated to Professor Roderick E. Wasylshen. We are pleased to contribute this paper to the Special Issue recognizing the scientific career of Rod Wasylshen, particularly for his contributions to solid-state magnetic resonance.

Fig. 1. (A) Single point imaging (SPI) sequence showing RF and gradient sections. During RF excitation the gradient is kept stable (interval P) while a single datum point is collected at an encoding time t_p . One set of encoding gradients is shown and up to three gradients can be used to perform 3D spatial encoding. The time intervals R and P represent the gradient ramp and plateau time, respectively. (B) As in (A), except sinusoidal gradient ramps were used.

resolution. Also, these techniques are not suitable for larger objects and in vivo MR imaging of, for example, human bones or teeth. Emid and Creighton⁶ proposed a spatial encoding technique based on pure phase encoding in combination with the collection of one point of the free induction decay after RF excitation, the so-called single point imaging (SPI) technique. As only one point of the NMR signal is collected, any convoluting effects from time-dependent perturbations such as the chemical shift, quadrupolar effect, etc., will not be observed in the acquired data. Resolution and sensitivity of the image will thus only be determined by the strength of the applied gradient.⁷ In general, when the gradient switching time becomes longer than the longitudinal relaxation (T_2^*) of the sample, SPI will provide a superior signal-to-noise ratio (SNR) compared with frequency-encoding techniques. The only drawback of the SPI pure phase-encoding technique is the replacement of a single line of frequency-encoded data with individual phase-encode steps, each requiring a relaxation delay. However, for many solids spin-lattice relaxation times (T_1) are relatively short, allowing fast repetition of the phase-encode steps without the risk of signal saturation. As an example we will show images of human teeth in which the water T_1 is relatively short vide infra.

A typical SPI experiment consists of three gradients, one for each spatial dimension, as shown in Fig. 1A. As soon as the ramped gradient has reached its plateau value, a RF pulse is applied and after an encoding time interval t_p (~50–400 μ s) a single datum point is collected. After the gradients are ramped down and sufficient time for T_1 relaxation is given, the experiment can be repeated. Typically repetition times range from 1 to 20 ms. The signal intensity S from the spin density ρ of a voxel in an SPI image acquired with a repetition time TR is given by⁸

$$[1] \quad S_{\text{SPI}}(t_p) = \rho \exp\left(-\frac{t_p}{T_2^*}\right) \left[\frac{1 - \exp\left(-\frac{\text{TR}}{T_1}\right)}{1 - \cos \alpha \exp\left(-\frac{\text{TR}}{T_1}\right)} \sin \alpha \right]$$

where α is the RF flip angle. Thus, $S_{\text{SPI}}(t_p)$ has an optimum for $\cos \alpha = \exp\left(-\frac{\text{TR}}{T_1}\right)$. Transverse and longitudinal weighting of the image can be introduced through careful selection of t_p and TR. For comparison, the signal intensity in a standard frequency-encoded spin-echo experiment can be described by

$$[2] \quad S_{\text{SE}}(\text{TE}) = \rho \exp\left(-\frac{\text{TE}}{T_2}\right) \left[1 - \exp\left(-\frac{\text{TR}}{T_1}\right) \right]$$

with TE defining the echo time, and assuming that diffusion can be ignored. Thus, in the case where TR is long and (or) the RF flip angle α is small:

$$[3] \quad S_{\text{SPI}}(t_p) = \rho \exp\left(-\frac{t_p}{T_2^*}\right)$$

resulting in a T_2^* weighted spin-density image in SPI. Frequency-encoding experiments use typical echo times of ~10 ms, whereas SPI-encoding times typically are ~100 μ s. Gravina and Cory⁷ discuss these differences in more detail.

In general, MRI experiments generate acoustic noise owing to the rapid switching of the magnetic field ($\frac{dB}{dt}$) in the gradient coils, inducing strong Lorentz's forces causing mechanical vibrations in the gradient coils and cryostat. SPI, with its high gradient duty cycle and relatively long experiment time, can cause dangerous levels of mechanical vibrations. This, of course, sets a limit on the speed of the data acquisition and could severely hamper SPI applications on medical MRI systems because of acoustic noise exposure limits set by

regulatory bodies. To improve the efficiency of the SPI sequence and significantly reduce the mechanical vibrations, the single point ramped imaging with the T_1 enhancement (SPRITE) technique was introduced⁹ followed by later modifications.^{10–12} SPRITE uses gradual gradient switching (step-function-like), avoiding large gradient ramps between successive phase-encoding steps. However, because of the large currents flowing through the gradient coils, overheating can become an issue for long experiments. Additional cooling delays will need to be introduced into the experiment to avoid equipment damage.¹³

Recently, we introduced an SPI modification different from SPRITE, using a sinusoidal gradient ramp that efficiently attenuates acoustic noise and at the same time has a lower power deposition into the gradient set.¹⁴ In a second modification we introduced SPI/VTI, an SPI modification using a variable phase-encoding time,¹⁵ introducing a substantial gain in SNR. Combining SPI/VTI with the use of soft shaped gradients resulted in an efficient pulse sequence for high-resolution solid-state MRI. These methods are described in detail in the following.

Theory

Acoustic noise reduction

Gradient coils in MR scanners have been observed to show a linear response to applied gradient pulses.¹⁶ Each gradient coil in each MR scanner can thus be characterized by its corresponding frequency response function (FRF).^{14,16} The product of the FRF and the frequency spectrum of the gradient pulse determines the frequency output of the gradient set. The FRF of a gradient coil can be determined as the Fourier transform of its response to a frequency-limited sinc-shaped gradient pulse. This type of excitation can be achieved using a 10 ms long 40-lobed sinc pulse that has a flat frequency spectrum in the human audible, most sensitive frequency range of 0–4000 Hz. Figure 2 shows the typical FRFs for a transverse gradient coil (Fig. 2A) and the longitudinal coil (Fig. 2B).

The gradient pulse program for SPI is straightforward and can be described as a train of trapezoids with amplitude A , ramp time R , a plateau duration P , and a repetition time TR (see Fig. 1A). Assuming these phase encode gradients are an even periodical function, the gradient train $G(t)$ can be written as a Fourier series:¹⁷

$$[4] \quad G(t) = \frac{a_0}{2} + \sum_{n=1}^{\infty} a_n \cos\left(nt \frac{2\pi}{TR}\right)$$

$$[5] \quad a_n = \frac{2}{TR} \int_0^{TR} G(t) \cos\left(nt \frac{2\pi}{TR}\right) dt, \quad \text{with } n = 0, 1, 2, \dots$$

where a_n ($n = 0, 1, 2, \dots$) are the expansion coefficients of the Fourier series. Expressions for the coefficients a_n for the specific case of linear and sinusoidally ramped gradients were presented by Latta et al.¹⁴ Each coefficient a_n corresponds to a frequency $f_n = \frac{n}{TR}$. The spectral power (sound pressure level, SPL) for a specific gradient pulse can now be calculated for various combinations of $\frac{R}{TR}$ and $\frac{P}{TR}$. It has been shown that the power of the fundamental frequency, corresponding to $f_1 =$

Fig. 2. (A) Frequency response function (FRF) of a transverse gradient coil. (B) FRF of the longitudinal gradient coil. Both spectra were obtained on the 11.7 T wide bore vertical bore system. The FRF magnitudes were scaled to 1 a.u. The transverse gradients show a typical reduced response for the low-frequency region up to ~800 Hz, whereas this is absent for the longitudinal coil. However, the sound pressure level for the longitudinal coil is ~18 dB (in C mode), below that of the transverse coils.

$(TR)^{-1}$, is relatively insensitive to changes in $\frac{R}{TR}$ and $\frac{P}{TR}$. However, for the remaining frequencies a minimum value in the total power can be obtained.¹⁴ Moving the fundamental frequency to the low-frequency range of the FRF of the gradient set by fixing $TR > f_R^{-1}$, with f_R the cut-off frequency in the FRF, audible acoustic noise is significantly reduced. In addition, choosing sufficiently long P and R values adds to the total accomplished acoustic noise reduction.

SNR optimization

When imaging an object with MR, data is collected in k -space (the spatial frequency domain). After acquisition the data is Fourier-transformed to obtain a map of the spin-density distribution. The spatial frequency vector in the raw-data map is given by the product of the gradient vector ($\mathbf{G}$) and the encoding time t_p :

$$[6] \quad \mathbf{k} = \frac{\gamma_H \mathbf{G} t_p}{2\pi}$$

Here, $\mathbf{G}$, is kept constant during t_p and the position in k -space is thus determined by the product of the gradient vector and the encoding time. During the standard SPI experiment, k -space is traversed by changing the gradient vector while leaving the encoding time constant. The SPI/VTI method traverses k -space differently than an SPI experiment, introducing a variable encoding time¹⁵ (Fig. 3). To obtain the same spatial frequency mapping we introduced a scaling constant that ensures encoding with the same spatial frequencies, however, with additional enhancement in the spin-density mapping, resulting in an increase in SNR. In short, the re-scaled parameters are

$$[7] \quad t_p = c T_p$$

Fig. 3. (A) Single point imaging (SPI) / VTI pulse diagram, showing the variable encoding delay t_p . Close to the centre of k-space (t_p varies as follows: $T_D < t_p < T_p$ (where T_D is the minimal encoding time possible and T_p is the maximum phase encoding time). (B) The SPI experiment, with a constant t_p , displays a constant signal amplitude (solid line); the SPI/VTI experiment shows a larger signal amplitude for $T_D < t_p < T_p$, close to the centre of k-space, while reducing at the extremities of k-space to the value obtained in the SPI experiment (broken line).

$$[8] \quad G'_i = \frac{1}{c} G_i, \quad \text{with } i = x, y, z$$

where c is the scaling constant, T_p is the phase encode interval, G'_i is the rescaled gradient magnitude, and G_i is the original gradient magnitude.

This allows for the variation of t_p ; $T_D < t_p < T_p$, where T_D represents the minimal encoding time possible set by hardware limitations. The sensitivity, defined as SNR per unit time, of SPI and SPI/VTI can be compared as both sequences use the same bandwidth and thus have an identical noise level per measurement.⁷ Comparing the total signal amplitude

for the two sequences in one phase-encode cycle we obtained:¹⁵

$$[9] \quad \frac{S_{\text{SPI}}}{S_{\text{SPI/VTI}}} = \frac{1}{T_p} \left[(T_D + T_2^*) e^{\frac{(T_D - T_p)}{T_2^*}} - T_2^* \right]$$

A phase-encode interval $T_p = \frac{T_2^*}{2}$ provides the optimal sensitivity for SPI.⁷ However, for this ratio, the maximal gain in sensitivity using SPI/VTI would be 1.28 for $\frac{T_D}{T_p} = 0.2$. However, with increasing $\frac{T_p}{T_2^*}$ sensitivity enhancements by a factor of 3–4 can easily be achieved.¹⁵ Thus, using SPI/VTI becomes beneficial when experimental limitations prevent the selection of a sufficiently short T_p . The ratio $\frac{T_D}{T_p}$ can be used to control image resolution, which in turn can be affected by

Fig. 4. Top and bottom rows represent two different orientations of a human tooth. Data was collected with the silent 3D single point imaging (SPI) sequence (see text) and reconstructed according to Šrámek et al.¹⁸ (A) Surface reconstruction, (B) volume reconstruction, and (C) object reconstruction, with particular attention to the area with high water content (caries), indicated in the crown area.

Fig. 5. Maximum intensity projections¹⁷ of the same tooth shown in Fig. 4. The two orientations shown in the top and bottom row of Fig. 4 correspond to (A) and (B), respectively.

cause of the influence of T_2^* related line broadening as SPI/VTI uses a variable encoding delay.

Experimental

The new methods were tested on human teeth and rubber stoppers. All MRI experiments were performed on an Avance DRX Bruker console (Bruker, Karlsruhe, Germany) using a

72 mm self-shielded gradient system SGRAD 123/72/S (o.d./i.d.) (Magnex, UK) installed in a vertical bore, 11.7 T magnet (Magnex, UK). Acoustic noise was measured using a Bruel and Kjaer (Nærum, Denmark) 2238 Mediator sound level meter equipped with a prepolarized free-field condenser microphone type 4188. For all measurements the microphone was centred 20 cm deep into the magnet and connected to

Fig. 6. Images of a phantom using different pulse sequences. Each image represents a slice in the phantom acquired at exactly the same position, using the following parameters: (A) 3D single point imaging (SPI), encoding time (t_p) = 270 μ s, repetition time (TR) = 10 ms. (B) 3D-SPI/VTI, 30μ s $\leq t_p \leq 120 \mu$ s, TR = 10 ms. (C) 3D gradient echo, echo time (TE) = 4 ms, TR = 40 ms. (D) 3D spin-echo, TE = 15 ms, TR = 1000 ms. All data were acquired with a matrix size of $64 \times 64 \times 8$ and a field of view (FOV) of $11 \times 11 \times 25$ mm³. These images show that SPI/VTI is not sensitive for susceptibility artifacts despite the variable encoding time used in the phase encoding. The phantom consists of a cylinder of water containing an air bubble (visible at top in the images) and two smaller cylinders with different diameters and thicknesses placed at the bottom.

the sound meter by a double-shielded cable. In all acoustic measurements the mean equivalent SPL was recorded (as defined by IEC 1672) and the results were obtained in both C and A modes. The latter takes into account the physiological characteristics of human hearing.

Experiments were performed using a home-built probe containing a Helmholtz RF coil tuned to 500 MHz for ¹H imaging and an orthogonally positioned three-turn RF solenoid tuned to 202 MHz for ³¹P experiments. The sample chamber had a diameter of 12 mm (i.d.) and a length of 12 mm.

¹H images were obtained using a data matrix of $96 \times 96 \times 32$ with a field of view (FOV) of $3 \times 3 \times 4$ cm³. Unless mentioned otherwise, SPI experiments used a detection time of $t_p = 125 \mu$ s in combination with a 10 μ s excitation pulse and a repetition time of 15 ms. The gradients were sinusoi-

dally shaped with a ramp time of 1.0 ms for optimal acoustic noise reduction (Fig. 1B). The resulting total experiment time was 80 min.

Teeth were obtained with consent from University of Manitoba Faculty of Dentistry, Winnipeg, Manitoba, patients. After sterilization, samples were stored in distilled water while refrigerated. Prior to an MRI experiment a tooth was removed from the container, blotted dry with tissue paper, and acclimatized to room temperature. Teeth used in this study did not contain living dental pulp.

Results and discussion

Using the simple modification of gradient shape adjustment as shown in Fig. 1B, SPI experiments can be made effectively silent for the 3D measurement of teeth. Restricting the TR of the experiment to values set by the flat part in the

Fig. 7. Single point imaging (SPI) images with a resolution of $64 \times 64 \times 16$ points, encoding time (t_p) = $120 \mu\text{s}$ and a total acquisition time of 11 min; signal-to-noise ratio (SNR) = 8.25. (B) SPI/VTI with a resolution of $64 \times 64 \times 16$ points, $40 \mu\text{s} \leq t_p \leq 120 \mu\text{s}$ and a total acquisition time of 11 min; SNR = 32.6. All other parameters were kept identical using the same maximal gradient amplitude. (A) and (B) Central slice of the 3D experiment. (C) and (D) A zoomed detail of the object, indicated by the square in (A). The SPI/VTI (B) sequence exhibits an $\sim 4\times$ higher SNR compared with SPI (A).

FRF in Fig. 2A, containing frequencies up to ~ 800 Hz; $\text{TR} > 800^{-1}$ s, will result in significant modulation of the fundamental acoustic frequency. Excellent acoustic noise suppression was achieved for the 11.7 T gradient system when the $R = 175 \mu\text{s}$ linear gradient ramps were replaced by a 1 ms sinusoidal slope while keeping the duration of the gradient at $P = 0.4$ ms and $\text{TR} = 15$ ms. These parameters resulted in an approximate sound pressure level of ~ 68 dB, close to the background value (measured in A scale) in the laboratory.

SPI with shaped gradients also significantly reduced the power deposition into the gradient system compared with SPRITE.⁹ The ratio of power deposition for the two sequences can be calculated as¹⁴

$$[10] \quad \frac{P_{\text{ShapedSPI}}}{P_{\text{SPRITE}}} = \frac{0.625R + P}{\text{TR}}$$

where $P_{\text{ShapedSPI}}$ and P_{SPRITE} represent the power deposited by the shaped SPI and the SPRITE sequences, respectively. For a gradient ramp time $R = 1$ ms, a plateau time $P = 0.4$ ms, and a repetition time $\text{TR} = 15$ ms, a reduction by a factor of

15 can be obtained. Of course, SPRITE was designed for samples with smaller T_1 values, allowing for a shorter TR.

SPI was used to obtain 3D maps of the mobile proton distribution in teeth, shown in Figs. 4 and 5. Most of the observed signal will arise from water; however, a small negligible fraction of the signal arises from proteins. Intact teeth (no caries) show T_1 relaxation with a relaxation constant of ~ 500 ms.^{19–21} T_2 relaxation is bi-exponential, arising from water in dentin tubules and in enamel. Water in the latter compartment is more restricted in mobility and thus has a shorter relaxation time. In Fig. 4B, which shows a volume reconstruction of the SPI data, the root canals can clearly be observed. Roots are mainly composed of dentin and show a higher water content. Also visible in Figs. 4B and 4C are the areas with tooth decay. Caries makes enamel porous, allowing the absorption of water, hence providing a larger proton density on the SPI images. SPI can be used as an analytical tool to study both tooth surface as well as volume. Images can be used to provide detailed 3D information on roots that can be used to construct accurate inner root canal fixations for tooth restoration purposes. Figure 5 shows a gray scale

image of the tooth, reconstructing the proton spin density in a maximum intensity projection.

Figure 6 shows the effect of susceptibility changes at the air–water interface. Images of a water phantom filled with an air bubble were obtained with SPI, SPI/VTI, gradient echo, and spin echo sequences (see Fig. 6 for details). The reconstructed images reveal that despite severe inhomogeneity artifacts observed in the gradient and, to a lesser extent, spin echo acquisition, no artifacts were detected for the SPI/VTI acquisition when compared with the SPI image, which is known to be artifact free. The influence of the variable time encoding in the SPI/VTI sequence did not significantly affect image resolution.¹⁵ In vivo applications of SPI are being developed, especially with an increasing number of people having bones replaced (hip, knee, etc.).²² These applications would benefit from the use of SPI/VTI and (or) silent SPI.

The use of the SPI/VTI sequence was demonstrated by imaging a small object made of soft rubber with $T_2^* = 73 \pm 4 \mu\text{s}$. Figure 7 shows the results comparing the central slice of 3D SPI (Figs. 7A and 7C) experiment with that of the SPT/VTI (Figs. 7B and 7D) experiment. Both experiments were performed with the same data matrix ($64 \times 64 \times 16$), gradient strengths, 1 average, FOV = $20 \times 20 \times 20 \text{ mm}^3$ and a 32° RF pulse. The SPI/VTI experiment, with $40 \mu\text{s} \leq 120 \mu\text{s}$, resulted in an SNR of 32.6, whereas the SPI experiment, with $t_p = 120 \mu\text{s}$, only resulted in an SNR of 8.25. The finer details of the images (see Figs. 7C and 7D)) show that the SPI/VTI acquisition, despite the partial filtration effect owing to the variable encoding time, is capable of reproducing the geometric details of the object while providing substantial SNR enhancement compared with SPI.

Conclusions

In summary, two modifications of the SPI sequence were discussed and tested experimentally. The first modification was based on a general reduction of the SPL of SPI, associated with the fast switching of encoding gradients. Shaped gradients were shown to significantly reduce acoustic noise while offering an attractive alternative to the SPRITE sequence, reducing the overall power deposition into the gradient set of the magnet. The second modification was based on a reduction of signal attenuation during the phase-encoding period (SPI/VTI sequence), using a variable reduction of the encoding time. Used on materials with short T_2^* values, the SPI/VTI sequence showed a significant increase in SNR with minimal distortions owing to susceptibility effects.

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