Enhancement Gradient Pulse Waveforms in MR Tomography

E. Gescheidtova and R. Kubasek
Department of Theoretical and Experimental Engineering
Faculty of Electrical Engineering and Communication, Brno University of Technology
Kolejni 2906/4, 612 00 Brno, Czech Republic

Abstract— The magnetic resonance (MR) imaging techniques of tomography and spectroscopy are exploited in many applications. For the MR instruments to function properly it is necessary to maintain a high quality of homogeneity of the fundamental magnetic field. The pre-emphasis compensation of the generated gradient field increases the homogeneity of the generated magnetic field and reduces the minimum switching times of the gradients. This enables measuring the MR images of incisions in the human body, the relaxation properties of nuclei, self-diffusion processes, flows of liquids and movements of solids faster and more accurately.

1. INTRODUCTION

When defining the area being measured in localized spectroscopy and tomography, the gradient field is excited by very short pulses of sufficient magnitude. This gives rise to a fast changing magnetic field, which induces eddy currents in the conducting material near the gradient coils. These currents then cause retrospectively unfavourable deformation of the total magnetic field [1]. The basic idea of a method that compensates this effect consists in replacing the missing leading edge of the field waveform by an overshoot of excitation current as shown in Fig. 1. To have the best possible compensation it is necessary to find an optimum shape of excitation pulse. Basically, this consists in obtaining the spectrometer response pulse, inverting it and using this inversion to filter the excitation (usually square) pulse. The term pre-emphasis compensation method is based on the fact that the compensation filter is in the nature of a differentiating element (high-pass filter).

At the present time, pre-emphasis filters are implemented by digital means, most frequently digital signal processors. At the Institute of Scientific Instruments, Academy of Sciences of the Czech Republic in Brno, pre-emphasis filters are implemented on a Motorola 96002 DSP as a fifth-order IIR filter in the first canonical form.

The principle of measuring the waveform of gradient pulse consists in determining the changes in instantaneous frequency [2] of an MR signal produced by the resonance of nuclei excited in two thin layers positioned symmetrically about the gradient field centre. The instantaneous frequency of MR signal is directly proportional to the induction of magnetic field $B(\pm\alpha_n, t)$. The average inductions of magnetic field $B(\alpha_n, t)$ are measured in the excited layer in the $+\alpha_n$ and $B(-\alpha_n, t)$ positions in the $-\alpha_n$ layer; α is one of the (x, y, z) directions.

From the differences of the two inductions measured the magnitude of gradient can be calculated according to the relation

$$G_\alpha(t) = \frac{1}{2\alpha_n}\left[B(\alpha_n, t) - B(-\alpha_n, t)\right]. \tag{1}$$

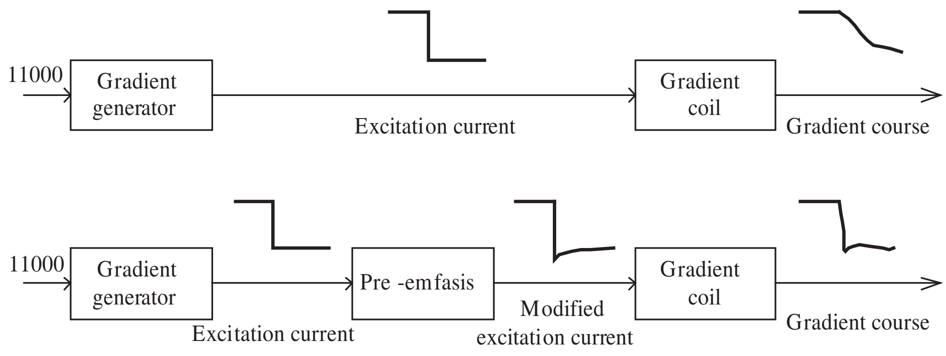

Figure 1: Principle of pre-emphasis compensation.

The measured waveform of magnetic inductions $B(\alpha_n, t)$ and $B(-\alpha_n, t)$ is much distorted due to noise. The gradient $G_r(t)$ calculated according to Equation (1) is even more distorted, due to the incorrelatibility of noise. To obtain an accurate and undistorted waveform of the gradient $G_\alpha(t)$, we must remove the noise.

2. DENOISING

The principle of sub-band denoising is shown in Fig. 2. Using an analysing filter bank the input signal is divided into a series of sub-band filters. Partial sub-band signals are thresholded appropriately in order to optimally suppress noise without influence on the useful signal. The magnitudes of thresholds p_i are calculated by within-block threshold estimation. In the block of synthesis filter bank the sub-band signals are again synthesized into the resultant signal.

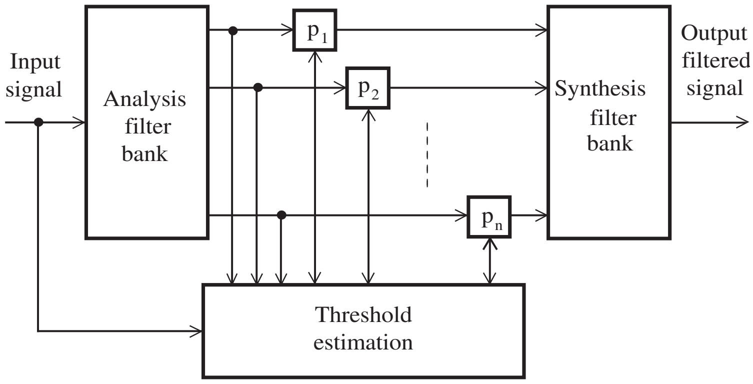

Figure 2: Block diagram of sub-band denoising.

The choice of the parameters of individual parts of blocks depends in the first place on the input signal properties. The filter bank base is chosen in dependence on the distribution of input signal spectral density. The length of pulse characteristics and the type of filter bank are chosen with a view to the required attenuation in the stop band, computation complexity, and the potential occurrence of transition phenomena. To suppress noise, filter banks without downsampling are usually used. Using a filter bank with downsampling will reduce the number of operations and the application demands on memory but at the cost of worse results. The threshold magnitudes calculated in block threshold estimation are usually calculated from the standard deviations of noise δ. Perhaps the best known relation derived for white additive noise is that with Gaussian distribution for global threshold $p = \delta\sqrt{2\ln(L)}$, where L is the input signal length. This value is usually very high, it is necessary to choose the threshold magnitude to be $p = \delta \cdot K$, where K is the empirically obtained constant. There are many types of thresholding [3], the most widely used thresholding is the soft or the hard type. By analysing the operation of calculating instantaneous frequency from the FID signal it is possible to obtain the noise parameters of instantaneous frequency signal necessary for automatic threshold adjustment [4]. The instantaneous phase of FID signal will be calculated using

$$\varphi = \arctan\left(\frac{\mathbf{Im}\,[\mathrm{FID}]}{\mathbf{Re}\,[\mathrm{FID}]}\right). \tag{2}$$

It is exactly $\Delta\varphi$ that represents the noise contained in the instantaneous frequency signal of FID signal. The magnitude of the change Δ in the real and the imaginary parts of the signal is directly linked with the magnitude of standard noise deviation, $\delta \approx \Delta$. Since the standard noise deviation of FID signal is constant, $\Delta\varphi$ changes in dependence on the magnitude of the FID signal and is thus non-linearly dependent also on SNR. Fig. 3 gives the waveforms of magnetic induction $B(\alpha_{n,t})$ before and subsequent to filtering. As can be seen, there has been considerable noise attenuation. The gradient $G_\alpha(t)$ is free from noise and, at the same time, measured with sufficient accuracy to be used in determining the coefficients of the pre-emphasis filter.

3. PREEMFASIS COMPENSATION

The lay-out of pre-emphasis filters is as shown in Fig. 4. Pre-emphasis filters are basically inverse filters to the model of tomographic scanner. The task is to generate a waveform of the gradient

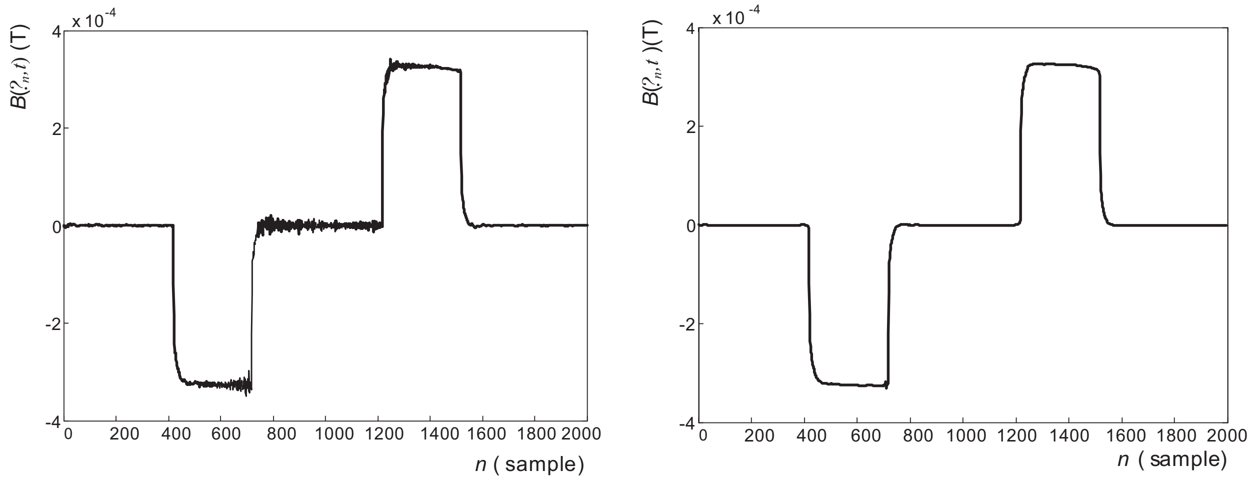

Figure 3: Waveforms of magnetic inductance $B(\alpha_n, t)$, (a) before filtering, (b) after denoising filtering.

$G'_\alpha(t)$ pre-distorted to such a degree that subsequent to the action of eddy currents the generated gradient $g_\alpha(t)$ is of the required waveform, i.e., $G_\alpha(t)$. The following equation must hold

$$g_\alpha(t) = G_\alpha(t), \tag{3}$$

and there must not be any change in the basic magnetic field $B_0(t)$. For a direct pre-emphasis filter $P_\alpha(z)$ it holds

$$G_\alpha(z)P_\alpha(z)M_\alpha(z) = g_\alpha(z). \tag{4}$$

$$P_\alpha(z) = M_\alpha^{-1}(z) = \frac{G_\alpha(z)}{g_\alpha(z)}. \tag{5}$$

The effect of eddy currents will be compensated under the conditions (5). The measurement itself must be performed for the setting $P_\alpha(z) = 1$, preferably for $P_{\tilde{\alpha}}(z) = 0$. The transfer of the coil $M(z)$, which compensates the basic magnetic field B_0 will be measured for quite the opposite setting, namely $P_\alpha(z) = 0$ and $P_{\tilde{\alpha}}(z) = 1$. The basic field B_0 is excited directly by the gradient in the respective direction, and for $M_0(z)$ it holds

$$G_\alpha(z)P_{\alpha 0}(z)M_0(z) = \Delta B_0(z). \tag{6}$$

The cross transfer of the gradient in the given direction $G_\alpha(z)$ to the basic field B_0 via $M_{\alpha 0}(z)$ is given

$$\Delta B_0(z) = G_\alpha(z)P_\alpha(z)M_{\alpha 0}(z). \tag{7}$$

The compensation of the basic field using the cross pre-emphasis filter $P_{\alpha 0}(z)$ will take place if a basic field of opposite polarity is generated,

$$G_\alpha(z)P_{\alpha 0}(z)M_0(z) = -\Delta B_0(z). \tag{8}$$

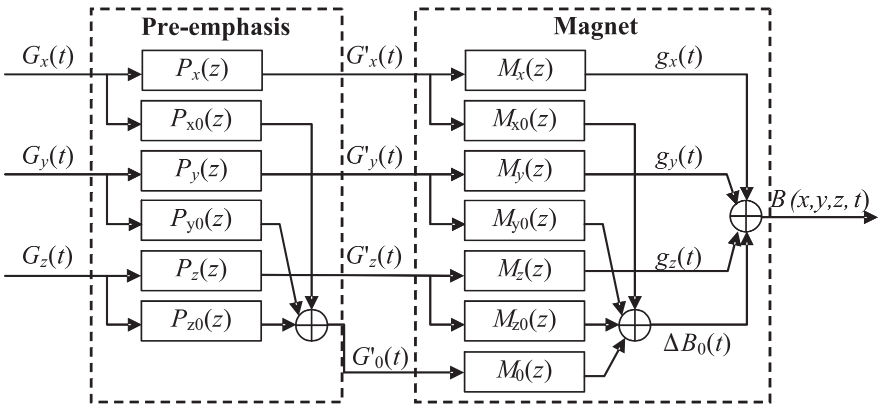

Figure 4: Pre-emphasis compensation and model of tomograph.

Thus it must hold

$$P_{\alpha 0}(z) = -\frac{\Delta B_0(z)}{G_\alpha(z) M_0(z)}. \tag{9}$$

Equation (9) says that the cross transfer of the gradient in the given direction $G_\alpha(z)$ to the basic field B_0 via $M_{\alpha 0}(z)$ must be measured for an already set direct pre-emphasis filter $P_\alpha(z)$. Its calculation must include an earlier determined transfer of the compensation coil $M_0(z)$. The sequence in which measurements are performed cannot be changed. The calculation of pre-emphasis filters by relations (4), (6) and (9) is always determined from two signals, one input signal and one output signal. The ratio of their Z-images gives the desired transfer characteristic of the pre-emphasis filter. The directly found characteristic is basically always unstable; none of its poles are inside the unit circle. We must find a stable IIR filter, which by its properties is the closest to the ideal inversion. There are several ways how to find this IIR filter. The best results have been obtained using the Matlab function *stmcb*, which employs the Steiglitz-McBride iteration process [5] used to calculate the approximation error by the LMS method.

4. EXPERIMENTAL RESULTS

Figure 5 gives the waveforms of positive gradient impulse G_x, G_y, and $G_z$with and without pre-emphasis compensation. The most essential change consists in the shortening of the leading and trailing edges of the gradient impulse $-G_x$ rise time from 250 µs to 70 µs. The leading edge has been shortened ca. three times. The lengths of the edges of the gradient impulses G_y and G_z were reduced from 350 µs to 60 µs.

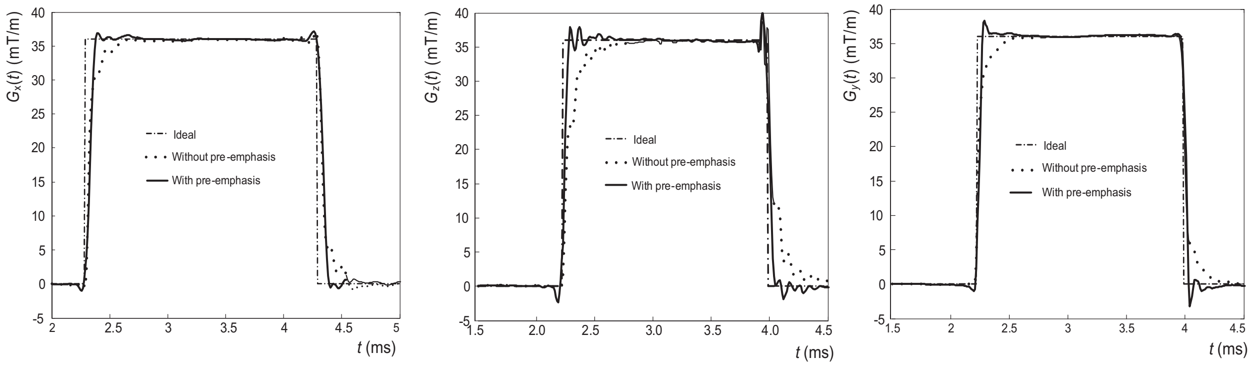

Figure 5: Comparison of positive gradient impulse G_x, G_y, and G_z with and without pre-emphasis compensation.

5. CONCLUSIONS

The application of pre-emphasis compensation of gradient magnetic field has led to a qualitative improvement in the parameters of the magnetic field generated in an MR tomograph. It can therefore be expected that better results will be obtained in all regions of MR tomography and spectroscopy where the generation of gradient fields of a defined waveform with minimum switching times is required. Today the minimum applicable switching time of gradients in MR tomography is a limiting factor in the application of fast imaging sequences in MRI. Shortening the gradient edges and improving the magnetic field homogeneity after a gradient impulse result in shortening the minimum applicable switching time of magnetic field gradients and the waiting time between gradient impulses, when the magnetic field is required to drop to the level of the homogeneity of the basic magnetic field B_0. In that case the images of incisions in the human body can be measured faster and more accurately.

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