Correcting K-trajectory by Using Multiple Function Models of Gradient Waveform for Ultrashort TE(UTE)

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Introduction:

The UTE sequence, based on radial sampling, acquires echo signal from central to outer parts of k-space [1]. For this kind of sequence, K-trajectory error caused by gradient system response becomes a major problem; exact gradient output is necessary for obtaining good images. For clinical use of UTE it is important to be able to change imaging conditions and parameters for a general oblique imaging plane without causing K-trajectory errors. Even if there are errors on the gradient output, the K-trajectory can be estimated by calculating the error strictly [2]. Thus, the total error of the gradient system (gradient timing error, eddy current, gradient system response) can be corrected.

Subject and Method:

In this study, we assume multiple function models of the gradient waveform. Figure 1 shows an example of four function models that divide the gradient waveform into four periods. This model consists of exponential functions with a different coefficient for each period. In this case, the beginning of ramp-up part (A) and the end of ramp-up part (C) are defined as exponential functions and other parts (B, D) are defined as linear functions, respectively. For image reconstruction, the resulting gradient waveforms calculated from multiple function models are used for gridding. The optimal coefficients for multiple function models are calculated for each physical axis (X/Y/Z) separately. Then the resulting gradient waveforms are combined to fit the K-space rotation angle for radial sampling for a general oblique imaging plane by using a rotation matrix. A 1.5T MRI (Echelon, Hitachi Medical Corporation, Tokyo, Japan, with 33 mT/m of maximum amplitude and 150 T/m/s of maximum slew rate) and dedicated knee coil were used for evaluation of a phantom and for volunteer imaging. Optimal coefficients for each function were adjusted by using a phantom before the examination. For volunteer evaluation, an oblique imaging plane was selected. For the evaluation of the ability to change imaging parameters of the UTE sequence, the optimal coefficients were modified according to gradient amplitudes which were used for imaging.

Results and Discussion:

Figure 2 compares gradient waveforms calculated by original and by multiple function methods. In this case, the time constant of the function near the beginning of ramp-up part was different from that near the end of the ramp-up part. Figure 3 shows the resulting images of a phantom acquired by UTE sequence. Acquisitions were made with two different gradient amplitudes. For both experiments, the images were distorted and non-uniform in the case of no correction (a,c). By applying multiple function models, image quality was improved in both cases (b,d). Figure 4 shows the resulting images of volunteer knee menisci regions acquired by the UTE sequence. In this case, the optimal coefficients for multiple function models were calculated before the examination by using a phantom and were then used for all reconstructions. Images made without our method have fatal signal loss due to loss of the center part of K-space caused by K-Trajectory error. Images made using our method showed drastically improved image quality, even for oblique imaging planes. As a result, knee menisci were shown with high signal intensity in the subtracted image (4e).

For most systems K-trajectory error caused by gradient imperfections are corrected by a single transfer function method that uses an equivalent circuit model [3] and an accurate gradient waveform measurement method [4]. Many gradient systems use some feedback process on gradient output in order to bring gradient waveforms as near as possible to a perfect trapezoidal shape. From our experience, the single transfer function method has not worked well for this kind of gradient system. The gradient waveform measurement method can correct the K-trajectory error if enough signals for calculation are acquired. This method uses the phase information from the acquired signals to calculate the gradient waveform. The accuracy of the phase information is sensitive to imaging conditions such as subject size, position, tissues and so on. Improving the stability of the calculation of the gradient waveform requires high signal to noise ratio of the signals, resulting in extended scan time for clinical use. In our method, we can eliminate the error in K-trajectory caused by the gradient system response effectively without increasing scan time for clinical use. Optimal coefficients can be calculated by using a phantom at the installation phase of the system, thus the stability for changing the imaging condition will not be a problem.

Conclusion:

Our multiple function models that divide gradient waveform into several periods and use different coefficients for fitting gradient waveform in each period can eliminate the error on K-trajectory caused by the gradient system response effectively without increasing scan time.

References:

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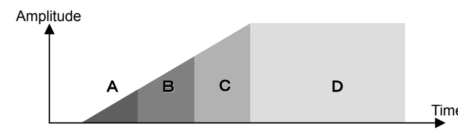

Figure 1 Four regions used for multiple function models on gradient waveform. In each region, the different coefficients can be used for calculation.

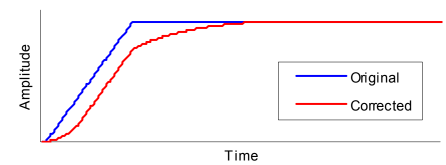

Figure 2 Gradient waveforms calculated from original (blue) and optimal (red). Note that the time constants of the function between the beginning of ramp-up part and the end of ramp-up part were different in the optimal case.

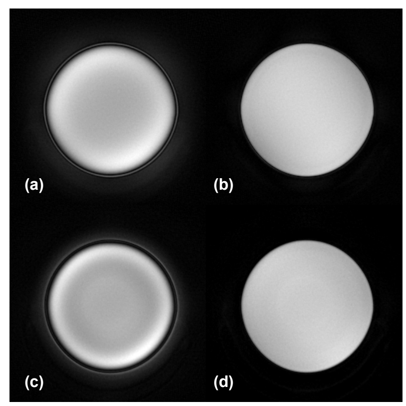

Figure 3 Phantom images acquired by UTE sequence without (a,c) and with (b,d) multiple function models. Gradient amplitudes are 4.7 mT/m (a,b) and 9.4 mT/m (c,d).

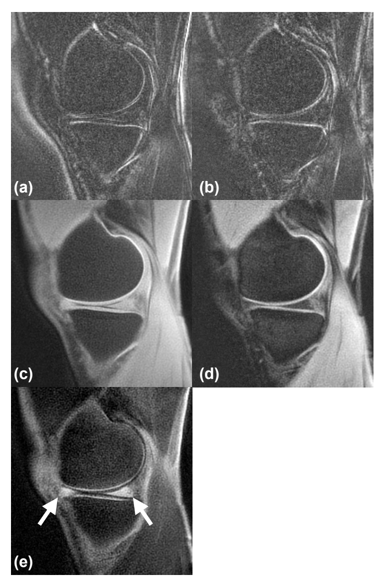

Figure 4 Volunteer knee menisci images acquired by UTE sequence without (a,b) and with (c,d,e) multiple function models. Echo times are 0.36 ms (a,c) and 10.0 ms (b,d). Knee menisci shows high signal in subtracted image (e) (arrow).