Appendix E1

Sequence Design and Testing

Wideband LGE Pulse Sequence Design and Simulation

To address the inversion pulse spectral bandwidth issue of the current LGE sequence, we sought to design and implement a wideband adiabatic inversion pulse to ensure proper inversion of the myocardial signal affected by the device generator.

Adiabatic radiofrequency inversion pulses are preferred for inversion recovery sequences such as LGE because adiabatic pulses are insensitive to B_1 inhomogeneity and produce excitation uniformly. The most commonly used adiabatic inversion pulse is the hyperbolic secant pulse, which consists of amplitude and phase modulation functions:

$$A(t) = A_0 \operatorname{sech} \beta t \tag{E1}$$

and

$$\phi(t) = \mu \ln(\operatorname{sech} \beta t) + \mu \ln A_0 \, , \tag{E2}$$

where A_0 is the peak B_1 field, β is a frequency modulation parameter (in units of rad per second), t is time, and μ is a dimensionless phase parameter. For the adiabatic passage condition to be fulfilled, the peak B_1 amplitude, A_0, must exceed the threshold given by the following expression:

$$A_0 >> \frac{\sqrt{\mu\beta}}{Y} \, . \tag{E3}$$

The center of the modulation frequency can be offset by adding a time-frequency-offset product, tf_{offset}, to the phase modulation function.

The spectral bandwidth of a hyperbolic secant pulse is given by the product of the frequency modulation parameter and the phase modulation parameter: μβ. Thus, by altering either or both of these parameters, the bandwidth can be modified.

The conventional LGE sequence on our 1.5-T MR imaging unit uses a hyperbolic secant inversion pulse with β = 672 rad/sec and μ = 5, which yields a bandwidth of 1.1 kHz. From the literature and our prior experience, we have seen that this bandwidth is insufficient for LGE imaging in the presence of an implanted cardiac device. The metal casing of the ICD induces off-resonance in the myocardium greater than 1 kHz. As a result, the off-resonant myocardial spins are not fully inverted when the inversion pulse is applied, and these spins give rise to the hyperintensity artifacts that are typically seen. To overcome this artifact, we designed new hyperbolic secant inversion pulses with wider bandwidths. During the initial phase of our study, we implemented a pulse with β = 750 rad/sec and μ = 10, which yielded a bandwidth of 2.4 kHz. Subsequently, the bandwidth was increased even further to 3.8 kHz by using β = 750 rad/sec and μ = 16. The potential issue with wideband adiabatic inversion pulses is that they require a higher B_1 field to satisfy the adiabatic passage condition in Equation (E3). We chose to make only a small increase to β and a larger increase to μ, in comparison to the conventional inversion pulse, because the minimum B_1 amplitude for adiabatic inversion scales linearly with β but only as the square root of μ (as in Eq [E3]). The minimum B_1 amplitude required to fulfill the adiabatic condition is 8.9 μT for the 2.4 kHz pulse and 11.2 μT for the 3.8 kHz pulse. The new inversion pulses were implemented in the existing two-dimensional inversion recovery LGE sequence, replacing the standard 1.1-kHz inversion pulse. The new inversion pulses were tested with a Bloch equation simulator in Matlab (Mathworks, Natick, Mass). Inversion performance and bandwidth were tested by generating plots of the longitudinal magnetization against a range of off-resonance frequencies.

All MR imaging was performed with 1.5-T MR imaging units, which were equipped with 40 mT/m gradients with maximum slew rates of 200 T/m/sec. Phantom experiments were performed by using an eight-channel spine coil, and volunteer and patient imaging was performed by using a six-channel body coil. In the phantom and healthy volunteer experiments, an ICD (Atlas II, model #V-268; St Jude's Medical) was used to simulate the off-resonance effects of these devices in patients.

Phantom Experiments

The modified wideband LGE sequences were tested in the American College of Radiology (ACR) standard phantom. The optimal inversion time for the solution in the ACR phantom was determined by using a Look-Locker sequence (14,15) in the absence of the ICD. The ICD was positioned at a distance of 6 cm from the phantom. LGE images were acquired by using the standard LGE sequence (using the 1.1-kHz inversion pulse) and the modified LGE sequences (using both 2.4- and 3.8-kHz bandwidth inversion pulses).

To study the relationship between ICD-caused off-resonance and the distance from the ICD, the ICD was placed 5 cm away from the edge of the phantom. The B_0 field map was measured by using two single-echo gradient-echo sequences with 2.5- and 3.5-msec echo times. The field map was calculated as the phase difference between the two images divided by the difference in echo time. Phase was unwrapped in the phase difference image by applying the Matlab function "unwrap" along columns on the phantom pixels; phase at the top of the phantom in the phase difference image was unaffected by the ICD because the ICD was at the bottom edge of the phantom (Fig E3a).

Results

Wideband LGE Pulse Sequence Design and Simulations

The properties of the inversion pulses, including amplitude, frequency, and phase modulation functions, as well as the longitudinal magnetization profile produced, are shown in Figure E2. The standard hyperbolic secant inversion pulse used in the conventional LGE sequence is shown in Figure E2a. This pulse used $\beta = 672$ rad/sec and $\mu = 5$. This yielded a bandwidth of 1.1 kHz and required a minimum B_1 amplitude of 5.6 μT. This bandwidth is adequate for routine imaging in the absence of cardiac devices but is not adequate for large off-resonances, as those introduced by the presence of a cardiac device. Figure E2b shows the modulation functions of the 3.8-kHz inversion pulse with a center frequency shift of 1.5 kHz. In all of our in vivo studies, this center frequency was shifted by 0 to ±1500 Hz to ensure that both the on-resonance and off-resonance tissue stayed within the spectral bandwidth of the new wideband inversion pulse.

A pulse sequence diagram for the LGE sequence, with the new 3.8-kHz bandwidth inversion pulse implemented, is shown in Figure E2c. The phase modulation function shown in the sequence diagram is wrapped about $\pm 2\pi$ and shows rapid changes. LGE sequences used clinically are segmented gradient-echo sequences, with 15-25 lines per k-space segment.

Field Map and Phantom Results

Figure E3a shows the results of the field map measurements. The left image shows the field map generated across an ACR phantom when an ICD is placed 5 cm from the phantom. The graph on the right shows the field profile (along the dotted line on the field map). The off-resonance due to the ICD is about 6 kHz at a distance of 5 cm, which drops to about 2 kHz at 10 cm. Typically, an ICD is implanted at a distance approximately 5–10 cm from the heart. Thus, the field map data suggest that the off-resonance values experienced at the myocardium because of an ICD lie roughly in the 2–6 kHz range.

The conventional LGE sequence uses an inversion pulse with a 1.1-kHz bandwidth. This is sufficient to invert spins that are off-resonance up to ±535 Hz. According to the graph in Figure E3a, this suggests that if the distance of an ICD from the myocardium is less than approximately 16 cm, hyperintensity artifacts will be produced when the conventional LGE sequence is used. Thus, if a patient has an ICD implanted at the right shoulder, the conventional LGE sequence would be adequate.

Figures E3b and E3c illustrate the effect of the conventional and new inversion pulses on an ACR phantom with an ICD placed 6 cm away. Figure E3b shows a gradient-echo image (no inversion pulse) of the phantom setup. In Figure E3c (top left), the phantom was imaged with the conventional LGE sequence by using the 1.1-kHz inversion pulse applied at an inversion time of 110 msec. In the absence of an ICD, this inversion pulse is sufficient to nullify the entire phantom image. However, off-resonance due to the ICD prevents inversion of the spins near the ICD, resulting in a hyperintensity artifact that spans more than one-third of the phantom. Figure E3c (top right and bottom left) also shows the LGE images obtained in the phantom with inversion pulses of 2.4 and 3.8 kHz bandwidths, which are sufficient to invert off-resonances up to 1.2 and 1.7 kHz, respectively. The increasing spectral coverage of the inversion pulse causes the hyperintensity artifact to become progressively smaller. Figure E3c (bottom right) shows an image with a 3.8-kHz bandwidth inversion pulse in which the center of the modulation frequency was shifted by 1500 Hz. This frequency offset increases the spectral coverage up to 3.2 kHz off-resonance, which is why the hyperintensity artifact is even smaller than in the bottom left image.

Geometric distortions of the ACR phantom grid due to ICD off-resonance can be seen in Figures E3b and E3c. Grid squares close to the ICD were distorted from left to right, which is the frequency encoding direction for these images. Dimensions of the grid squares were 1.5×1.5 cm, and the grid was 2 cm from the circumference. Thus, the distortions become noticeable at a distance of about 12–13 cm away from the ICD. In these five images, the receiver bandwidth was 500 Hz/pixel, which is the value used in conventional LGE.