Cine DENSE MRI with Dual Displacement Encoding

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Introduction Cine DENSE (Displacement Encoding with Stimulated Echoes) MRI acquires a temporal sequence of images where the phase of each image is encoded for tissue displacement. More specifically, the phase of the DENSE stimulated-echo signal is given by $\theta = k_e\Delta x$, where Δx is the tissue displacement and k_e is the user-specified displacement-encoding frequency (a parameter analagous to the "venc" in velocity-encoded phase contrast MRI). The selected value for k_e determines the degree of the phase shift for a given displacement (and, consequently, whether or not phase wrapping occurs), and the k_e value also affects the degree of intravoxel dephasing and associated signal loss in tissues that deform such as cardiac and skeletal muscle (1). Thus, an optimal value for k_e depends on the magnitude of the tissue displacement and deformation, where larger k_e values provide greater sensitivity for smaller displacements and deformations and smaller k_e values provide less phase wrapping and intravoxel dephasing for larger displacements and deformations. Since cine DENSE typically acquires a temporal sequence of images, often with smaller tissue displacements early in the sequence and larger displacements later, no single value of k_e is optimal for all times. The purpose of the present study was to develop a cine DENSE sequence with dual displacement encoding, where two distinct stimulated echoes with different displacement-encoding frequencies are simultaneously stored along the longitudinal axis, and where either can be recalled at any particular time to provide a better match between k_e and Δx.

Methods. As shown in Fig. 1, a dual-encoded sequence was developed using 3 hard nonselective RF pulses separated by 2 gradient pulses (rather than the usual 2 RF pulses separated by a single gradient pulse) for the displacement-encoding module. Following these pulses, the longitudinal magnetization is modulated as

$$M(x) = M(A\cos(k_{e1}x) + B\cos(k_{e1}x + k_{e2}x) + C\cos(k_{e1}x - k_{e2}x)) \qquad [1]$$

By using $k_{e2} = 2k_{e1}$, the longitudinal magnetization is given by

$$M(x) = M(A + C)\cos(k_{e1}x) + MB\cos(3k_{e1}x) \qquad [2]$$

where A, B and C are coefficients related to flip angles ϕ_1 and ϕ_2 of the second and third RF pulses in displacement-encoding module. With this modulation of longitudinal magnetization, stimulated echoes with phases of $\theta = k_{e1}\Delta x$ (low sensitivity) or $\theta = 3k_{e1}\Delta x$ (high sensitivity) can be separately recalled by using either a smaller or larger displacement unencoding gradient in the readout module (Fig. 1). Extraneous signals other than the desired stimulated echoes are suppressed using phase cycling and though-plane dephasing (2). The dual-encoded cine DENSE sequence was implemented on a 1.5T Avanto MRI system (Siemens, Erlangen, Germany), and was evaluated using a deformable phantom (3) and, with informed consent and under IRB-approved protocols, by imaging the hearts of normal volunteers.

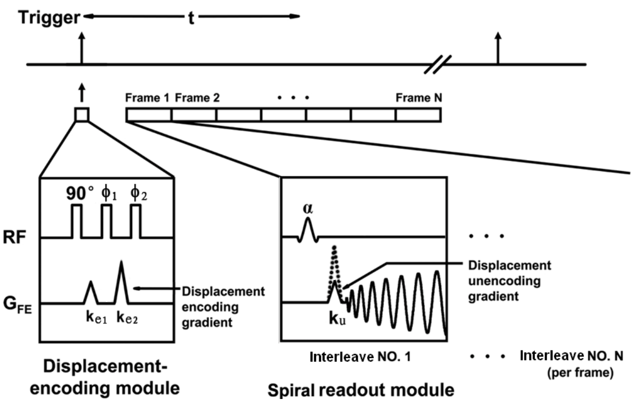

Fig. 1. Pulse sequence timing diagram for cine DENSE with dual displacement encoding.

Results In Fig. 2, *k*-space data are shown to illustrate the different signals generated by this pulse sequence. In total, 5 primary echoes are generated, where their location in the k_x direction depends on the area of the unencoding gradient and where 2 of the echoes can be considered stimulated echoes whose phases are given by $\theta = k_{e1}\Delta x$ (red arrows) and $\theta = 3k_{e1}\Delta x$ (yellow arrows). Either of the stimulated echoes can be made to refocus in the center of *k*-space by application of the appropriate unencoding gradient pulse during the readout module, and all of the other, unwanted, echoes can be suppressed by phase cycling and application of gradient pulses applied in the through-plane direction (2). Selected frames from a temporal sequence of dual-encoded cine DENSE images of a deformable motion phantom are shown in Fig. 3. For an early frame with small displacements (A,B), the stimulated echo with a high encoding frequency was acquired, and, for a later frame with larger displacements (C,D), the stimulated echo with the low encoding frequency was acquired. These images illustrate the larger phase shift of the early time frame due to the use of a higher displacement encoding frequency, and the smaller phase shift from the later time frame due to the smaller encoding frequency. MRI of volunteers demonstrated the same properties for *in vivo* heart imaging.

Conclusions To better accommodate a temporal sequence of images with different amounts of displacement and deformation at different times, a dual-encoded cine DENSE sequence was developed. Two stimulated echoes with different displacement-encoding frequencies are simultaneously stored along the longitudinal axis, and either may be individually recalled at any given time using the proper unencoding gradient applied prior to signal readout. Extraneous signals are eliminated using phase cycling and though-plane dephasing gradients. Higher displacement-encoding frequencies can be used to sample smaller tissue displacements (for higher sensitivity) and lower displacement-encoding frequencies can be used to sample larger tissue displacements and deformations (to minimize phase wrapping and intravoxel dephasing). The theory of the technique was developed and the experimental feasibility of the technique was demonstrated using a deformable phantom. The technique was also demonstrated for cardiac imaging in healthy volunteers. A disadvantage of the method is decreased signal-to-noise ratio, as the relatively low signal of the stimulated echo is further divided into 2 stimulated echoes. For this reason, the method may be most useful when used for applications with high SNR.

References (1) Spottiswoode et al. IEEE Trans Med Imag. 2007;26(1):15-30. (2) Zhong et al. MRM 2006;56:1126-31. (3) Young et al. Radiology 1993;188:101-108.

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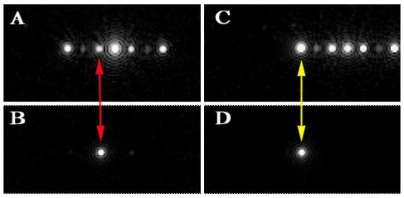

Fig. 2. *k*-space data illustrate the echoes observed using dual-encoded DENSE applied without artifact suppression (A,C). The echoes may be shifted in *k*-space by changing the area of the unencoding gradient (A vs C). The 2 echoes that maintain the properties of displacement-encoded stimulated echoes with encoding frequencies of k_{e1} (red arrow) and $3k_{e1}$ (yellow arrow) are highlighted. Either of these can be centered in *k*-space, and the other echoes can be suppressed using phase cycling and through-plane dephasing (B,D), providing isolated stimulated echoes suitable for image reconstruction.

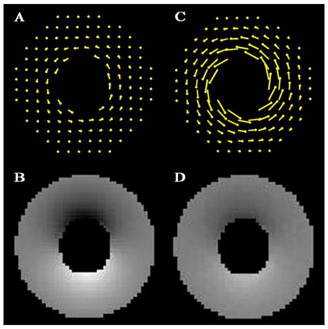

Fig. 3. Using a high encoding frequency, small displacements (A) yield larger phase shifts (B). Later in the sequence of images, when displacements are larger (C), use of a lower encoding frequency results in smaller phase shifts (D). Phase wrapping is avoided.