Solving the Problem of Misregistration of Arterial Input Function Pixels along the Phase-Encoding Direction by the Gradient Recalled-Echo Sequence

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INTRODUCTION

Dynamic contrast-enhanced (DCE) MRI is a powerful tool for the evaluation of tissue perfusion, blood flow and vascular permeability. However, for accurate quantification the simultaneous acquisition of an arterial input function (AIF) along with target tissues is necessary. Furthermore extracting the pixels to estimate the AIF is problematic. It has been reported that AIF pixels shift along the phase-encoding direction during the dynamic studies using an echo-planar imaging pulse sequence (1,2). This was explained as a magnetic susceptibility effect caused by high gadolinium concentration and low bandwidth per pixel in the phase-encoding direction. In a gradient recalled-echo sequence, Rausch et al. (1) suggested that the frequency shift didn't affect either readout or phase-encoding directions because of the wide bandwidth per pixel compared with the frequency shift. Conversely, Conturo et al. (3) suggested that a linear gadolinium concentration change would be expected to cause small-scale misregistration along the phase-encoding direction. In agreement with Conturo et al (3), we report a phenomenon of shifting pixels in the rat abdominal aorta on DCE-MRI obtained by a saturation recovery spoiled gradient recalled-echo T1-weighted imaging (SPGR T1WI) pulse sequence with the semikeyhole acquisition scheme.

METHODS

MR images were acquired using a 4.7 T horizontal MRI system (UNITYINOVA, Varian) with a 6-cm quadrature birdcage coil (Varian). Parameters for the saturation recovery SPGR T1WI sequence were: TR/TE = 13.9/2.3 ms, flip angle of the excitation pulse = 15°. Saturation was performed using a 90° RF pulse followed by a crusher gradient pulse. The delay for saturation recovery between the saturation pulse and the excitation pulse was 10 ms. Slice thickness = 5 mm. Field-of-view = 6 x 6 $(cm)^2$, matrix = 128 x 96. Each k-space constructed was zero-filled to 128 x 128. 2 averages. Bandwidth was 100 kHz.

Animal studies were performed in full compliance with the UK Animals (Scientific Procedures) Act, 1986. Athymic rats (240 g) were anesthetised with isoflurane/air. Before the DCE-MRI acquisition, SPGR T1WI was obtained without saturation recovery. DCE-MRI was acquired using the semikeyhole acquisition scheme, in which sequential acquisitions of peripheral lines of k-space were interleaved with the acquisition of central lines. Temporal resolution between acquisitions of the central lines was 0.5 s. 0.8 mmol/kg body weight Gd-DTPA (Magnevist, Schering AG) was injected from a tail vein (injection speed 0.5 mL/s) during the DCE-MRI acquisition.

Data processing was carried out using software developed in-house using MATLAB (MathWorks). Sliding-window technique (Fig. 1a) was used to construct a series of k-spaces and compared to another unique technique which introduces a discontinuity at the injection peak (Fig. 1b); that is, the sliding of the k-space window was stopped at before the peak of the AIF, and a completely new k-space constructed using only lines acquired after the AIF peak. The conventional sliding window technique was subsequently resumed. The time of the AIF peak was determined by images constructed by the conventional sliding-window technique retrospectively.

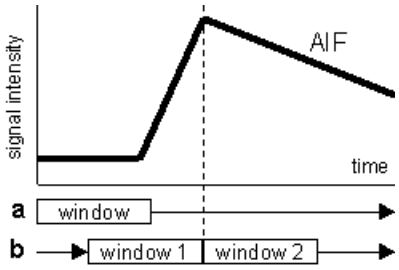

Fig 1 Sliding-window reconstructions. (**a**) conventional sliding. (**b**) sliding with a discontinuity at the peak of AIF.

RESULTS

The image of the aorta lumen shifted along the phase-encoding direction toward the edge of the FOV when compared to the precontrast SPGR T1WI (Figs. 2a and 2b). Even though the axes of readout and phase-encode were exchanged in another rat DCE-MRI study, the shift again occurred in the phase-encoding. The maximum 4-pixel shift was observed in this study. No shifting of the aorta lumen image along the readout direction was observed in either of the rat DCE-MRI studies. Figs. 2c and 2d show the time evolution of the shifting aorta lumen image along the phase-encoding direction on DCE-MRI around the Gd-DTPA injection time. In this example the shifting began with the start of the enhancement when the 141st image was acquired. Subsequently, the aorta lumen image returned to its original position, as observed on the precontrast SPGR T1WI (Fig. 2b). On DCE-MRI by the sliding-window reconstruction technique with a discontinuity at the AIF peak, the aorta lumen image shifted only during the rise time of the AIF and returned to its original position when the AIF was at the peak (Fig. 2d).

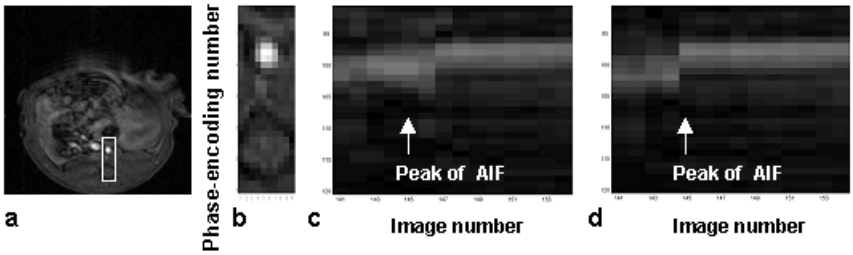

Fig 2 **a**: Rat abdominal 2D SPGR T1WI (TR/TE = 13.9 ms/2.3 ms, flip angle = 15°) before DCE-MRI acquisition. **b**: Portion including an aorta and a spine from the image **a**. **c-d**: At the start of the enhancement due to Gd-DTPA injection, the aorta image shifted in the phase-encoding direction (vertical) and then returned to its original position shown in the precontrast image **b**. Images **c** and **d** are constructed by sliding-window techniques shown in Figs. **1a** and **1b** respectively.

DISCUSSION

Paramagnetic contrast agents affect the bulk magnetic susceptibility and hence the Larmor frequency. The effect of contrast agent phase shift ($\Phi_{Gd}(t)$) can be expressed as (3): $\Phi_{Gd}(t) = P \cdot TE \cdot C_p(t)$, where $C_p(t)$ is the paramagnetic concentration, and P is a function of the paramagnetic agent and the paramagnetic compartment geometry and orientation. Thus, the net phase ($\Phi(t)$) can be expressed as: $\Phi(t) = \Phi_{PE} + \Phi_{Gd}(t)$, where Φ_{PE} is the phase induced by a given phase-encoding gradient. From comparison among the different reconstruction techniques, it was clear that the shift was observed on images whose corresponding inverse (k-space) images contained peripheral lines acquired during the rise time of the gadolinium first-pass (Fig. 2d). Thus the variation of the $\Phi_{Gd}(t)$ during k-space sampling throughout the AIF rise time was large enough to modulate the Φ_{PE}. Consequently, the enhanced aorta lumen image shifted along the phase-encode direction. This misregistration of the AIF pixels must be corrected when measuring AIF. No misregistration was seen along the readout direction as expected from the wide bandwidth per pixel in agreement with Rausch et al. (1).

REFERENCES

1. Rausch M, et al. Magn Reson Imaging 2000;18:1235-1243. 2. Bammer R, Moseley ME. 12th Proc Intl Soc Mag Reson Med 2004 p. 362. 3. Conturo TE, et al. Magn Reson Med 1992;27:375-390.