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Simulation study of the effect of golden-angle KWIC with generalized kinetic model analysis on diagnostic accuracy for lesion discrimination

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ABSTRACT

Purpose: To quantitatively evaluate temporal blurring of dynamic contrast-enhanced MRI data generated using a k-space weighted image contrast (KWIC) image reconstruction technique with golden-angle view-ordering.
Methods: K-space data were simulated using golden-angle view-ordering and reconstructed using a KWIC algorithm with a Fibonacci number of views enforced for each annulus in k-space. Temporal blurring was evaluated by comparing pharmacokinetic model parameters estimated from the simulated data with the true values. Diagnostic accuracy was quantified using receiver operator characteristic curves (ROC) and the area under the ROC curves (AUC).
Results: Estimation errors of pharmacokinetic model parameters were dependent on the true curve type and the lesion size. For 10 mm benign and malignant lesions, estimated AUC values using the true and estimate AIFs were consistent with the true AUC value. For 5 mm benign and 20 mm malignant lesions, estimated AUC values using the true and estimated AIFs were 0.906 ± 0.020 and 0.905 ± 0.021, respectively, as compared with the true AUC value of 0.896.
Conclusions: Although the investigated reconstruction algorithm does impose errors in pharmacokinetic model parameter estimation, they are not expected to significantly impact clinical studies of diagnostic accuracy.

1. Introduction

Dynamic contrast enhanced MRI (DCE-MRI) is an important tool for the diagnosis of cancerous lesions. It provides morphological as well as pharmacokinetic information about lesions that can be combined to improve diagnostic accuracy. In an effort to balance morphological and kinetic information in DCE-MRI data sets, the clinical community often needs to trade-off spatial versus temporal resolution. This is particularly true, for example, when high-spatial resolution data are required for a large three-dimensional field of view, such as encountered in bilateral breast imaging [1].

There have been many approaches to balancing these two conflicting system requirements. One method is to improve scanner hardware (e.g., gradients, parallel imaging, coil arrays) and streamline pulse sequences so that data acquisition is faster. In this way, both spatial and temporal resolution can improve simultaneously. However, even when these methods are employed, current state-of-the-art hardware alone cannot fulfill the ever-increasing demand for higher spatial and temporal resolution. Another approach is to take advantage of information redundancy in images to extract higher spatial and temporal resolution information from undersampled data [2]. Some examples of this method include keyhole imaging [3,4], compressed-sensing [5], and view sharing [6].

Although these algorithms show great promise, they can affect image quality in complicated, unexpected ways that hinder their acceptance by the clinical community [7]. In order to realize the full potential of these temporal acceleration techniques and facilitate their transition into the clinical workflow, image quality must be rigorously evaluated.

Two examples of methods that take advantage of information redundancy are k-space weighted image contrast (KWIC) [8] and golden-angle radial view sharing [9]. The combination of these two methods can be referred to as golden-angle KWIC (GA-KWIC). KWIC is a view sharing technique that uses a radial k-space acquisition and exploits the fact that the center of k-space is oversampled to reduce the number of views that contribute to lower resolution regions of k-space. Since the center of k-space is the primary determinant of image contrast, the temporal resolution can be increased to approximately the time it takes to acquire the reduced number of views in this region. This is achieved by continuously acquiring radial views and shifting the reconstruction window by the number of views in the center of k-space. With this method, the outer region of k-space maintains a full sampling of radial views in order to preserve spatial resolution. In this way, very high temporal resolution can be achieved in post-processing without detrimentally affecting the spatial resolution of the images. However, since different numbers of views contribute to

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different regions of k-space, the temporal resolution of a KWIC reconstructed data set is a complicated, poorly-defined quantity that depends on the combination of low- and high-frequency structures that make up an object under evaluation. Although temporal blurring of KWIC reconstructed data has been qualitatively examined [10], the authors are unaware of any published quantitative evaluations of this effect.

Golden-angle radial imaging is a method of continuously acquiring radial k-space data such that each view is separated by a constant angle, 111.25°, the so-called "golden angle" [9,11]. By acquiring data in this fashion, any number of consecutive views will provide a relatively even sampling of k-space, and optimal k-space coverage is achieved when a Fibonacci number of consecutive views is selected [9,12,13]. This technique has the advantage that temporal resolution can be arbitrarily selected and optimized on an ex post facto basis. This property complements the KWIC algorithm since it allows the exact view weighting to be optimized in post-processing.

In this study, we quantitatively evaluate the temporal blurring of DCE-MRI images reconstructed using a GA-KWIC algorithm by examining the effect on estimated pharmacokinetic model parameters. Temporal blurring of both lesion and arterial input function (AIF) curves is taken into consideration. The dependence of model parameter estimation on kinetic curve type is examined since we expect quickly varying curves to be more strongly impacted than slowly varying curves. In addition, since KWIC reconstruction applies different weighting to low- and high-resolution regions of k-space, we explore how pharmacokinetic model parameter estimation depends on lesion size. Errors in model parameter estimation are directly translated into the impact of the GA-KWIC algorithm on diagnostic accuracy for the separation of benign and malignant lesions.

2. Methods

2.1. Object model

The simulations in this study were performed using two different object models; object A to assess the temporal blurring depending on lesion size and object B to assess the temporal blurring depending on contrast enhancement characteristics (see Fig. 1). Both objects consist of a two-dimensional (2D) circular component (150 mm diameter), representing normal tissue, in a flat background, representing air. Each object contains a set of internal circular shapes to represent lesions or arteries. Kinetic curves of the lesions and arteries are described by the generalized kinetic model (GKM) [14] and a population AIF [15], respectively.

Object A contains a 10.0 mm diameter artery and two sets of nine lesions. The first set of lesions, located in the outer region of the object, has kinetic curves with $K^{trans} = 0.08\ \text{min}^{-1}$ and $v_e = 0.30$, where K^{trans} is the volume transfer constant between blood plasma and extravascular extracellular space (EES) and v_e is the volume of the EES per unit volume of tissue, and diameters of 2.0, 4.0, 5.0, 6.0, 8.0, 10.0, 12.5, 15.0, and 20.0 mm. The second set of lesions is located in the inner region of the object and has $K^{trans} = 0.33\ \text{min}^{-1}$ and $v_e = 0.42$ and the same diameters as the first set of lesions.

Object B has an artery, whose diameter can be selected to be either 5, 10, or 20 mm, and twenty circular lesions all with the same diameter, selected to be either 5, 10, or 20 mm. The K^{trans} values of the twenty lesions are 0.01 and 0.1 to 1.9 min^{-1} in increments of 0.1 min^{-1} and $v_e = 0.35$ for all the lesions.

For both objects, the relaxation times of normal tissue and lesions were set to $T_1 = 1495.0$ ms and $T_2 = 66.0$ ms. These values were taken from published values measured *in vivo* of breast fibroglandular tissue in normal volunteers at 3 T [16,17]. The relaxation times of blood were set to $T_1 = 1932.0$ ms and $T_2 = 275.0$ ms [18]. In all cases, T_2^* was arbitrarily assumed to be equal to half of T_2. Relaxivities were set to $r_1 = 3.3\ \text{s}^{-1}\ \text{mM}^{-1}$ and $r_2 = 5.2\ \text{s}^{-1}\ \text{mM}^{-1}$ [19] for all tissue types.

2.2. System model

In order to generate simulated DCE-MRI data, a spoiled gradient echo (SPGR) imaging sequence (flip angle = 12 degrees, repetition time = 3.6 ms, echo time = 1.47 ms, and 1.0 mm × 1.0 mm pixel resolution) with golden-angle radial view-ordering scheme [9] and a three-dimensional (3D) stack-of-stars acquisition was assumed. Although data were generated in 2D, it was assumed that 35 total slices were acquired so that the time between acquisition of each radial view was 35 × 3.6 ms = 0.126 seconds, as in a 3D acquisition.

Each acquired radial view was simulated by first calculating the relaxation values of each object component assuming the fast exchange limit and generating a true MR signal map with 1.0 mm × 1.0 mm pixels using the SPGR signal equation [20]. The Fourier transform of this true MR signal map was then calculated to generate a Cartesian-sampled version of k-space, and the values for the radial view were estimated using bilinear interpolation. Relaxation values were updated for each radial view, but it was assumed that all the samples for a given view were acquired instantaneously. Each radial view was oversampled in the readout direction by a factor of two in order to simulate a two-fold oversampling scheme often used in radial sampling sequences.

Reconstructed images were produced from the series of radial views by selecting a continuous set of 610 radial views and then, optionally, applying a modified version of the *k*-space weighted image contrast (KWIC) algorithm [8]. The KWIC algorithm has been previously implemented using views that are acquired at sequential

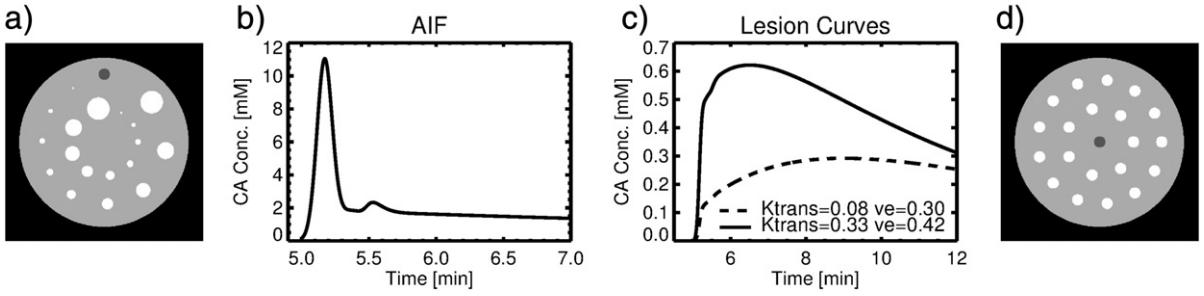

Fig. 1. a) Object A consists of a circular region of normal tissue that contains one circular artery and two sets of 9 lesions with varying sizes. One set of lesions (outer ring) has a slow enhancing kinetic curve and the second set (inner ring) has a fast enhancing kinetic curve. b) Kinetic curve for the artery (a population AIF). c) Slow and fast enhancing kinetic curves for the lesions. d) Object B consists of one 10 mm circular artery and twenty circular lesions (diameter = 5, 10, or 20 mm) in a circular background of normal tissue (diameter = 150 mm). Each of the lesions has a different kinetic curve. Objects A and B are shown with the background areas cropped for ease of visualization. See text for detailed object specifications.