Moving-buffer *k-t* BLAST for real-time reconstruction: Cartesian & simplified radial cases

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INTRODUCTION Recently, we introduced *k-t* BLAST and *k-t* SENSE, which allow significant acceleration in dynamic imaging by exploiting spatiotemporal correlations within the data [1]. The underlying premise is that for most natural objects, the signal intensities in dynamic images can be represented compactly in *x-f* space (*x* = spatial position, *f* = temporal frequency). Since the *f* axis in *x-f* space is related to time *t* by a Fourier transform and the Fourier transform is integrated over all time points, the straightforward implementation of *k-t* BLAST requires the reconstruction to start only after complete data collection.

To make on-the-fly reconstruction possible, we propose in this work to replace the Fourier transform with a short-time Fourier transform. This is equivalent to using a moving buffer that holds the data from several latest time frames, and repeatedly reconstructing the data in the buffer. This approach opens up the possibility of real-time applications that require near-instantaneous visualization, such as MR guided interventions. To reduce computational load, we applied *k-t* BLAST separately to each coil and combined the resulting images from all coils by root mean square. Both Cartesian and radial *k*-space trajectories were examined. For the latter, we show an approximation of the reconstruction formula, which reduces to an efficient Wiener filter. Finally, we address two key questions of the moving-buffer approach: 1) the needed length of the moving buffer, and 2) the impact of "edge effects" due to discontinuities between the first and last time frames of the buffer. The latter determines the minimum latency of the reconstructed image.

THEORY, MATERIALS & METHODS Simulation was performed to investigate the effects of the buffer length and latency, using a set of fully sampled 2D cardiac images with 24 cardiac phases acquired in the short-axis view within a breath-hold. Assuming a typical heart rate of 60 beats per minute, these images were representative of the image contents seen at 24 frames per second in real-time cardiac imaging. The images were decimated to simulate 8x acceleration and were reconstructed by *k-t* BLAST using 5 central phase-encode lines (at full density) as training data [1]. The reconstruction error was quantified as the root-mean-square difference between the reconstructed and original images. This was repeated with the moving buffer placed at different starting points in the cardiac cycle and with equivalent *k-t* sampling patterns (which are circularly shifted from one another).

Real-time cardiac imaging was performed with a balanced steady-state free precession (SSFP) sequence on a Philips Intera 1.5T scanner (Philips, Best, the Netherlands) using a 5-element phased-array coil. For Cartesian *k*-space sampling, the parameters were: TE/TR/flip=1.52ms/3.04ms/60°, 8x acceleration with interleave order (1,4,7,2,5,8,3,6), 15 phase-encode lines/frame with 8 central lines as training data, 2.43x2.71x8.00mm^3 resolution, and 48ms/image. For radial sampling, the parameters were: TE/TR/flip=1.38ms/2.80ms/60°, 5x acceleration with interleave order (1,3,5,2,4), 14 radial lines/frame, 2.22x2.22x10.00mm^3 resolution, and 39.2ms/image. Training data for radial sampling were obtained directly from the oversampled *k*-space center. Reconstruction of the radial data was simplified by exploiting the observation [4] that the *x-f* point spread function is sufficiently close to a delta function, except for aliasing that occurs as ring-like structures at discrete harmonic frequencies (Fig. 1). In general, these aliasing rings are diffuse and therefore rather benign. The exception is the aliasing originating from the time-invariant (i.e. DC) portion of the image signals. Since the DC signals are very strong, even small signal leakage from them can affect image quality significantly. In *k-t* BLAST, the separate treatment of the DC term [1] takes care of such signal leakage, so this problem is largely resolved. For the rest of *x-f* space, it is sufficient to approximate the *x-f* point spread function as a delta function (i.e. encoding matrix becomes an identity matrix in *x-f* space). In that case, each *x-f* voxel can be reconstructed separately. The reconstruction formula [1] reduces exactly to a Wiener filter in *x-f* space:

k-t BLAST [1]: $$\rho = \underline{\rho} + \mathbf{\Theta E^H}\left(\mathbf{E\Theta E^H} + \mathbf{\Psi}\right)^{\dagger}\left(\rho_{alias} - \mathbf{E}\underline{\rho}\right)$$

Approximation: $$\rho_i \approx \underline{\rho}_i + \frac{\mathbf{\Theta}_{i,i}}{\mathbf{\Theta}_{i,i} + \mathbf{\Psi}}\left(\rho_{alias} - \mathbf{E}\underline{\rho}\right)$$

where ρ, $\underline{\rho}$, $\mathbf{\Theta}$, $\mathbf{E}$, $\mathbf{\Psi}$, ρ_{alias}, $(\cdot)_i$ and $(\cdot)_{i,i}$ represent the reconstructed *x-f* voxel signals, the time-averaged (i.e. DC) *x-f* voxel signals, the signal covariance, the encoding matrix, the noise variance, the aliased *x-f* voxel signal, the *i*-th element, and the *i*-th diagonal element, respectively.

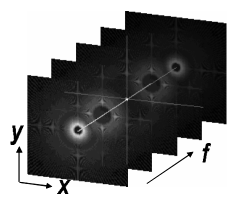

Fig. 1 Point spread function in *x-f* space for radial sampling at 5x acceleration.

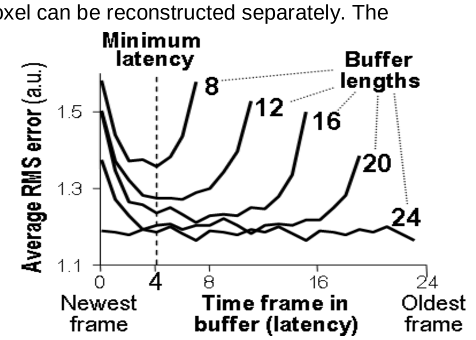

Fig. 2 Simulation results evaluating the effects of buffer length and latency on average reconstruction error.

RESULTS Fig. 2 shows the simulation results. The reconstruction error significantly increased at the first and last frames of the buffer, due to temporal discontinuity as expected [1]. This error upshoot was not observed for the largest buffer length of 24, since that buffer length covered the entire cardiac cycle, thereby eliminating the discontinuity. In general, error decreased with increasing buffer length, but there was a diminishing return with buffer lengths greater than twice the acceleration factor (16 in this case). In general, if the imaged contents remain similar (e.g. with no abrupt change in view plane), longer buffers are preferred, albeit at the expense of increased computation. Practically, a buffer length of 3- to 5-fold the acceleration factor is recommended to minimize error. For a given buffer length, the reconstruction error decreased quickly after a short latency and remained at the same level towards the middle of the buffer. A latency corresponding to ~50% to 100% of the acceleration factor (i.e. ~4-8 frames for 8x acceleration here) is generally recommended. These recommendations (buffer length ~ 3-5 x acceleration, latency ~ 50%-100% of acceleration factor) were used for reconstructing the data from real-time cardiac imaging (Fig. 3). In both Cartesian and radial sampling, the high acceleration led to considerable undersampling. The effects of undersampling were significantly minimized in both cases after *k-t* BLAST reconstruction. Overall image quality was good, and the myocardial contraction was depicted accurately. However, slight residual artifacts were discernible occasionally.

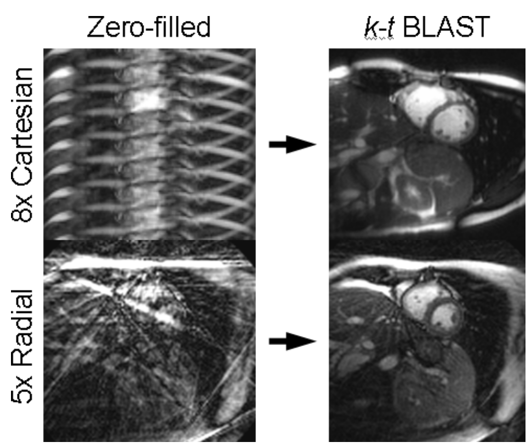

Fig. 3 Real-time cardiac imaging with Cartesian (top) and radial (bottom) *k*-space sampling. Images obtained by (left) zero-filled Fourier reconstruction and (right) *k-t* BLAST.

DISCUSSION & CONCLUSION We showed that by using a short-time Fourier transform, it was possible to apply the *k-t* approach to real-time reconstruction with a small amount of latency. The computational load is sufficiently small and it can be highly parallelized to be handled in real-time with today's hardware.

REFERENCES [1] Tsao J, *et al*. MRM; 50:1031-42, 2003. [2] Madore B, *et al*. MRM; 42:813-28, 1999. [3] Tsao J, *et al*. MRM; 46:652-60, 2001. [4] Larson A, & Simonetti O. ISMRM, abstract 77, 2002.