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Computer modeling of the spatial resolution properties of a dedicated breast CT system

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Abstract

Computer simulation methods were used to evaluate the spatial resolution properties of a dedicated cone-beam breast CT system. x-ray projection data of a 70 μm nickel-chromium wire were simulated. The modulation transfer function (MTF) was calculated from the reconstructed axial images at different radial positions from the isocenter to study the spatial dependency of the spatial resolution of the breast CT scanner. The MTF was also calculated in both the radial and azimuthal directions. Subcomponents of the cone beam CT system that affect the MTF were modeled in the computer simulation in a serial manner, including the x-ray focal spot distribution, gantry rotation under the condition of continuous fluoroscopy, detector lag, and detector spatial resolution. Comparison between the computer simulated and physically measured MTF values demonstrates reasonable accuracy in the simulation process, with a small systematic difference ($\sim 9.5 \pm 6.4\%$ difference, due to unavoidable uncertainties from physical measurement and system calibration). The intrinsic resolution in the radial direction determined by simulation was about 2.0 mm^{-1} uniformly through the field of view. The intrinsic resolution in the azimuthal direction degrades from 2.0 mm^{-1} at the isocenter to 1.0 mm^{-1} at the periphery with 76.9 mm from the isocenter. The results elucidate the intrinsic spatial resolution properties of the prototype breast CT system, and suggest ways in which spatial resolution can be improved with system modification.

Keywords

cone-beam CT; image quality; spatial resolution; MTF; computer simulation; breast cancer

I. INTRODUCTION

The spatial resolution of a computed tomography (CT) system is dependent on a number of system parameters, each of which has an influence on the overall modulation transfer function

(MTF) of the system. The modulation transfer function, as an important quantitative parameter to evaluate the spatial resolution of a CT system, has been extensively explored.^{1–10} Common methods to measure the MTF of a CT system include using a line-pair pattern phantom,^{5,10} a thin metal wire,^{9,11} or a thin metal foil² to generate a line spread function (LSF) or point spread function (PSF). Pendant-geometry cone-beam CT systems have been developed and evaluated in our laboratory as a potential imaging strategy for breast cancer screening and diagnosis over the past six years.^{12–16} In a previous report,⁸ the MTF of a prototype breast CT system was physically measured, and the system MTF was determined for a number of different imaging conditions. To better understand the role that each of several subcomponents of the imaging system plays on the system MTF, computer simulations were used to model the spatial resolution properties of the prototype breast CT scanner. The computer simulations required that some of the physical properties of the scanner be characterized and thus the two-dimensional x-ray distribution of the focal spot, the MTF of the flat panel detector in planar acquisition mode, and the temporal lag properties of the flat panel detector were physically measured. The physical measurements of these subcomponents of the breast CT scanner imaging chain are described in Sec. II. The computer simulation technique is described in Sec. III. Finally, the simulated MTFs are compared with physical measurements of the breast CT scanner in Sec. IV.

II. METHODS: PHYSICAL MEASUREMENTS

A. Focal spot measurement

The x-ray focal spot distribution was measured with a 30 μm pin hole camera made from a 90:10 gold-platinum alloy (07-613, Nuclear Associates, Carle Place, NY). Because the integration times necessary for producing high quality pinhole images were quite long, high resolution mammography film (AD Mammo Fine, FUJIFILM) was used to integrate the pinhole image. A magnification factor of 5:1 was applied. The processed film image of the magnified focal spot was digitized with a pixel size of 118 μm . While the two-dimensional image of the focal spot distribution was measured, the resolution properties of the breast CT scanner were considered only in the native reconstruction plane. Accordingly, the two-dimensional focal spot distribution was integrated along the z dimension to produce a one-dimensional distribution. This profile was used in computer simulation to model the focal spot distribution.

B. Detector lag measurement

For large area flat panel detectors using a-Si:H photodiodes, charge trapping and other capacitive effects can and do cause detector lag.^{17–20} When a detector is working in continuous acquisition mode, as in fluoroscopy or some cone-beam CT systems working under continuous fluoroscopy mode, some fraction of the signal generated in a given frame will be recorded in subsequent frames. Detector lag is normally a combined result from different charge releasing mechanisms, each of which can be modeled as an exponential function with different time constants. For an individual detector element, or a photodiode, the output signal $g(t)$ is determined by the convolution of the input signal (no lag) $x(t)$, with the detector response function $h(t)$ in the time domain.

To measure the temporal response of the detector used in the breast CT system, the detector panel was illuminated with a number of different x-ray intensities (0.5, 1, 3, 5, and 7 mA). The input signal was assumed to be a right-sided Heaviside function during the acquisition and is defined as

$$x(t) = \begin{cases} 0 & (t < 0) \\ kA & (t > 0) \end{cases}, \quad (1)$$

where A is the signal level and k is a constant of proportionality. A is linearly proportional to the mA values ($A = 0.5, 1, 3, 5, 7$), and t is frame number, $t = 1, 2, 3, \dots, 530$.

A total of 530 frames were acquired continuously with a frame rate of 30 frames per second. The response function $g_0(t)$ ($t = 1, 2, \dots, 530$) was assumed to be uniform between individual detector elements. The gray scale values from a $60 \text{ pixel} \times 90 \text{ pixel}$ region, corresponding to a $23 \text{ mm} \times 35 \text{ mm}$ region of interest located near the center of the image, were averaged to determine $g_0(t)$ for each frame.

The measured function $g_0(t)$ was fit to a function $g(t)$ (Matlab Curve Fitting Toolbox, Mathworks, Natick, MA), which was a summation of multiple exponential functions with different time constants, defined by

$$g(t) = \sum_i f_i A_i [1 - \exp(-t/T_i)], \quad (2)$$

where f_i is a constant scalar. a_i and T_i are the weighting factor and time constant of each term, respectively.

The detector response function or the rising-edge step-response function (SRF) $h(t)$ was determined by the deconvolution of $x(t)$ from $g(t)$. Once the detector response function $h(t)$ was determined, it was convolved with the simulated projection image in the time domain to generate simulated detector lag.

The rising-edge step-response function $h(t)$ determined in this work is different from the delta-function impulse-response function (IRF) or the falling-edge SRF,¹⁷ since the main focus in this work was to simulate the signal built up between frames during the specific acquisition mode in our system; i.e., continuous fluoroscopy with 30 frames per second and 500 frames per rotation. Given the physical aspects of the acquisition mode, the use of the rising-edge SRF is a better mathematical construct than the falling edge SRF or delta-function IRF. Image noise properties were also not considered in this simulation, since the focus here is on spatial resolution.

C. MTF of the detector

The projection MTF of the detector was measured at 80 kVp with an edge phantom, the same tool used for MTF measurement in the ACRIN DMIST trial.²¹ This phantom consists of a square sheet of niobium mounted on a larger square of aluminum. The phantom was placed on the surface of the detector to eliminate the influence of the focal spot. An oversampled LSF was synthesized from the data in the slightly angled edge image, and the oversampled MTF was computed.⁶

D. Physical measurement of the breast CT scanner MTF

The MTF of the breast CT system was measured previously in our lab and results were reported elsewhere.⁸ The breast CT system uses a CsI based a-Si:H flat panel detector (PaxScan 4030CB, Varian Medical Systems, Salt Lake City, UT), with a detector element dimension of $388 \mu\text{m} \times 388 \mu\text{m}$ when working in the 2×2 pixel binning mode, corresponding to a 1024×768 image matrix. An x-ray generator (HF160, Pantak, East Haven, CT) was coupled with an

x-ray tube (MXR-160/ 20, COMET, Flamatt, Switzerland) with a nominal 0.4 mm focal spot (IEC 336 standard). The scanning parameters used for the breast CT scanner were 80 kVp, with 500 views acquired over 360° at 30 frames per second in 16.7 s. The completed scan was acquired under continuous fluoroscopy mode. For the comparisons used in this work, CT images were reconstructed using a ramp filter.

To measure the MTF, a 70 μm nickel-chromium wire was used to generate the point spread function (PSF) in axial CT images. Oversampling of PSF was implemented by aligning the wire at a small angle ($<2^\circ$) with respect to the axis of system rotation. The PSFs produced in this manner were integrated along one direction to produce the line spread function (LSF). The LSFs and MTFs were measured at four different radial positions from the isocenter of the scanner, where $r = 2.6, 27.2, 51.9$, and 76.9 mm. These physically measured system MTFs were used for comparison with computer simulation results in this work.

III. METHODS: COMPUTER SIMULATION

A. Computer simulation overview

X-ray projection images of a 70 μm wire were simulated by analytical calculation. The x-ray photon path length through the wire, modeled as a cylinder, was computed and the x-ray intensity was projected onto the plane of the detector. The projection images were acquired over 360°, and the PSF and MTF were computed from reconstructed CT images. As illustrated in Fig. 1, the simulations were carried out in a serial manner with a number of physical assumptions. The simulation started with the most optimal situation including an ideal focal spot (point source), no motion artifacts of the wire (similar to a pulsed-fluoroscopy CT acquisition), no detector lag, and with ideal detector spatial response. A series of MTFs were then generated with various combinations of the degradations from the subcomponent MTFs. These system MTFs are defined as:

MTF₀: ideal focal spot, no motion blurring (pulsed-fluoroscopy CT acquisition), no detector lag, ideal detector MTF (MTF = 1);

MTF₁: measured focal spot distribution, no motion blurring (pulsed-fluoroscopy CT acquisition), no detector lag, ideal detector MTF (MTF = 1);

MTF₂: measured focal spot distribution, with motion blurring (continuous fluoroscopy CT acquisition), no detector lag, ideal detector MTF (MTF = 1);

MTF₃: measured focal spot distribution, with motion blurring (continuous fluoroscopy CT acquisition), with measured detector lag, ideal detector MTF (MTF = 1);

MTF₄: measured focal spot distribution, with motion blurring (continuous fluoroscopy CT acquisition), with measured detector lag, actual detector MTF.

Details of each step are described in the following sections.

B. Simulation geometry

The system geometry is illustrated in Fig. 2. This system has the same geometric parameters as the physical breast CT system. The distance between the x-ray focal spot to the isocenter (source-to-axis distance) was $\text{SAD} = 457.5$ mm. The distance between the x-ray focal spot and the detector plane (source-to-detector distance) was $\text{SDD} = 877.1$ mm. The corresponding magnification factor was 1.92.

Since this work focused on the spatial resolution in the axial plane, a two-dimensional (2-D) fan beam system model was developed. A one-dimensional (1-D) detector array was simulated

to have 16 384 detector elements with a pitch of 24.25 μm (1 / 16 of the physical detector elements).

After a piece-wise polynomial fitting of the profile of the focal spot distribution, a total of N_s source-let samples were used to simulate the finite sized focal spot distribution. Each source-let with index i , was assumed to be an ideal point source with photon intensity $S(i)$, $i = 1, 2, \dots, N_s$.

Each individual x-ray beam path BP_{ij} was emitted from each source-let with index i , and terminated at each detector element with index j ($i = 1, 2, \dots, N_s$; $j = 1, 2, \dots, 16\,384$). The simulated wire was defined as a solid cylinder area with a diameter of 70 μm , centered at (x_c, y_c) for orthogonal coordinates, or (ρ, θ) in polar coordinates. If a beam path BP_{ij} crossed the wire cylinder, the chord length l_{ij} was calculated and used to determine the attenuation of x-ray photons.

C. Simulation of x-ray generation, interaction, and absorption

The x-ray spectrum was modeled as 80 kVp with tungsten anode and beryllium window using the TASMIP algorithm.²² A thickness of 0.3 mm copper filtration was added to this spectrum to match the measured half-value layer, 5.3 mm Al, of the physical x-ray beam.

For each x-ray beam path, $S(i)$ was the total number of photons emitted from the corresponding source-let with $\Phi(E)$ as the x-ray spectrum. If the chord length l_{ij} was greater than zero, this photon group interacts with the wire, which has a total linear attenuation coefficient $\mu_w(E)$.

Every photon group is eventually absorbed by the CsI scintillator and the output signal is proportional to the total energy deposition in the scintillator. The output signal I_j , from detector pixel j , can be expressed as

$$I_j = \frac{1}{N_s} \sum_{i=1}^{i=N_s} \left[\int_{E=0}^{E=80} S(i)\Phi(E)e^{-\mu_w(E)l_{ij}} \times [kE(1 - e^{-\mu_D(E)d})]dE \right], \quad (3)$$

where $\mu_D(E)$ is the total linear absorption coefficient of CsI detector and d is its thickness. k is a constant to determine the efficiency of converting x-ray photons into visible light photons. The exact value of k has no effect on the MTF since it will be cancelled out during the inverse logarithm procedure to calculate the attenuation coefficient in CT reconstruction. In this simulation, the x-ray focal spot has N_s source-let to simulate its finite size, and the resultant projection image was averaged over all the source-lets.

As mentioned previously, the focus of this investigation was on the spatial resolution properties of the breast CT scanner, and thus noise issues are not a part of this work. Consequently, quantum and other sources of noise in the CT images were not modeled here.

D. Gantry motion modeling

The CT gantry of the physical breast CT system rotates continuously while the detector operates under continuous fluoroscopy mode during the acquisition. To model the acquisition of the wire image using the 500-view scan that is used on the physical scanner, a total of 9000 projections were first simulated with the gantry operation in a pulsed-fluoroscopy mode or a virtual “step and shoot” mode, such that the wire was stationary in each of the 9000 time samples during the rotation. One of every 18 views from these 9000 views was used for reconstruction, for a total of 500 projections, to simulate the data set acquired without motion during the acquisition of each image. To simulate the physical acquisition environment, 18

adjacent projections from the 9000-view data set were averaged to generate a 500-view data set. This procedure approximated the motion of the gantry around the object for the 33.3 ms frame time of the detector, which was working under continuous fluoroscopy mode. Each of the 18 projections occurred in this 33.3 ms time window corresponding to a 1.85 ms acquisition, with the assumption that the detector was working under pulsed-fluoroscopy mode.

E. Detector lag modeling

The detector response function or the rising-edge step-response function (SRF) $h(t)$ was determined from physical measurements as described in Sec. II B. For an individual detector element in each projection image, to simulate the signal with detector lag, the “lag-free” signal was convolved with the detector response function $h(t)$ in the time domain. This time domain convolution was applied to the 500-view projection data set, following the gantry motion modeling.

F. Image reconstruction

A standard filtered backprojection algorithm was used for image reconstruction.²³ CT images were reconstructed as an 8192×8192 matrix with a pixel size of $19 \mu\text{m}$. Since the effect of reconstruction filter kernel on image resolution has been well studied,⁸ only the ramp filter was used in this work. The measured detector MTF was multiplied by the ramp filter in the frequency domain before the reconstruction to simulate the actual detector response. It is worth noting that in most clinical practices, apodization filters are normally applied to balance the tradeoffs in spatial resolution and noise. Although the implementation of apodization filters may affect the system MTF, the effects from system subcomponents are independent of the apodization filter.

For the physical measurements of system MTFs, CT images were reconstructed as a 512×512 matrix, for a field of view of $156 \text{ mm} \times 156 \text{ mm}$, corresponding to a pixel dimension of $304 \mu\text{m}$ on a side. In the computer simulation, the same field of view was reconstructed with an 8191×8219 matrix, corresponding to a 16-fold higher sampling pitch in each direction of the image. Thus, the oversampling strategy used in physical measurements, by tilting the wire with a small angle from the axis of rotation, was not necessary given the over sampling of the LSFs in the simulated CT images.

G. MTF calculation

To evaluate the shift variant characteristics of the CT MTF, four different wire positions were simulated, with distances of 2.6, 27.2, 51.9, and 76.9 mm from the isocenter, as shown in Fig. 3. These four positions were chosen to match the experimental settings in the physical measurement.⁸ Due to the circularly symmetric nature of CT images, the one-dimensional MTFs in both radial and azimuthal directions (ρ, θ) were calculated, instead of along a Cartesian based coordinate system. The two-dimensional PSF was integrated along the radial (ρ) direction (from isocenter outward) to compute the azimuthal MTFs, and the PSF was integrated along the angular (θ) direction to compute the radial MTF, as illustrated in Fig. 3.

IV. RESULTS

A. Focal spot measurement

The focal spot image and profile are shown in Fig. 4. A piece-wise fitting curve was generated from the measured data with linear interpolation between adjacent data points. In the simulation, the source distribution was sampled from this fitting curve with a sampling space of $4 \mu\text{m}$. A total of 330 source-lets were used to model a source distribution over a 1.32 mm dimension.

B. Detector lag measurement

Detector output signals as a function of time (frame) were measured with different input signal levels (different x-ray intensities). It was found that a three-term exponential function was adequate to model the detector temporal response with different input signal levels (average $r^2 > 0.99$). Thus, Eq. (1) was simplified to

$$g(t) = fA \{ a_1 [1 - \exp(-t/T_1)] + a_2 [1 - \exp(-t/T_2)] + a_3 [1 - \exp(-t/T_3)] \}, \quad (4)$$

where f is a constant scalar. a_i and T_i are the weighting and time constant of each term, respectively.

For the purpose of comparison, $g(t) / \int_{t=0}^{t=500} g(t)$ is plotted in Fig. 5. These results demonstrate that a_i and T_i are not strongly dependent on the input signal level A . Compared with previously reported works,¹⁷ for this breast CT system, the variation of lag between different input signal levels were relatively small. This could be attributed to the fact that different detectors and different frame rates were used. The input signal level range was also relatively smaller compared to previous work (only from 16% to 55% of saturation, compared with from 5% to 90% of saturation), due to the limitation of patient dose in the breast CT application and the x-ray tube output. The input signal for a wire image was almost constant due to the small attenuation generated by the wire. Thus one curve can adequately model the lag in this study. The values used in the computer simulation were from the average of measured results:

$$\begin{aligned} a_1 &= 0.9393, & T_1 &= 0.2067, \\ a_2 &= 0.0406, & T_2 &= 11.0846, \\ a_3 &= 0.0201, & T_3 &= 103.8508. \end{aligned} \quad (5)$$

With a normalized output function $g(t) / \int_{t=0}^{t=500} g(t)$, and a normalized constant input function $x(t) = 1$, the detector response function $h(t)$ was produced from deconvolution of the input signal from the output signal. Figure 6 shows one example output function $g(t)$ and the calculated response function $h(t)$.

C. MTF of the detector

The measured MTF of the detector is shown in Fig. 7. This MTF is consistent with the resolution properties of CsI based thin film transistor flat panel systems with the 388 μm sampling aperture and pitch.²⁴ The measured data were fit into a smooth curve function and implemented in the computer simulation.

D. MTF results comparison with physical measurements

The MTF results from computer simulation were compared with the physical measurement results in Fig. 8. For the physical measurements, the MTFs were measured along the azimuthal direction (about a 4° difference). Thus, only the azimuthal MTFs from simulation were compared with the physically measured MTFs. The quantitative comparisons are shown in Table I. Overall, the average difference between the simulated and physical measured MTFs

was 9.5% (sd = 6.4%), illustrating reasonably good correspondence between the simulations and the physically determined MTF values. The reasons for small systematic discrepancies between the results from simulation and physical measurements could be:

- a. The measurement error from geometrical calibration of the cone-beam system. Even though the calibration error normally has negligible effects on the CT image quality,²⁵ for the extreme situations such as MTF measurement using a thin metal wire, the unavoidable measurement error from system calibration will generate a noticeable effect on the MTF results. As shown in Fig. 9, a small error (0.5 pixels on the detector plane) in the determination of the central ray location²⁵ will cause a noticeable difference of system MTF.
- b. The detector MTF is dependent on the radial position where the LSF was measured, due to x-ray beam divergence; however a single MTF for the detector, measured near the center of the detector, was used for the simulations.
- c. The 2-D distribution of the focal spot was integrated along the z axis for the simulation, but for the physical measurements the 2-D distribution was at play and the wire had a small angle (along the z axis), and this would potentially cause a small discrepancy between the simulation and physical measurements.
- d. As demonstrated from the results of computer simulation, the system MTF was asymmetric. The integration of the 2-D PSF into a 1-D LSF is direction dependent. Small differences in the integration direction between the physical and simulation experiments may therefore lead to slightly different MTFs.
- e. The system MTF is dependent on the radial position where the PSF was measured, and determining this location precisely in the physical experiment was complicated by the fact that a small angulation of the wire in the z dimension was used.

E. MTF results from different system components

Calculated MTF results from computer simulation are shown in Fig. 10–Fig. 13. The azimuthal and radial MTF were calculated separately, for each of the different radial positions. The physically measured MTF was plotted in each figure together with the simulated results, and was labeled “MTF_m.” In each figure, the simulated MTF results show consistent degradation from the ideal MTF₀.

From the MTF results plotted in Fig. 10–Fig. 13, a number of observations can be made:

1. The x-ray focal spot distribution and detector MTF determine the optimal system MTF, both azimuthal and radial MTF, consistently through the field of view, in the absence of motion.
2. Motion blurring caused from continuous fluoroscopy CT acquisition degrades the azimuthal MTF greatly, but has little effect on the radial MTF. The magnitude of the degradation increases with distance away from the isocenter. For the locations where $r = 2.6, 27.2, 51.9,$ and 76.9 mm, the motion paths during the 33 ms image acquisition time are 33, 342, 652, and 966 μm , respectively.
3. Detector lag has a minimal effect on MTF degradation for the modeled prototype breast CT scanner.
4. The detector MTF consistently determines the radial MTF throughout the field of view. For the azimuthal MTF, the effect of detector MTF declines with increasing motion blur caused from continuous fluoroscopy CT acquisition.

5. The azimuthal MTF consistently drops from the isocenter to the periphery. This MTF falloff is caused primarily by gantry motion during continuous fluoroscopy CT acquisition. The azimuthal MTF of this system has a cutoff frequency of 2.0 mm^{-1} at the isocenter and degrades to 1.0 mm^{-1} at the periphery. The simulated results demonstrated the same trend as the physical MTF measurements.
6. The radial MTF does not depend on the location and shows good consistency from the isocenter to the periphery. The radial component of the MTF therefore is relatively immune from the degrading effects of gantry motion during continuous fluoroscopy CT acquisition. The radial MTF of this system has a cutoff frequency of 2.0 mm^{-1} .
7. If we consider 10% MTF as the limit of the system resolution, the best 10% MTF value is around 1.7 mm^{-1} (for the radial MTF, immune to motion blurring caused from continuous fluoroscopy CT acquisition). If the system resolution is only limited by the focal spot size and detector MTF, the smallest object pair can be resolved from the CT image is around $294 \mu\text{m}$, if there is no noise in the image. It is worth noting that in clinical practice, when there is noise present and some type of apodization filter is applied, this number could be quite different. An example was shown in Fig. 14 that demonstrates the change on system MTF from different reconstruction filters. A more detailed discussion of this was given by Kwan *et al.*⁸

V. SUMMARY

Computer simulation methods were used to evaluate the spatial resolution characteristics of a dedicated breast CT system. In the simulations, a much smaller detector element size was chosen to adequately sample the PSF. The role that different subcomponents of the breast CT scanner played in regards to spatial resolution was elucidated by the simulation process. Comparison between the computer simulations and physically measured MTF values also demonstrated reasonable accuracy in the simulation process.

The intrinsic spatial resolution properties of a prototype breast CT system were evaluated systematically, and the results also suggest ways in which spatial resolution can be improved with system modification. For example, it is obvious that gantry motion during continuous fluoroscopy CT acquisition causes a significant reduction in the MTF towards the periphery of the field of view on the current prototype scanner. Modifications to correct this include the use of a pulsed x-ray source synchronized to the detector frame rate, or increasing the frame rate and collecting more views per gantry rotation. As shown in Fig. 15, with the increase of total number of views per rotation, azimuthal MTF degradation caused by motion blurring from continuous fluoroscopy CT acquisition was recovered accordingly. Clearly, the technical capabilities of the x-ray generator, tube, detector, frame grabber, and gantry motor have important ramifications on optimal design options as well.

The physical configuration of the scanner will also impact the amount of detected scattered radiation, and thus a trade off between lower magnification factors and higher scatter levels exists. MTF values as a function of the system magnification factor are shown in Fig. 16. The radial MTF at the isocenter, which is not affected by motion blur caused from continuous fluoroscopy CT acquisition, is determined mainly by the focal spot MTF and detector MTF, projected to isocenter. The system magnification factor determines the weighting between these two MTFs. Higher magnification corresponds to more weighting on the focal spot MTF and less weighting on the detector MTF. Similarly, with lower magnification, the system MTF is determined mainly by the detector MTF. Optimal system magnification factor is achieved when the focal spot MTF and the detector MTF are tuned with respect to each other (Fig. 16). The complexity of these and other tradeoffs¹¹ suggest that computer simulation techniques will continue to be valuable for optimizing the design of cone beam CT scanners, such as the breast scanner in our laboratory.

Another issue left to be addressed for future work is to evaluate the MTF along the axis perpendicular to the axial plane (or the z -axis MTF). Z -axis MTF of this breast CT system has been measured,⁸ and was reported to be better than the measured axial plane MTF. For the z -axis MTF, the 2-D distribution of the x-ray focal spot (both horizontal and vertical) and the detector MTF in the vertical direction will play a more important role than motion blurring and detector lag.

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FIG. 1.

Simulation diagram. At each step, CT images were reconstructed from simulated projection images and corresponding PSF and MTF were achieved.

FIG. 2. CT system geometry. Top view of the CT system geometric parameters applied in the computer simulation.

FIG. 3. MTF calculation. Four different wire positions were simulated. For each position, the 2-D PSF was integrated along the ρ axis to achieve the 1-D LSF (and the associated MTF) in azimuthal direction. Similarly, the 2-D PSF was integrated along the θ axis to calculate the radial LSF and MTF.

FIG. 4. X-ray tube focal spot measurement. (a) Focal spot image digitized from a mammography film. (b) By integrating the image in (a) vertically, horizontal distribution of the focal spot was measured and fitted to model the focal spot distribution in computer simulation.

FIG. 5. Detector lag measurement. Detector output signals measured as a function of time (frame) with different input signal levels (different x-ray intensities). All the results were fit to a three-term exponential function and normalized for comparison (please refer to the text for details).

FIG. 6.

Detector lag modeling. (a) Detector output signal $g(t)$. (b) Detector response function $h(t)$; note the three different slopes in logarithmical scale, corresponding to three different time constants in the three-term exponential function.

FIG. 7. Projection MTF of the detector. Measured horizontally at 80 kVp with an edge phantom placed on the surface of the detector. The averaged data points were fit into a smooth curve function and applied in computer simulation.

FIG. 8.

MTF results comparison. Azimuthal MTF results from physical measurement and simulation are compared as a function of (a) spatial frequency and (b) radial position. Results from simulation matched well with measurement and demonstrated very similar trends for the azimuthal MTF.

FIG. 9.

Effect of CT system calibration on MTF results. Azimuthal MTF was calculated at 51.9 mm from the isocenter with physical focal spot distribution, motion blurring, detector lag, and real detector MTF. For ideal calibration result (dashed line), exactly same geometrical parameters were used for simulation and reconstruction. For calibration with measurement error (solid line), 0.5 pixels difference in the determination of central ray location (Ref. 25) was used for reconstruction. The results demonstrated that unavoidable measurement error had a noticeable effect on system MTF measurement and was one of the major causes for the small systematic discrepancy between the results from simulation and physical measurements.

FIG. 10. MTF results at 2.6 mm from isocenter. (a) Azimuthal MTF. (b) Radial MTF.

FIG. 11. MTF results at 27.2 mm from isocenter. (a) Azimuthal MTF. (b) Radial MTF.

FIG. 12. MTF results at 51.9 mm from isocenter. (a) Azimuthal MTF. (b) Radial MTF.

FIG. 13.
MTF results at 76.9 mm from isocenter. (a) Azimuthal MTF. (b) Radial MTF.

FIG. 14.

MTF results with different reconstruction filters. Azimuthal MTF was calculated at 2.6 and 76.9 mm from the isocenter with physical focal spot distribution, detector lag and detector MTF. Two reconstruction filters were applied for each position, Ramp and Shepp Logan. By using apodization filters, the system MTF degrades accordingly.

FIG. 15.

MTF results at different CT acquisition modes. Azimuthal MTF was calculated at 76.9 mm from the isocenter with physical focal spot distribution, without detector lag, and with ideal detector MTF. For continuous fluoroscopy modes, different number of views per rotation corresponds to different x-ray pulse duration and duty cycle. With the increase of total number of views per rotation, azimuthal MTF degradation caused by motion blurring from continuous fluoroscopy CT acquisition was recovered accordingly. The system eventually works under pulsed-fluoroscopy mode.

FIG. 16. System MTF as a function of magnification. Only the x-ray focal spot distribution and detector MTF were considered. Magnification of cone beam CT system ranged from 1.1 to 5. Optimal magnification determined from system MTF was in the range from 1.4 to 2.2.

TABLE I

MTF comparison. Azimuthal MTFs from computer simulations were compared with measurement. (Measured MTF values lower than 0.2 were not calculated.)

Radial distance from the isocenter (mm)	Spatial frequency (cycles/mm)						
	0.2	0.4	0.6	0.8	1.0	1.2	1.4
2.6	2.0%	2.3%	6.5%	8.1%	10.4%	6.1%	8.7%
27.2	1.0%	1.2%	1.5%	4.1%	9.4%	9.5%	7.7%
51.9	2.1%	6.9%	16.7%	34.6%	—	—	—
76.9	2.2%	14.0%	42.3%	—	—	—	—
Average	1.8%	6.1%	16.7%	15.6%	9.9%	7.8%	8.2%
Standard deviation	0.5%	5.8%	18.2%	16.6%	0.7%	2.4%	0.7%
							Average
							6.3%
							4.9%
							15.1%
							19.5%
							9.5%
							6.4%