Published in final edited form as:

Med Phys. 2007 March ; 34(3): 1062–1075.

Modeling and measurement of the detector presampling MTF of a variable resolution x-ray CT scanner

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Abstract

The detector presampling MTF of a 576-channel variable resolution x-ray (VRX) CT scanner was evaluated in this study. The scanner employs a VRX detector, which provides increased spatial resolution by matching the scanner's field of view (FOV) to the size of an object being imaged. Because spatial resolution is the parameter the scanner promises to improve, the evaluation of this resolution is important. The scanner's pre-reconstruction spatial resolution, represented by the detector presampling MTF, was evaluated using both modeling (Monte Carlo simulation) and measurement (the moving slit method). The theoretical results show the increase in the cutoff frequency of the detector presampling MTF from 1.39 cy/mm to 43.38 cy/mm as the FOV of the VRX CT scanner decreases from 32 cm to 1 cm. The experimental results are in reasonable agreement with the theoretical data. Some discrepancies between the measured and the modeled detector presampling MTFs can be explained by the limitations of the model. At small FOVs (1–8 cm), the MTF measurements were limited by the size of the focal spot. The obtained results are important for further development of the VRX CT scanner.

Keywords

variable resolution x-ray (VRX) detector; detector presampling MTF; spatial resolution measurement; Monte Carlo simulation

1. INTRODUCTION

Computed tomography (CT) is a proven method of diagnostic imaging. It provides high-contrast sectional and three-dimensional surface-rendered images of *in vivo* anatomy. Depending on the size of anatomy being imaged, CT can be divided into two general domains: clinical CT and micro-CT. Clinical CT scanners are used primarily for whole-body imaging; they have a field of view (FOV) of 40–50 cm but only moderate spatial resolution of 2–3 cy/mm.^{1,2} Although clinical CT scanners can image objects down to a few centimeters in diameter, there is no improvement in the spatial resolution as the object size decreases. Micro-CT scanners, on the other hand, are very appropriate for imaging small objects. Such scanners have spatial resolution approaching 100 cy/mm.^{2,3} However, the FOV of micro-CT scanners is typically only a few centimeters.

A novel type of a CT scanner has been proposed that can image large as well as small objects with greatly improved spatial resolution.^{4,5} The scanner is based on a variable resolution x-ray (VRX) detector – a one-dimensional (1D) discrete detector placed at an acute angle with respect to an incident x-ray beam. By angulating the detector (i.e., orienting the detection medium substantially in the longitudinal, x-ray, direction), the width and sampling distance of physical or virtual detector cells appear to be smaller (a “projective compression” principle), and the spatial resolution can be increased by at least two orders of magnitude (Fig. 1). One of the powerful properties of this technique is that all components of the detector blurring –

including the deposited energy cloud, charge or light spreading, and the physical cell width – are reduced.

The VRX detector has four main advantages. First, it provides a great increase in the spatial resolution. Second, the spatial resolution and FOV of the VRX detector can be varied by changing its angulation. When the detector is angulated according to the object size, large as well as small objects can be imaged, and the highest possible spatial resolution is achieved at each particular FOV. Third, for a relatively thin detection medium, the quantum efficiency of the VRX detector improves as its spatial resolution increases. Fourth, the detector is well suited for the CT scanner geometry.

Although the most obvious application of the VRX detector is a basic, single-slice third-generation CT scanner, the VRX detection technique can also be extended to a multi-slice or cone-beam CT system, which employs a stack of 1D detectors or a planar two-dimensional (2D) detector such as a flat-panel array. An angulated stacked detector will clearly provide an increase in the in-plane spatial resolution, while leaving the cross-plane resolution dependent solely on the height of the detector cells as well as on collimation. In the case of a flat-panel detector, it may be possible to enhance spatial resolution in both directions simultaneously by angulating the single-piece detector accordingly, but this potential has not yet been investigated. One possible limitation of applying the VRX detection technique to CT scanners with 2D detectors is non-isotropic spatial resolution. Another limitation is the difficulty of using any post-patient anti-scatter collimator (like a Bucky grid), which cannot be attached to the entrance surface of the angulating detector but must be maintained perpendicular to the x-ray beam direction.

After establishing the feasibility of the VRX CT scanner concept with a theoretical analysis and experimental results from a 16-channel solid-state detector⁴ as well as from a storage-phosphor plate,⁵ a 576-channel solid-state dual-arm VRX detector has been built and has shown promising results in terms of the signal level, SNR, and spatial resolution.⁶ An analytical and Monte Carlo study of this 576-channel detector has demonstrated several advantages of the dual-arm detector configuration compared with the single-arm configuration.⁷ Therefore, the 576-channel dual-arm detector has been used to construct an experimental VRX CT scanner. Initial assessment of the scanner has included a Monte Carlo study of the x-ray cross-talk and has indicated the need for anti-scatter collimation and optimization of the x-ray tube voltage.⁸ Further examination of the scanner's performance has become possible after development of an accurate reconstruction algorithm specific to the VRX CT scanner. As part of achieving high-quality reconstruction, a method to calibrate the scanner by scanning an off-axis metal pin has been established.⁹ Also, an alternative, four-arm VRX detector for target CT imaging has recently been developed and preliminarily tested, showing the potential for yielding even higher spatial resolution in a target region compared with the original VRX detection technique.¹⁰

An important step in the development of the experimental VRX CT scanner is a comprehensive evaluation of the scanner's performance. In general, the evaluation of the imaging system performance includes such standard metrics as the modulation transfer function (MTF), noise-equivalent number of quanta (NEQ), and detective quantum efficiency (DQE). Because the primary purpose of the VRX CT scanner is to improve spatial resolution, the initial phase of the scanner evaluation has focused on the MTF. Specifically, an intensive theoretical and experimental study of the detector presampling MTF (representing the pre-reconstruction spatial resolution) and the scanner reconstruction MTF (representing the post-reconstruction spatial resolution) has been undertaken in our lab. The results for the detector presampling MTF are presented in this paper. A similarly intensive study of the NEQ and DQE of the VRX CT scanner will be the subject of future investigations. It should be pointed out, however, that

both the spatial resolution and the quantum efficiency of the VRX detector increase as the angle between the detector and the incident x-ray beam decreases, as mentioned previously.

The results of the comprehensive theoretical and experimental study of the detector presampling MTF are presented in this paper. The detector presampling MTF was modeled by the Monte Carlo method and measured by the moving slit technique. The theoretical results show the increase in the cutoff frequency of the detector presampling MTF from 1.39 cy/mm to 43.38 cy/mm as the FOV of the VRX CT scanner decreases from 32 cm to 1 cm. The experimental results support the theory, except for the small FOVs (< 4 cm), where the accurate MTF measurements up to the cutoff frequency could not be made due to the focal spot limitation. The discrepancies between the measured and the modeled detector presampling MTFs can be explained by the limitations of the model.

2. THEORY

2.1. VRX CT scanner

A typical VRX CT scanner (Fig. 2) includes a dual-arm VRX detector. Such a detector is preferable for use in CT scanners because it provides left-right symmetry, low system magnification, small variations in the detector performance from one end of the detector to the other, and a compact system design.⁷ The scanner also includes an x-ray source and support for an object being imaged. On the diagram, the object is represented by a circle. The diameter of this circle (i.e., the maximum size of the object) determines the scanner's FOV.

2.2. Detector presampling MTF

For an imaging system, the MTF is a standard measure of spatial resolution. The MTF describes the transfer of sinusoidal inputs through the system.¹¹ For a digital imaging system, the most relevant measure of spatial resolution is the detector presampling MTF.¹² The detector presampling MTF includes only a detector aperture and blurring in the detection medium. Hence, this MTF describes inherent resolution of one cell in a discrete detector, without the influence of the focal spot, magnification, sampling, or reconstruction. In our study, the detector presampling MTF represents the pre-reconstruction spatial resolution of the VRX CT scanner.

Experimental evaluation of the detector presampling MTF usually consists in measuring the corresponding line spread function (LSF) or edge spread function (ESF). The LSF is defined as the radiation intensity distribution in the image of a line object of unit intensity.¹¹ Similarly, the ESF represents the radiation intensity distribution in the image of a perfectly attenuating edge of unit intensity. Measurement of either the detector presampling LSF or ESF is relatively straightforward in principle. For 1D discrete detectors, the detector presampling LSF is measured by the moving slit/wire method, whereas the corresponding ESF is measured by the moving edge method. For 2D discrete detectors, the primary methods to measure the detector presampling LSF and ESF are with the stationary slanted slit^{13,14,18} and edge,^{15–18} respectively.

Once the detector presampling LSF or ESF is measured, the corresponding MTF can be computed as

$$MTF(f) = k | F\{LSF(x)\} |, \quad (1)$$

$$\text{or } MTF(f) = k \left| F\left\{\frac{d}{dx}ESF(x)\right\} \right|, \quad (2)$$

where $F\{ \}$ represents the Fourier transform, k is a normalization constant, x and f are the spatial and frequency coordinates, respectively.

The same principle applies to the modeling: the detector presampling LSF (or ESF) is first modeled, and the detector presampling MTF is then computed. Thus, the evaluation of the detector presampling MTF, both theoretically and experimentally, reduces to the evaluation of the corresponding LSF (or ESF).

3. METHODS

3.1. VRX CT scanner parameters

To model and measure the detector presampling MTF, a number of parameters for the VRX CT scanner were selected and computed. These include detector parameters, scanner geometrical parameters, and optimum x-ray beam parameters.

3.1.1. Detector parameters—The dual-arm VRX detector used in the experimental VRX CT scanner employed a linear scintillator-photodiode array as its detection system. The array consisted of cadmium tungstate (CdWO_4) scintillator crystals (cells) arranged in 24 modules (12 on each arm) of 24 cells per module (Fig. 3). The cells in every module were separated by inner lead (Pb) separators. Two outer lead separators were placed on the module's ends. Also, there was aluminum oxide (Al_2O_3) reflective paint coating between the cells and the separators in a module. The main parameters of the VRX detector are given in Table 1.

3.1.2. Scanner geometrical parameters—Geometrical parameters of the VRX CT scanner were selected and computed based on the schematic diagram (Fig. 2). There were three groups of parameters: (1) constant parameters, (2) an independent parameter, and (3) dependent parameters. The constant parameters included the source-vertex distance, source-object distance, and active arm length. These parameters were the same throughout the study. The independent parameter was the FOV; it was varied in a specified range, and the detector presampling MTF was modeled and measured as a function of this FOV. The dependent parameters included the opening half-angle, incident angle, and source-cell distance. These parameters were computed as functions of the FOV. The two latter parameters were also functions of the cell under study. A partial list of the scanner geometrical parameters is given in Table 2.

The active arm length was computed assuming that only 256 out of 288 cells on each arm were used to image an object. These were the active cells. The remaining 32 cells at the end of each arm were reference cells; they were used to correct for variations in the x-ray flux. The number of active/reference cells was selected based on the acceptable maximum error (1.18%) in corrected air-scan images. This error was defined as the standard deviation divided by the mean of the corrected (for the gain, offset, and flux variations) signal from one active cell; the maximum value was obtained by computing and comparing the errors for all active cells.

3.1.3. X-ray beam parameters—The x-ray beam parameters included the tube voltage, tube current, exposure time, filtration, and beam thickness. These parameters were computed from a simple theoretical optimization study. The study consisted in minimizing patient dose while maintaining a fixed contrast SNR and limiting the tube heat.

The contrast SNR for one projection, not for a reconstructed image, was considered. The contrast SNR was computed as

$$SNR_C = \frac{C}{\sigma_C}, \quad (3)$$

where C is the contrast, and σ_C is the standard deviation of the contrast. The contrast was defined by

$$C = 2 \frac{|E_T - E_B|}{E_T + E_B}, \quad (4)$$

where E_B and E_T are the x-ray energies absorbed in a cell under study after x rays passed through the background and target regions of a phantom, respectively. Assuming these energies are independent, the detector is ideal, the system is quantum-limited, and the signal variations are due to photon-counting (Poisson) statistics only, the following expression was derived for the contrast SNR:

$$SNR_C = \frac{|E_T^2 - E_B^2|}{2\sqrt{E_T^2\sigma_B^2 + E_B^2\sigma_T^2}}, \quad (5)$$

where σ_B and σ_T are the standard deviations corresponding to E_B and E_T . Each of these standard deviations was computed according to

$$\sigma = \sum_i^{\text{all energy intervals}} S_i \xi_i^2 \Delta \xi, \quad (6)$$

where $\Delta \xi$ is the energy interval, S_i is the i -th interval value of the respective absorbed spectrum for a cell under study, and ξ_i is the mid-interval energy. A threshold contrast SNR of five was assumed for the optimization study.

The patient dose was represented by the “skin” dose, which was the dose for the outer 2 mm of a water phantom along the scanner centerline (the line connecting the focal spot and the detector vertex). This dose was computed assuming a narrow-beam geometry and scatter in the phantom.^{19,20} In units of cGy, this dose was given by

$$D = \frac{1.602 \cdot 10^{-11}}{\rho} \int_0^{\xi_{\max}} n \xi \left(1 - e^{-\mu \ell} \left(\frac{\mu^*}{\mu} + 0.1 \right)\right) d\xi, \quad (7)$$

where ℓ is the “skin” thickness in cm, ρ is its density in g/cm³, μ and μ^* are the corresponding linear attenuation and energy absorption coefficients, respectively, in 1/cm, 0.1 is the factor describing scatter (according to an assumption²⁰ for energies above 50 keV), n is the input photon fluence in 1/cm², $d\xi$ is the energy interval in keV, ξ is the mid-interval energy in keV, and $\xi_{\max}$ is the maximum energy in the x-ray spectrum in keV. The linear attenuation and energy absorption coefficients to compute both the dose in the phantom and the absorbed spectrum in a detector cell were taken from the NIST XCOM database.²¹

The beam thickness was set by a slice-thickness collimator placed at 70 cm from the focal spot. The collimator opening was proportional to the FOV, with the scale factor of 0.02. The optimization was done for the cell #128, which was the middle active cell on an arm.

A cylindrical water phantom was assumed for the study. The diameter of the phantom was equal to the FOV. As mentioned before, the contrast was formed by x rays passing through two different regions of the phantom. In the background region, x rays were attenuated only by water with the density of 1.0 g/cm³. In the target region, x rays passed through the same water background plus a thin target, which had different attenuation properties than the water background. The diameter of the target was FOV/100. Two types of the target were considered. The first type consisted of water too, but with the density of 0.9 g/cm³. This target provided the contrast based on the density difference, and the corresponding phantom represented a simplified “tissue” phantom. In the second type of the target, 1% of calcium by mass was added

to the water with the density of 1.0 g/cm^3 . The latter target provided the contrast based on the atomic number difference. The phantom with this target represented a simplified “bone” phantom.

The use of water (simulating normal biological soft tissue) and calcium (representing the most attenuating biological substance appreciably present in mammals) in the optimization study is believed to set low and high bounds on the optimum tube voltage. Also, in the authors’ opinion, simple mechanical phantoms (e.g., the simplified “tissue” and “bone” phantoms) containing materials that represent the attenuation limits of biological subjects are preferable, for general system analysis, to complex, task-based anthropomorphic phantoms.

To simplify the optimization procedure, no external filter was considered. Only constant internal filtration (1 mm of Al equivalent in the tube and 2 mm of Al equivalent in the tube collimator) was used. Under the condition of having the contrast SNR of five, the skin dose was computed as a function of the tube voltage at each FOV. The tube voltage that provided the minimum skin dose was selected as the optimum tube voltage at a particular FOV. Because there was no external filter, the required tube heat for the typical exposure time (4 sec) was found below the allowable limits for the Varian G1582BI x-ray tube (0.6 mm nominal focal spot; active cooling)²² used in the VRX CT scanner. Therefore, no additional limiting of the tube heat (hence, the tube current) was necessary.

The computed optimum tube voltage was found independent of the target density, fraction of calcium in the target, and target thickness. The target density was varied in the $0.1\text{--}0.9 \text{ g/cm}^3$ range for the “tissue” phantom, whereas the mass fraction of calcium was changed in the 1–5% range for the “bone” phantom. Also, the FOV-to-target thickness (diameter) ratio was varied from 5 to 100 for both “tissue” and “bone” phantoms. In all cases, the changes only shifted the dose curve up or down on the y-axis without affecting its shape, thus without changing the corresponding optimum tube voltage. The independence of the optimum tube voltage on the target thickness provided strong evidence that spatial-frequency-dependent effects were absent in the cases examined.

Plots of the optimum tube voltage as a function of the scanner’s FOV for “tissue” and “bone” phantoms are shown in Fig. 4. Because the operating voltage range for the x-ray tube and generator was 40–125 kVp, when the optimum tube voltage fell outside this range, the corresponding limiting voltage was used as the optimum voltage. The summary of the main VRX CT scanner parameters used to model and measure the detector presampling MTF are given in Table 3.

3.2. Modeling of the detector presampling MTF

The detector presampling MTF of the VRX CT scanner was modeled by Monte Carlo simulation. Because the VRX detector was a 1D discrete array, the MTF modeling was done by “moving”, with a very small step, a perfect (zero-thickness) pencil x-ray beam along this array and simulating the energy deposited in a cell under study as a function of the beam position. This energy represented the detector presampling LSF, from which the corresponding MTF was then computed. Because of the ability to use a perfect pencil beam in the Monte Carlo program, there was no need to simulate a slit.

3.2.1. VRX detector model—The VRX detector model was similar to one used in the Monte Carlo study of x-ray cross-talk in the VRX detector.⁸ The model was based on the actual VRX detector, with the main parameters given in Table 1.

In the model, 24 cadmium tungstate (CdWO_4) cells, represented by rectangular parallelepipeds, composed one detector module (Fig. 3). The cells were separated by lead (Pb) separators. Inner

separators were placed between the cells in the module; two outer separators were placed on the edges of the module. Between the cells and the separators, there were gaps corresponding to a reflective paint in the actual detector. Each module included an aluminum oxide (Al_2O_3) base behind the cells. Twelve modules formed one arm of the VRX detector. Only one module with discrete cells was used; all the other modules on that arm and all the modules on the opposite arm were replaced by continuous blocks from a uniform material. These blocks had the same x-ray attenuation properties as the replaced modules.

3.2.2. Monte Carlo simulation—The three-dimensional, ACCEPTP ITS 3.0 Monte Carlo code²³ was used for the simulation. This code provides state-of-the-art Monte Carlo solution of linear time-independent coupled electron/photon radiation transport problems, with or without the presence of macroscopic electric and magnetic fields. Physical rigor is maximized by employing some of the best available cross sections and sampling distributions, as well as the most complete physical model for describing the production and transport of the electron/photon cascade from 1.0 GeV down to 1.0 keV.

The Monte Carlo code was run in the photon-only transport mode. This mode substantially increased the simulation speed while producing results with no significant differences from the results obtained in the coupled electron/photon transport mode. A polychromatic point x-ray source was used to create a pencil beam. The source spectra were generated by the method of Birch and Marshall.²⁴ This method has been shown to produce spectra that are not significantly different from experimental spectra.²⁵ For the spectrum generation, the target angle of the anode of 10 deg and intrinsic aluminum (Al) filtration of 3 mm were chosen, according to the parameters of the x-ray tube used in the experimental VRX CT scanner.²² To verify that the generated spectra were consistent with the actual spectra used for the MTF measurement, the corresponding half-value layers (HVLs) were determined and compared. For the tube voltage of 80 kVp and no external filtration, both the theoretical and experimental HVLs were found to be equal to 6.0 mm of Al within an uncertainty of ± 0.4 mm.

To simulate the detector presampling MTF, only one arm of the VRX detector was considered. The simulation was done at six opening half-angles, corresponding to the six selected FOVs (Table 3). At each opening half-angle, the simulation was repeated for three cells: #1 (vertex cell), #128 (middle active cell), and #256 (last active cell). At 1 cm and 2 cm FOVs, the simulation was also repeated for the cell #283, in order to compare the modeling results with results of the measurement, which could only be done for the cell #283 because of this cell's lower system magnification. For each of these cells, the simulation was repeated twice, with the values of the optimum tube voltage for the "tissue" and "bone" phantoms. At 1 cm and 2 cm FOVs, those values were the same, therefore only one simulation was done with the optimum tube voltage; the second simulation was done with the "typical" tube voltage of 80 kVp, which was the tube voltage most frequently used, according to the literature, to measure the detector presampling MTF. Thus, in total 40 detector presampling LSFs and corresponding MTFs were simulated.

The simulation parameters were the following. The number of samples in each simulated LSF was 512. This corresponded to 512 runs of the Monte Carlo code (each run with a slightly shifted, along the detector array, x-ray source position) to obtain one LSF. The amount of shifting (LSF sampling distance) was equal to 1/16 of the projected cell width. The number of photon histories in each run was 10,000. These values of the simulation parameters were chosen to ensure that the statistical error in the simulated detector presampling MTFs was always less than 0.3%, while the aliasing and truncation errors were always below 0.5%.

3.3. Measurement of the detector presampling MTF

The detector presampling MTF of the VRX CT scanner was measured by the moving slit method,⁴ to obtain a sufficiently oversampled LSF. According to this method – and similarly to the modeling – a narrow slit was moved along the VRX detector, across a cell under study, and the signal (containing many time points) from that cell was recorded. This signal represented the measured LSF. The measured MTF was then computed and corrected for the effects of the focal spot and slit, to obtain the detector presampling MTF.

The moving slit method appeared to be the most suitable technique for measuring the detector presampling MTF in the VRX CT scanner. Because the VRX detector was a 1D discrete array, none of the methods developed for 2D arrays (i.e., with a stationary slanted slit or edge) could be used. Among all the methods for 1D arrays (i.e., with a moving slit, wire, or edge), the use of a slit was preferable due to a higher SNR when measuring high-frequency response.²⁶ Also, as mentioned previously, the ability to move the slit with an arbitrary small step (limited only by hardware) allowed sufficient LSF oversampling, to avoid any aliasing errors.

3.3.1. Measurement setup—An experimental 576-channel VRX CT scanner (Fig. 5) was constructed according to the typical diagram (Fig. 2), except an object being imaged was placed on a rotary table. In addition to the VRX detector, x-ray tube (with its own collimator), and rotary table, the experimental scanner included a slice-thickness collimator, detector electronics, and several controlling computers. The VRX detector could be positioned as far as 2.6 m from the x-ray tube. The design of the scanner allowed vertical as well as lateral alignment of the x-ray tube, slice-thickness collimator, rotary table, and VRX detector.

To ensure accurate angular alignment of the VRX detector with respect to the x-ray tube, the detector was placed on an acrylic plate that had the common pivotal point with the detector vertex (Fig. 6). The plate could be manually rotated with the 0.1 deg step in the ± 2 deg range from the scanner centerline. The plate also had the pre-defined opening half-angles corresponding to the six selected FOVs (Table 3). During the alignment, the scanner centerline was imitated by a thin wire tightened between the detector vertex and the point over the tube's focal spot. The line representing the zero angle on the acrylic plate was then aligned with the wire, i.e. with the scanner centerline.

A custom built slit²⁷ was used for the measurement (Fig. 6). The slit was formed by two 1.1 mm thick tungsten jaws covered with 0.5 mm thick lead plates on both sides. Each jaw was controlled by a separate micrometer (2 μ m graduation) to allow accurate adjustment of the slit width. The slit was placed on a MM-4M-F50 micropositioner (National Aperture, Inc.), which provided the scan speed up to 6 mm/sec and the positioning resolution of 0.5 μ m. The micropositioner was placed on a short aluminum bar, with a pivotal point on one side. This short bar was, in turn, placed on a long aluminum bar, also with a pivotal point on the same side. By rotating the short and long bars around their pivotal points, the slit was “rotated” around the vertical axis with coarse and fine steps, respectively. This allowed very accurate angular alignment of the slit with respect to x rays hitting a cell under study. The rotation step for the long bar (fine alignment step) was 0.05 deg, with the range of ± 2 deg.

To measure the detector presampling MTF, only one arm of the VRX detector was used (Fig. 6). The left arm, as viewed from the x-ray tube, was chosen. This arm was set at the opening half-angle corresponding to the FOV of interest. The other, right arm was set at the 90 deg angle with respect to the scanner centerline. Several cells on the right arm were used as reference cells, allowing post-acquisition correction for x-ray flux variations. To avoid signal saturation in the reference cells, an attenuator was placed in front of these cells. The attenuator was either a 1.0 mm thick sheet of lead (used with high kVp settings) or a 25.4 mm thick piece of acrylic (used with low kVp settings).

The slit was positioned between the VRX detector and the slice-thickness collimator. Reasonable attempts were made to position the slit as close to the detector as possible, to provide minimum system magnification. This was not always possible, however, because a few-millimeter gap had to be left between the slit and the detector edge to allow angular slit alignment, which involved movement of the slit in the source-detector direction. The slice-thickness collimator was placed at 70 cm from the focal spot. The VRX detector, slit, slice-thickness collimator, and x-ray tube were also aligned in the vertical direction.

3.3.2. Measurement procedure—The measurement of the detector presampling MTF of the VRX CT scanner included several steps. First, the VRX detector was set at an FOV of interest (i.e., the left arm was set at the corresponding opening half-angle), the slit was removed, and an air scan was acquired. For the air scan, the opening of the slice-thickness collimator was 0.5 mm, the tube voltage and current were 80 kVp and 100 mA, and the exposure time was 1 s. The resulting image was processed to obtain a file with the gains for the detector channels.

Then, the slit was put back in such a position that x rays passing through the slit would hit a cell under study during the slit movement. The cells #1, #128, and #256 were typically chosen as the cells under study. At 1 cm and 2 cm FOVs, however, only the cell #283 was considered because high system magnification for cells closer to the detector vertex prohibited reliable measurement of the detector presampling MTF for those cells.

The cell #1 (vertex cell) was chosen for the measurement because it was important to examine cells in the extreme positions of the detector array (all other cells were intermediate) since they would give the extreme MTF curves. The fact that there was a difference in the response of a vertex cell compared with other cells, because a vertex cell had no adjoining cell on one side making full contact as all other cells do, was not a problem. This was because the complete MTF was measured using only one cell (presampling MTF). However, even if a series of adjacent cells were used to measure the MTF (postsampling MTF), the vertex cell should still be included. There would just be two slightly lower points (from the two vertex cells) on the LSF, but the result would represent the actual situation.

While placing the slit in front of the cell under study, rough initial alignment of the slit was done, to ensure an approximate 90 deg angle between the slit and the x rays hitting the cell. Then, more careful rotational alignment of the slit was performed. A series of LSF measurements was made by rotating the slit in increments of 0.05 deg. The integrals of the measured LSFs were computed and plotted versus the slit rotation angle. The angle with the maximum LSF integral was the optimum slit rotation angle. The slit was set at this angle and considered aligned with the cell under study.

For the slit alignment, the opening of the slice-thickness collimator and the slit gap were 3.8 mm and 10–20 μm , respectively. These settings ensured a high x-ray flux and, hence, a high SNR in the LSF measurements. The tube voltage, current, and exposure time were 80 kVp, 100–250 mA, and 4 s, respectively.

After the slit was aligned, a procedure was performed to find the effective slit gap. The procedure included a series of LSF measurements with the decreasing slit gap. Several values of the slit gap in the range of 6–50 μm were used. The integrals of the measured LSFs were computed and plotted as points versus the slit gap. A linear fit to these points was then found. The abscissa of the intersection of this fit with the x-axis represented a “zero” slit gap. The effective slit gap was then computed by subtracting the “zero” slit gap from the actual slit gap, given by the micrometer readings. Because the “zero” slit gap was usually negative (i.e., x rays were passing even through the fully closed slit), the effective slit gap was usually greater than

the actual slit gap. For this procedure, the slice-thickness collimator had the opening of 0.5 mm, the tube voltage and current were 80 kVp and 100–250 mA, and the exposure time was 4 s.

After the slit was aligned and its effective gap was found, the actual LSF data used to compute the detector presampling MTF were acquired. The data were taken at three different tube voltages: (1) the optimum tube voltage for the “tissue” phantom; (2) the optimum tube voltage for the “bone” phantom; and (3) the “typical” tube voltage of 80 kVp, which was the most frequently-used voltage, according to the literature, for measuring the detector presampling MTF. Ten LSF recordings were acquired at each tube voltage. This normally resulted in 30 LSF recordings for each cell under study at each selected FOV. The situation was slightly different at 1 cm and 2 cm FOVs, at which the optimum tube voltages for the “tissue” and “bone” phantoms were the same. Thus, at each of these two FOVs, only 20 LSF recordings were acquired for each cell under study.

When taking the actual LSF data, the opening of the slice-thickness collimator was 0.5 mm. The exposure time was 4 s. The tube current was the maximum allowable by the x-ray tube and generator for the chosen tube voltage and exposure time; this current ranged from 250 mA to 400 mA. The scan speed of the micropositioner was chosen to provide the total length of each LSF approximately equal to 10 projected cell widths. Each LSF included about 4000 samples; this resulted in the LSF sampling distance of roughly 1/400 of the projected cell width. Such fine LSF sampling ensured aliasing-free MTF results. It also ensured that the blurring in the LSF signal due to the slit motion during acquisition was extremely small (around 0.25% of the projected cell width) and could be neglected. The actual (given by micrometer readings) slit gap was typically 10 μm . At lower tube voltages at smaller FOVs, however, the slit gap was increased to provide an acceptable SNR. Thus, the 40 kVp measurements at 1 cm, 2 cm, and 4 cm FOVs were done with the slit gap of 20 μm , 14 μm , and 12 μm , respectively. The increased slit gap did not affect the measurement results, which were already limited by the increased focal spot size due to the focal spot “blooming” (discussed later) at 40 kVp. Also, at 80 kVp at 1 cm FOV, the slit gap was 6 μm , to allow an accurate LSF measurement with still an adequate SNR.

All LSF data – either actual, used to compute the detector presampling MTF, or auxiliary, used to align the slit and find the effective slit gap, – were processed in the following way. An acquired image was first corrected for gain variations among the detector channels, using the channel gains from the file created during the air scan. The image was then corrected for the electronic offsets by subtracting the same cells’ readings taken when x rays were off. Next, the image was corrected for the x-ray flux variations in time. This was done by dividing the image, sample by sample, by the average of the reference cells’ signals. Finally, the measured LSF was extracted. For an auxiliary LSF (used to align the slit or find the effective slit gap), the extracted data were used with no additional processing. For an actual LSF (used to compute the detector presampling MTF), the extracted LSF was further processed.

The further processing of an actual LSF included, first, an additional offset correction, to make sure the tails of the LSF were at the zero level. This was done by selecting two regions, one on each tail (where the LSF data stopped falling), fitting a straight line through the data in both regions, and subtracting the fitted line from the original LSF. After this correction, the next step was to find the system magnification and LSF sampling distance. Although these parameters could be computed from the measured distances (the source-vertex distance and source-slit distance) and experimental settings (the detector opening half-angle, acquisition sampling time, and micropositioner scan speed), the values would not be accurate enough due to difficulties in accurately measuring the focal spot and slit positions. Therefore, the system magnification and sampling distance were determined using a slit “trace” in the acquired image

(Fig. 7-a). This trace was formed by x rays passing through the slit as the slit was moving across the detector cells. From the trace, the cells traversed by the slit were first found (Fig. 7-b). Then, based on the known physical cell spacing and the detector opening half-angle, the distance S (Fig. 7-c) covered by the slit trace as projected onto the cell's under study plane (i.e., the plane orthogonal to the line passing through the source and the cell under study) was computed. This distance was divided by the number of samples to get the LSF sampling distance. The system magnification was then determined by dividing the sampling distance by the actual distance the slit moved between two samples.

After the LSF sampling distance and system magnification were found as part of the further LSF processing, the measured MTF was computed from the offset-corrected LSF. This MTF was corrected for the effects of the focal spot and slit, and the final, detector presampling MTF was obtained. The MTF correction was done by dividing the measured MTF by the properly scaled slit and focal spot MTFs. The slit MTF was given by a sinc-function corresponding to a rectangular LSF with the width equal to the effective slit gap. The focal spot MTF was accurately measured using the same moving slit method. For this measurement, the slit was placed close to the focal spot, to provide large system magnification (4.7). In addition, the VRX detector arm set at the 90 deg angle with respect to the scanner centerline was used, to exclude any effects of the detector angulation. The focal spot MTF was measured at 80 kVp (250 mA) and 40 kVp (400 mA), yielding the measured focal spot size of 1.23 mm and 1.50 mm, respectively. These values significantly exceeded the nominal focal spot size of 0.6 mm. Also, the 22% increase in the focal spot size at 40 kVp indicated the "blooming" of the focal spot, caused by repelling of the electrons in the tube's electron beam at such low voltage and high current, due to the space charge effect.²⁸ The 40 kVp measured focal spot MTF was used to correct the 40 kVp detector presampling MTFs; all the other detector presampling MTFs were corrected with the 80 kVp measured focal spot MTF. The error introduced by such correction was believed to be negligible because the measured, not assumed, focal spot data were used.

After the detector presampling MTF was measured for one cell under study, the slit was repositioned for another cell under study at the same FOV, and the entire sequence – slit alignment, finding the effective slit gap, and actual MTF measurements – was repeated for the new cell. The same air scan data were used for all the cells under study at the same FOV. After the detector presampling MTF was measured for all the cells under study at one FOV, the detector was repositioned for another FOV, a new air scan was taken, and the entire measurement procedure was repeated.

To summarize, at each selected FOV, one air scan was acquired. For each cell under study at each selected FOV, a set of auxiliary LSF data (at different slit angles) was taken to align the slit, and another set of auxiliary LSF data (for different slit gaps) was taken to find the effective slit gap. Then, at each tube voltage for each cell under study at each selected FOV, 10 recordings of the actual LSF data were acquired. From these data, 10 measured LSFs were extracted, and 10 detector presampling MTFs were computed. From these 10 MTFs, the average MTF and the standard deviations were computed. This average detector presampling MTF with its standard deviations at all points represented the result of the measurement at one tube voltage for one cell under study at one selected FOV.

4. RESULTS

4.1. Modeling of the detector presampling MTF

A sample detector presampling LSF modeled by the Monte Carlo simulation at the "tissue" phantom tube voltage for the cell #256 at 8 cm FOV is shown in Fig. 8-a. The LSF is asymmetrical, due to the angulation of the VRX detector. The tail directed toward the end of the detector arm (left tail in the figure) is higher than the tail (right tail) directed toward the

vertex. A theoretical comparison of symmetrical and asymmetrical LSFs with the same full width at half maximum revealed that the LSF asymmetry does not significantly affect the MTF cutoff frequency but only lowers the MTF curve. A small “plateau” on the left side of the peak of the modeled detector presampling LSF (around -0.03 mm) corresponds to the x-ray beam positions, as the beam moves along the detector array, where x rays are passing only through the reflective paint before entering the cell under study. Because of no attenuation in the reflective paint (the reflective paint material was not modeled), there is a very little change in the energy deposited in the cell and, therefore, the LSF is almost flat over the region equal to the projected width of the reflective paint.

Results of the Monte Carlo simulation of the detector presampling MTF of the VRX CT scanner are shown in Fig. 9 as dashed lines. From these graphs, three important observations can be made.

First, at all FOVs, except for 1 cm and 2 cm, the MTF curves for the “tissue” phantom tube voltage are always lower than the MTF curves for the “bone” phantom tube voltage. This is because the “tissue” tube voltage is always higher than the “bone” tube voltage at those FOVs, and the higher tube voltage results in more cell-to-cell x-ray penetration, which degrades the detector presampling MTF. The discrepancy between the “tissue” and the “bone” MTF curves increases as the difference between the corresponding tube voltages becomes larger. At 1 cm and 2 cm FOVs, the “tissue” and “bone” tube voltages are the same, producing identical MTF curves.

Second, despite the differences in the MTF curves, the MTF cutoff frequencies for the “tissue” and “bone” tube voltages are very close. This is consistent with the idea that the limiting spatial resolution of the VRX detector should not depend on the energy of the x-ray beam. Also, the simulated MTF cutoff frequencies are within approximately 3% of the ideal cutoff frequencies.

Third, as the FOV decreases from 32 cm to 1 cm, the MTF cutoff frequency increases from 1.39 cy/mm (cell #256) to 43.38 cy/mm (cell #283). In addition, the result of the Monte Carlo simulation for the cell #1 at 1 cm FOV (not shown) predicts the MTF cutoff frequency of 53.64 cy/mm. These important modeling results clearly demonstrate the increase in the spatial resolution of the VRX detector with its angulation. Also, at each FOV, the MTF cutoff frequency increases by approximately 20% as one moves from the last active cell (#256) to the vertex cell (#1). Thus, at the smallest FOV modeled (1 cm), the MTF cutoff frequencies in the range of 43–54 cy/mm are predicted.

The Monte Carlo simulation was also used to show the effect of the x-ray beam quality on the detector presampling MTF of the VRX CT scanner (Fig. 10). It is clearly seen that at the incident angle of 11.58° (for the middle active cell at 8 cm FOV), increasing the beam energy from 29 keV to 69 keV (by raising the tube voltage from 40 kVp to 140 kVp and simultaneously increasing the external Al filtration from 0 mm to 5 mm) lowers the spatial frequency by as much as 35% for $\text{MTF} = 0.5$ (Fig. 10-a). But, as already demonstrated (Fig. 9), “hardening” of the beam does not significantly affect the MTF cutoff frequency. Although not generally appreciated, increasing the beam energy lowers the MTF even at normal incidence for all x-ray detectors, because of scattered x-ray and electron range effects. For the VRX detector at normal incidence (Fig. 10-b), the spatial frequency is reduced by 4% for $\text{MTF} = 0.5$ as the beam energy increases from 29 keV to 69 keV.

4.2. Measurement of the detector presampling MTF

A sample measured LSF used to compute the measured detector presampling MTF of the VRX CT scanner is shown in Fig. 8-b. This LSF was acquired at the “tissue” phantom tube voltage for the cell #256 at 8 cm FOV. Similarly to the modeled data, the measured LSF is

asymmetrical, due to the detector angulation. The LSF asymmetry, however, does not much affect the MTF cutoff frequency, but only degrades the MTF curve, as mentioned previously. A small “plateau” present on the left side of the peak in the modeled LSF is less evident in the measured data because of non-zero attenuation in the reflective paint and, hence, a larger change in the energy deposited in the cell under study over the projected width of the reflective paint. Also, the measured LSF, when corrected for the slit and focal spot effects, agrees well with the corresponding modeled detector presampling LSF (Fig. 8-a).

Results of the measurement of the detector presampling MTF of the VRX CT scanner by the moving slit method are shown in Fig. 9 as solid lines. Good quality MTF measurements were obtained for all the cells under study at 16 cm and 32 cm FOVs. At smaller FOVs (1–8 cm), reliable MTF measurements could not be made for the cells close to the detector vertex (i.e., the cells with increased system magnification) because the cutoff frequency of the projected focal spot fell well below the detector cutoff frequency for those cells. Therefore, the detector presampling MTF was measured only for the cell #256 at 8 cm and 4 cm FOVs and the cell #283 at 2 cm and 1 cm FOVs. All the measured MTF curves are relatively smooth despite the noisy LSF data because, first, each curve represents an average of 10 MTFs computed from individual LSFs and, second, the frequency of the noise is typically far above the MTF cutoff frequency.

Again, several important observations can be made based on the results. First, similarly to the modeling results, the measured MTF curves for the higher (usually “tissue”) tube voltage are lower than the measured MTF curves for the lower (“bone”) tube voltage, due to increased cell-to-cell x-ray penetration at the higher tube voltage. Also, the error bars are larger for the lower (“bone”) tube voltage because the SNR was lower in each low-kVp LSF recording and the same number of LSF recordings (10) was acquired to get the averaged MTF at the low and high tube voltages.

Second, the measured MTF cutoff frequencies for the “tissue,” “bone,” and “typical” tube voltages are very close at large FOVs (16–32 cm; except for the cell #1 at 16 cm FOV). At small FOVs (1–8 cm; also for the cell #1 at 16 cm FOV), those cutoff frequencies are difficult to compare because the MTF curves in most cases were terminated at spatial frequencies below the expected cutoff frequencies, to provide the MTF data not degraded by the projected focal spot. Specifically, in each case when the estimated focal spot cutoff frequency was close to or slightly below the detector cutoff frequency, the detector presampling MTF was terminated at a data point with the spatial frequency not exceeding 75% of the corresponding focal spot cutoff frequency. The observed fact that the 40 kVp MTF curves were terminated at much lower spatial frequencies or could not be measured at all is attributed to the increase in the focal spot size due to the “blooming” effect, discussed previously.

Third, the measured MTF curves are always lower than their modeled counterparts. The difference between the corresponding measured and the modeled MTF curves usually increases at a low tube voltage. The lower measured MTF curves can be explained, in the authors’ opinion, by physical phenomena present in the actual VRX CT scanner but not included in the Monte Carlo simulation. Those are the following phenomena: (1) deviations between the true and the modeled detector and slit; (2) cell-to-cell non-uniformities, both geometrical and electronic; (3) cell-to-cell optical cross-talk via the optical epoxy that attaches the scintillator crystals to the photodiodes; (4) x-ray scatter from the various detector components, rotary table, and nearby structures; (5) deviations between the true and the modeled system geometry; (6) electronic noise, cross-talk, and other imperfections; (7) system noise, especially at small angles; and (8) focal spot motion (target wobble, etc.).

Fourth, despite the explained discrepancies between the measured and the modeled MTF curves, their cutoff frequencies agree well in all the cases where those frequencies could be correctly estimated. The measured MTF cutoff frequency increases from 1.38 cy/mm to 5.50 cy/mm for the last active cell (#256) as the FOV decreases from 32 cm to 8 cm (the corresponding modeled values are 1.39–5.55 cy/mm).

The results of the measurement of the detector presampling MTF are partially summarized in Fig. 11, where spatial frequencies are plotted as functions of the incident angle for several MTF values. The data for the “typical” tube voltage of 80 kVp are presented because this voltage was the same at all selected FOVs, allowing a comparison in the entire range of the incident angles. The data clearly show the increase in the measured detector resolution of the VRX CT scanner as the incident angle decreases, even with the high-resolution curves (for $MTF = 0.1$ – 0.3) only partially measured at small incident angles due to the focal spot limitation.

5. DISCUSSION

Several important issues specific to the VRX CT scanner and revealed in the current study should be discussed. First, an x-ray tube with a small focal spot is required for achieving high spatial resolution of which the VRX detector is capable. Because of the relatively large focal spot used, the highest spatial frequency for which the detector presampling MTF could be measured was 20.90 cy/mm (for the cell #283 at 1 cm FOV). To be able to measure the detector presampling MTF up to the highest frequency predicted by the Monte Carlo simulation (53.64 cy/mm for the cell #1 at 1 cm FOV), the focal spot with the actual size less than 0.05 mm is required for the described scanner geometry. Thus, with the sufficiently small focal spot, the VRX CT scanner is expected to provide the pre-reconstruction spatial resolution of more than 50 cy/mm.

Next, the system magnification in the VRX CT scanner varies more from the scanner centerline to the periphery than in other CT scanners. For the described measurement setup, the system magnification changes from 1.035 for the cell #283 at 2 cm FOV to 1.350 for the cell #1 at 1 cm FOV. These values are slightly larger in the actual imaging mode because an object being imaged is positioned farther from the detector compared with the slit used for the MTF measurements. Such variation in the system magnification has a greater impact on the scanner’s spatial resolution, ultimately requiring a smaller focal spot for high-resolution CT imaging.

Also, the beam quality (determined primarily by the x-ray tube voltage) has a direct effect on the pre-reconstruction spatial resolution of the VRX CT scanner. The reason for this is two-fold. First, higher x-ray energy increases the depth of x-ray penetration, which, because of the inherent detector angulation, increases the lateral component of the x-ray penetration length. Second, higher x-ray energy increases the deposited energy cloud size. It should be noted that the second phenomenon occurs in all x-ray detectors.

The obtained results of the theoretical and experimental evaluation of the detector presampling MTF of the VRX CT scanner are of great importance. They show that the measured cutoff frequencies agree well with the modeling, and both the measured and modeled cutoff frequencies are close to the ideal values. The results also demonstrate reasonably good agreement between the measured and the modeled MTF curves. The maximum deviation between the measured and the modeled MTFs ($|MTF_{\text{meas}} - MTF_{\text{mod}}|$) is 0.23 units, and the average deviation, excluding the normalized zero-frequency points, is 0.05 units. In addition, the data indicate that although the detector presampling LSF asymmetry due to cell-to-cell x-ray penetration reduces the corresponding MTF slightly, it does not significantly affect the cutoff frequency. Furthermore, the results underscore the necessity of using a small focal spot

for high-resolution modes of the VRX CT scanner. Overall, the study supports the potential value of the VRX detection technique for high-resolution CT imaging.

6. CONCLUSIONS

A comprehensive evaluation of the detector presampling MTF of the VRX CT scanner was done for the first time and is presented in this paper. The detector presampling MTF was evaluated both theoretically (by Monte Carlo simulation) and experimentally (by the moving slit method). The theoretical results show the increase in the cutoff frequency of the detector presampling MTF from 1.39 cy/mm to 43.38 cy/mm as the FOV of the scanner decreases from 32 cm to 1 cm. The experimental cutoff frequencies agree well with the theoretical values in all cases where such comparison can be made. The agreement between the experimental and the theoretical MTF curves, however, is not as good, but the discrepancies can be explained by physical phenomena present in the actual VRX CT scanner but not included in the Monte Carlo simulation. The detector presampling MTF could not be measured for high-magnification cells at small FOVs (1–8 cm) because of the relatively large focal spot size. But the measurements that could be made support the validity of the MTF modeling by Monte Carlo simulation.

The presented study of the pre-reconstruction spatial resolution of the VRX CT scanner is an important step in the assessment of the scanner's performance. The results of the study will be used for evaluating other image-quality measures as well as for optimizing the scanner's design to maximize its spatial resolution.

Acknowledgements

The authors thank Lawrence Jordan, Herbert Zeman, Joseph Laughter, and Arun Gopal for their help and useful suggestions. This study was partially supported by the NIH Grant No. EB-00418.

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Fig. 1. Conventional (a) and VRX (b) detectors (vertical arrows indicate the projected cell width).

Fig. 2. Schematic diagram of a VRX CT scanner: D_v – source-vertex distance; D_o – source-object distance; D_c – source-cell distance; L_a – active arm length; α – opening half-angle; θ – incident angle.

Fig. 3.
Section of a VRX detector module.

Fig. 4.
Optimum x-ray tube voltage for the VRX CT scanner.

Fig. 5.
Experimental VRX CT scanner.

Fig. 6. Slit and VRX detector setup for measuring the detector presampling MTF (an attenuator in front of the reference cells is not shown).

Fig. 7. Slit trace (a), cells traversed by the slit (b), and geometry used to compute the LSF sampling distance and system magnification (c): α – opening half-angle; S – distance covered by the slit trace as projected onto the plane orthogonal to the line passing through the source and the cell under study.

Fig. 8. Modeled detector presampling LSF (a) and corresponding measured LSF (b) at the “tissue” phantom tube voltage at 8 cm FOV.

(a) FOV = 32 cm (opening half-angle = 53.43 deg)

(b) FOV = 16 cm (opening half-angle = 21.90 deg)

(c) FOV = 8 cm (opening half-angle = 10.60 deg)

(d) FOV = 4 cm (opening half-angle = 5.26 deg)

(e) FOV = 2 cm (opening half-angle = 2.63 deg)

(f) FOV = 1 cm (opening half-angle = 1.31 deg)

Fig. 9.

Measured and modeled detector presampling MTF at 32, 16, 8, 4, 2, and 1 cm FOVs.

Fig. 10. Dependence of the modeled spatial resolution of the VRX detector on the beam quality at 11.58 deg (a) and 90 deg (b) incident angles.

Fig. 11.
Dependence of the measured spatial resolution of the VRX detector on the incident angle at the “typical” tube voltage.

Table 1

Parameters of the VRX detector.

Number of cells	576
Cell material	CdWO ₄
Separator material	Pb
Reflective paint material	Al ₂ O ₃
Cell width	0.79 mm
Inner separator width	0.10 mm
Outer separator width	0.18 mm
Reflective paint width	0.05 mm
Cell height	20.14 mm
Cell thickness	3.00 mm

Table 2

Geometrical parameters of the VRX CT scanner.

Source-vertex distance	150 cm
Source-object distance	106 cm
Number of active cells per arm	256
Number of reference cells per arm	32
Active arm length	25.617 cm
FOV	1–32 cm
Maximum FOV	36.21 cm

Table 3
Modeling and measurement parameters for the VRX CT scanner.

FOV (cm)	Opening half-angle (deg)	Incident angle (deg)			Optimum tube voltage (kVp)	
		cell #1	cell #128	cell #256	"tissue" phantom	"bone" phantom
1	1.31	1.31	1.43	1.58	[40]	[40]
2	2.63	2.63	2.87	3.17	[40]	[40]
4	5.26	5.26	5.75	6.34	60	[40]
8	10.60	10.61	11.58	12.76	100	[40]
16	21.90	21.91	23.87	26.22	[125]	49
32	53.43	53.45	57.55	62.09	[125]	66

Values in [ ] are the tube and generator limits when actual optimum tube voltages fall outside these limits