Photon-counting hexagonal pixel array CdTe detector: Spatial resolution characteristics for image-guided interventional applications

Srinivasan Vedantham, Suman Shrestha, Andrew Karellas,^{a)} Linxi Shi, and Matthew J. Gounis

Department of Radiology, University of Massachusetts Medical School, Worcester, Massachusetts 01655

Ronaldo Bellazzini, Gloria Spandre, Alessandro Brez, and Massimo Minuti

Istituto Nazionale di Fisica Nucleare (INFN), Pisa 56127, Italy and Pixirad Imaging Counters s.r.l., L. Pontecorvo 3, Pisa 56127, Italy

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Purpose: High-resolution, photon-counting, energy-resolved detector with fast-framing capability can facilitate simultaneous acquisition of precontrast and postcontrast images for subtraction angiography without pixel registration artifacts and can facilitate high-resolution real-time imaging during image-guided interventions. Hence, this study was conducted to determine the spatial resolution characteristics of a hexagonal pixel array photon-counting cadmium telluride (CdTe) detector.

Methods: A 650 μm thick CdTe Schottky photon-counting detector capable of concurrently acquiring up to two energy-windowed images was operated in a single energy-window mode to include photons of 10 keV or higher. The detector had hexagonal pixels with apothem of 30 μm resulting in pixel pitch of 60 and 51.96 μm along the two orthogonal directions. The detector was characterized at IEC-RQA5 spectral conditions. Linear response of the detector was determined over the air kerma rate relevant to image-guided interventional procedures ranging from 1.3 nGy/frame to 91.4 $\mu\text{Gy}/\text{frame}$. Presampled modulation transfer was determined using a tungsten edge test device. The edge-spread function and the finely sampled line spread function accounted for hexagonal sampling, from which the presampled modulation transfer function (MTF) was determined. Since detectors with hexagonal pixels require resampling to square pixels for distortion-free display, the optimal square pixel size was determined by minimizing the root-mean-squared-error of the aperture functions for the square and hexagonal pixels up to the Nyquist limit.

Results: At Nyquist frequencies of 8.33 and 9.62 cycles/mm along the apothem and orthogonal to the apothem directions, the modulation factors were 0.397 and 0.228, respectively. For the corresponding axis, the limiting resolution defined as 10% MTF occurred at 13.3 and 12 cycles/mm, respectively. Evaluation of the aperture functions yielded an optimal square pixel size of 54 μm . After resampling to 54 μm square pixels using trilinear interpolation, the presampled MTF at Nyquist frequency of 9.26 cycles/mm was 0.29 and 0.24 along the orthogonal directions and the limiting resolution (10% MTF) occurred at approximately 12 cycles/mm. Visual analysis of a bar pattern image showed the ability to resolve close to 12 line-pairs/mm and qualitative evaluation of a neurovascular nitinol-stent showed the ability to visualize its struts at clinically relevant conditions.

Conclusions: Hexagonal pixel array photon-counting CdTe detector provides high spatial resolution in single-photon counting mode. After resampling to optimal square pixel size for distortion-free display, the spatial resolution is preserved. The dual-energy capabilities of the detector could allow for artifact-free subtraction angiography and basis material decomposition. The proposed high-resolution photon-counting detector with energy-resolving capability can be of importance for several image-guided interventional procedures as well as for pediatric applications. © 2016 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4944868>]

Key words: fluoroscopy, dual-energy imaging, angiography, photon-counting detector, modulation transfer function

1. INTRODUCTION

Minimally invasive, image-guided interventions such as endovascular treatment of unruptured and ruptured intracranial aneurysms have been shown to be a safe and effective alternative to invasive surgical treatments.¹ Endovascular treatment of arterial occlusions such as the use of mechanical devices² has shown favorable outcome for treatment of acute

ischemic stroke. Percutaneous coronary intervention is an established treatment for coronary artery disease. Real-time x-ray image guidance or fluoroscopy is essential for deployment of devices such as stents, flow-diverters, and embolic coils and for local delivery of drugs. Electrophysiology studies and ablation procedures for diagnosis and treatment of arrhythmias also require fluoroscopic guidance. Digital subtraction angiography (DSA) is vital for assessing the

degree of stenosis or occlusion and identifying the presence of perforators that guide device selection for treatment. Post-treatment DSA permits qualitative and quantitative evaluation of the effectiveness of the intervention.

Endovascular image-guided interventions³ are typically performed at dedicated angiography suites. Image intensifier-based systems were the mainstay for several decades, but suffered from veiling glare, distortions, and degradation of image quality with time. Hence, indirect conversion CsI:Tl scintillator coupled amorphous silicon (a-Si) flat-panel detector,^{4–6} tiled large-area charge-coupled devices (CCDs) coupled to CsI:Tl scintillator,^{7,8} and direct conversion amorphous selenium (a-Se) photoconductor-based flat-panel detector⁹ were developed. Current generation of interventional C-arm systems utilize flat-panel detectors with pixel pitch ranging from 143 to 200 μm , depending on the manufacturer and the intended application.^{5,6,10} The spatial resolution achieved by such systems poses a challenge in visualizing small details such as the struts of the stent or the flow-diverter that is needed to evaluate device apposition during deployment. Additionally, once the site of the abnormality to be treated or intervened is determined, imaging can be directed to a smaller field-of-view (FOV) to reduce radiation dose. Hence, high-resolution small field-of-view devices, referred to as microangiographic fluoroscope (MAF), have been developed.¹¹

Photon-counting detectors allow for discrimination and removal of system (electronic) noise with appropriate selection of energy threshold. Cadmium telluride (CdTe) has been studied for x-ray and gamma ray detection due to its high atomic number ($Z_{\text{Cd}} = 48$ and $Z_{\text{Te}} = 52$), high density (5.85 g/cm³), and large band-gap (1.5 eV). The ability to fabricate compact arrays is a contributing factor and CdTe and CdZnTe (CZT) detectors have been incorporated into systems for radionuclide imaging and are being investigated for spectral CT. In this study, we investigated the spatial resolution characteristics of a photon-counting CdTe detector capable of concurrently acquiring up to two energy-windowed images for high-resolution region-of-interest imaging during image-guided interventions. To our knowledge, such detectors have not been investigated for real-time imaging needs during image-guided interventions. The primary objective of this study is to characterize the spatial resolution properties of a thin-layer (0.65 mm) CdTe in combination with small pixel pitch (60 μm along apothem with hexagonal pixels). The secondary objective of this study is to describe the modifications to the methodology for determining the pre-sampled modulation transfer function (MTF) for detectors with hexagonal pixel array. The later objective is pertinent as hexagonal pixel array detectors have transitioned to clinical practice, such as for digital mammography. A preliminary version of this work was presented at the 57th Annual Meeting of the American Association of Physicists in Medicine.¹²

2. METHODS AND MATERIALS

A single module of a photon-counting CdTe Schottky detector (PIXIRAD-I, Pixirad Imaging Counters, s.r.l., Pisa, Italy) was investigated for potential use in fluoroscopic and 2D

digital cine acquisitions for image-guided interventions. The detector was evaluated under IEC RQA-5 spectral conditions¹³ in terms of linearity of response and presampled MTF. The CdTe sensor had hexagonal pixel matrix and provides the advantage of needing 13.4% ($100 \times [1 - (\sqrt{3}/2)]\%$) less pixels for aliasing-free imaging of circular bandlimited signals compared to square pixel matrix.¹⁴ Alternatively stated, the advantage of hexagonal pixel is that the sampling density is higher than a square pixel with similar pitch. For hexagonal arrays, the center-to-center spacing between any pixel and the surrounding six pixels is identical and is equal to $2a$, where a is the apothem. Since current image display devices utilize square pixel matrix, the data acquired with hexagonal pixel matrix require resampling to square pixels for distortion-free display. Hence, the optimum pitch for resampling to square pixels and the MTF after resampling were determined.

2.A. Detector description

Figure 1 shows the schematic and photograph of the investigated photon-counting CdTe Schottky detector. It has a 650 μm thick continuous CdTe layer (Acrorad Co., Ltd., Japan) with a continuous thin-film Pt layer as the cathode, and the anode consists of hexagonally patterned pixels made of multiple thin-film metallic layers (Au/Ni/Au/Ti/Al). The pixelated anode is flip-chip bonded to a complimentary metal-oxide semiconductor (CMOS) application-specific integrated circuit (ASIC) readout with identical hexagonal pixel pattern. The apothem of the hexagonal pixels was 30 μm resulting in pixel pitch of 60 and 51.96 μm along the two orthogonal directions. The study utilized a single module with an active pixel area of 31 $\times$ 25 mm with 512 $\times$ 476 pixels. Each module can be abutted on the two opposite sides along the edge corresponding to the 25 mm dimension, and sensors with eight modules providing a field-of-view of 25 $\times$ 2.5 cm have been developed. Each pixel in the CMOS ASIC has a low-noise charge amplifier (50 e⁻ rms) and two pairs of discriminators and counters, each with 15-bits depth. With appropriate selection of the threshold, the discriminators in each pixel allow detector noise to be removed with only the photons above the threshold being counted. In order to reduce detector noise and polarization effects, which would allow for selection of lower energy threshold that may be needed for some applications, the detector uses thermoelectric cooler with liquid circulation to reduce the operating temperature to approximately -30 to -20°C . The detector does not require cryogenic cooling and dry air is circulated to avoid condensation. Each pixel is equipped with a self-calibration circuit and a single global threshold can be applied to the entire detector. Key specifications of the investigated detector are summarized in Table I. Additional description of the investigated detector was provided in a prior report.¹⁵ For clarity, we refer to a discriminator-counter pair as a channel. The detector allows for several data acquisition modes: (i) single energy-windowed image that utilizes a single channel, wherein x-ray photons above the selected energy threshold (E_{th}) are counted, (ii) single energy-windowed dead-time free mode that utilizes both channels wherein one channel

FIG. 1. (a) Schematic of the CdTe detector. X-rays are incident on the continuous thin-film Pt cathode deposited on a continuous CdTe layer. Anode is patterned with hexagonal pixels and is flip-chip bonded to a CMOS ASIC with identically patterned pixel matrix. (b) Photograph of a single detector module.

acquires data for the next readout while the other channel is readout, and (iii) simultaneous operation of both channels, which based on the selected energy threshold for each channel could produce up to two energy-windowed images, one each from each of the channels. This version of the detector was equipped with a readout ASIC (PIXIE-1) that did not include circuitry to correct for charge sharing between adjacent pixels. A gigabit Ethernet interface is used for data communication.

X-rays are incident on the Pt cathode which is held at approximately -400 V . To minimize any possible polarization

effect on the charge collection efficiency, the very thin sensor is operated at low temperature (typically at -30 to $-20\text{ }^{\circ}\text{C}$). The operation at low temperature and the high applied electric field (approximately 6 kV/cm) greatly extends the time window for stable detector operation. At the expected radiation flux, the detector can be kept continuously under high-voltage and under radiation for more than 15 min. This is not a limiting factor for neurointerventional applications as the detector can be fully refreshed by rapidly recycling the applied high-voltage between acquisitions. The process for ramp-down and ramp-up of the high-voltage takes less than 2 s and the detector can be completely refreshed by switching-off the high-voltage for 4 s, for a total of 6 s. Upon x-ray interaction, electron-hole pairs are generated and drift toward the electrodes under the influence of the electric field and can diffuse laterally. An electrical signal equivalent to the electron-hole pairs is induced and is concentrated on that pixel as its dimensions are substantially smaller than the thickness of the detector (ratio of 1:10.8) resulting in “small pixel” effect.¹⁶ Indeed, the “weighting field” is high only around the pixel as the “weighting potential” drops-off by more than 90% within a distance of the same order of pixel size, resulting in a very effective small pixel effect.¹⁶ The peaking time of the on-pixel charge sensitive amplifier is 400 ns. Hence, the collection of holes, which is an order of magnitude lower in mobility, in the very weak and uniform “weighting” field does not contribute significantly to the signal.

TABLE I. Key specifications of the single-module CdTe Schottky sensor investigated in this study.

Parameter	Value
CdTe layer thickness	650 μm
Bias voltage	200–400 V
Leakage current	$\sim 5\text{ nA/cm}^2$ at 400 V ($-20\text{ }^{\circ}\text{C}$)
Pixel dead time	300 ns, or dead-time free mode
Maximum frame rate	200 frame/s
Pixel count rate (maximum)	$1 \times 10^6\text{ count/(pixel/s)}$ (dead-time corrected)
Global count rate (maximum)	$2.4 \times 10^{11}\text{ count/s}$
Active area	31 $\times$ 25 mm
Pixel matrix (hexagonal sampling)	512 $\times$ 476
Discriminators/counters per pixel	2
Counter depth	15 bits (32768 counts)

2.B. Experimental setup and imaging conditions

In this study, the detector was operated in a single-photon counting (SPC) mode (single energy window) using one of the two available channels to include x-ray photons of 10 keV or higher (i.e., $E_{th} \geq 10$ keV). The bias voltage was set to 400 V for all experiments and the detector was operated at -20°C . The x-ray system used for the experiments comprised a radiography/fluoroscopy high-frequency x-ray generator (Indico100, CPI, Inc., Canada) powering an x-ray tube (B-150 housing with A-192 tube insert, Varian Medical System, Salt Lake City, UT) with 0.6 and 1.2 mm nominal focal spot sizes and 12° anode angle. The manufacturer specified inherent filtration of the x-ray tube was 0.75 mm of Al at 75 kVp. The collimator housing included a 1.5 mm of Al precollimation filter. The detector performance was characterized at IEC RQA5 spectral conditions¹³ with the detector positioned at 100 cm from the source. This was achieved by adding 21.9 cm of type 1100 alloy of Al close to the tube port and at 70 kVp. The first half value layer (HVL) was determined from exposure measurements performed using a calibrated ionization chamber (MDH 1515, RadCal, Corp., Monrovia, CA). The empirically determined first HVL was 7.01 mm of Al. The detector incident spectrum was simulated using srs-78 software¹⁷ provided by the IPEM to match the measured first HVL and kVp. The spectrum normalized to unit area is shown in Fig. 2. From the definition of the Roentgen and the mass energy absorption coefficient of air from NIST, the estimated photon fluence per unit exposure ($\bar{q}_0/X$) was 267 photon/(mm² μR). Using a conversion coefficient of 8.76 mGy/R, the estimated photon fluence per unit air kerma ($\bar{q}_0/K$) was 30.48 photon/(mm² nGy) and is 1% higher than the IEC specified value for the RQA-5 spectrum.

2.C. Linearity and sensitivity

The linear response of the detector was measured over approximately five-orders of detector entrance exposure rate (air kerma rate) ranging from 0.15 $\mu\text{R}/\text{frame}$ (1.34 nGy/frame) to 10.4 mR/frame (91.38 nGy/frame). The lowest detector

FIG. 2. Normalized (to unit area) x-ray spectrum used in empirical studies.

entrance exposure rate investigated was an order of magnitude lower than the typical fluoroscopic exposure rate of 1.5–2.5 $\mu\text{R}/\text{frame}$ (Ref. 18) and the highest detector entrance exposure rate was an order of magnitude higher than that used for digital subtraction angiography.¹⁸ This range of air kerma rate was achieved by varying the tube current (mA) and the detector frame duration. The detector temperature was maintained at -20°C during image acquisition. The exposure-dependence of mean counts per pixel was used to assess the linearity of detector response. The mean sensitivity ($\bar{\Gamma}$) was computed as the slope from the linearity plot.

2.D. Presampled modulation transfer function

Several methods for determining the presampled MTF to characterize various imaging systems using slit or edge test devices,^{19–26} as well as comparison among various algorithms²⁷ using the edge-spread function (ESF), have been reported. The IEC recommended methodology is based on the edge test device and was adopted with modification to account for the hexagonal sampling matrix. A tungsten edge test device oriented at small angle (1° – 3°) with respect to the apothem, or at a small angle orthogonal to the apothem, was placed in contact with the detector cover. The spacing between the detector cover and the CdTe sensor is 4–5 mm. For the 0.6 mm focal spot and source to detector distance of 100 cm, the penumbra is less than 0.5%. Multiple image frames were acquired at IEC RQA-5 spectral conditions and averaged after conversion to double-precision floating point values.

In order to describe the modification that is needed to account for the hexagonal sampling matrix, the pixel coordinate system needs to be defined as the hexagonal sampling matrix can be defined in terms of quadrants, sextants, or layers. The coordinate system used in this study is shown in Fig. 3(a). Within the same row of pixels, the center-to-center pixel spacing is $\Delta x = 2a = 60 \mu\text{m}$, where the apothem, $a = 30 \mu\text{m}$. Within the same column of pixels, the center-to-center pixel spacing between adjacent rows is $\Delta y = \Delta x \sqrt{3}/2 = 51.96 \mu\text{m}$. However, pixels with odd y -coordinates are shifted from pixels with even y -coordinates by the apothem a along the x -direction.

In Fig. 3(b), the method used to generate the oversampled ESF along the y -axis, i.e., orthogonal to the apothem is described. Let us denote the pixel where the edge of the test device traverses through the pixel center as (x_0, y_0) and is oriented at a small angle (α_x) with respect to x -axis. Then, the sampling distance between the edge of the test device and the pixel center along the y -axis is dependent on whether the y -coordinate of the pixel is odd or even. For any pixel (x, y) , represented as $(x_0 + i, y_0 + j)$, the sampling distance $d_y(i, j)$ from the edge of the test device to the pixel center can be represented as

$$d_y(i, j) = \begin{cases} j \Delta y + i \Delta x \tan \alpha_x, & \text{when } j \text{ is even} \\ j \Delta y + (i + 0.5) \Delta x \tan \alpha_x, & \text{when } j \text{ is odd} \end{cases} \quad (1)$$

In Eq. (1), d_y is negative for pixels that are attenuated by the edge test device and is positive for pixels in the unattenuated

FIG. 3. (a) Illustration showing the orientation of the hexagonal pixel matrix and the pixel coordinate system. (b) Method used for computing the sampling distance between the pixel center and the edge test device, when it is oriented at a small angle α_x with respect to the apothem. (c) Method used for computing the sampling distance between the pixel center and the edge test device, when it is oriented at a small angle α_y with respect to the y -axis. (d) The angle (α_x or α_y) was determined from two subimages corresponding to odd and even y -coordinates using Hough transform, as standard implementations assume Cartesian coordinate system. The angle estimates were validated by linear regression that accounted for the hexagonal sampling matrix.

region. It is relevant to note that y_0 can be either odd or even and for each i , the sampling distance d_y is dependent on j being odd or even. The oversampled ESF is determined from n_x adjacent columns of data,²⁶ where n_x is represented in terms of the pixel coordinates and $n_x = 2 \Delta y / (\Delta x \tan \alpha_x)$. Since there is a $\Delta x/2$ offset between even and odd j for each i , the oversampled ESF is constructed from $2n_x - 1$ edge profiles.

Figure 3(c) illustrates the method used to determine the oversampled ESF along the x -axis, i.e., parallel to the apothem. From Fig. 3(c), it could be inferred that for any pixel $(x_0 + i, y_0 + j)$, the sampling distance $d_x(i, j)$ from the edge of the test device to the pixel center can be represented as

$$d_x(i, j) = \begin{cases} i \Delta x + j \Delta y \tan \alpha_y, & \text{when } j \text{ is even} \\ (i + 0.5) \Delta x + j \Delta y \tan \alpha_y, & \text{when } j \text{ is odd} \end{cases} \quad (2)$$

In Eq. (2), d_x is negative for pixels that are attenuated by the edge test device and is positive for pixels in the unattenuated region. The oversampled ESF is determined from N_y rows of data,²⁶ where $n_y = \Delta x / (\Delta y \tan \alpha_y)$.

In order to compute the number of columns (rows) of data needed to generate the oversampled ESF, the angle α_x (α_y) need to be determined. We used both the Hough transform and the linear regression-based methods^{26,27} for determining α_x and α_y . Standard implementations of Hough transform in computing software such as MATLAB® (The MathWorks, Inc.,

Natick, MA) or IDL® (Exelis Visual Information Solutions, Inc., Boulder, CO) are based on Cartesian coordinate system and some may also assume that the sampling distance, $\Delta x = \Delta y$. Hence, the images with the edge test device were decomposed to two subimages corresponding to even and odd j coordinates as shown in Fig. 3(d). For each subimage, the spacing between adjacent pixels is Δx and $2\Delta y$ along the two orthogonal axes. From each subimage, the angle (α_x or α_y) was determined and they were identical. For the image acquired with the edge test device oriented at a small angle with respect to the apothem, $\alpha_x = 1.99^\circ$ using the Hough transform method, and resulted in $n_x = 50$. The oversampled ESF, $\text{ESF}(d_y)$ was constructed from $2n_x - 1 = 99$ edge profiles based on Eq. (1). Linear regression based method resulted in identical number of edge profiles. For the image acquired with the edge test device oriented at a small angle along the direction orthogonal to the apothem, $\alpha_y = 1.7^\circ$ using the Hough transform method and resulted in $n_y = 40$. Linear regression yielded an identical value of $n_y = 40$. Prior to numerical differentiation to generate the line spread functions (LSFs), the oversampled ESFs were regularized with a third order polynomial function.²⁷ The MTFs along the two orthogonal axes were obtained by Fourier transform of the corresponding LSFs. Thus, the method is analogous to Algorithm A in Samei *et al.*²⁷ with the modification to account for hexagonal sampling matrix. Algorithm A (Ref. 27)

corresponds to determining the angle of the edge test device using linear regression, computing the number of consecutive edge profiles needed for the oversampled ESF from the slope of the regression line, numerical differentiation of the ESF to provide the LSF, and Fourier transforming the LSF to provide the MTF. The method was repeated for multiple nonoverlapping segments along the edge test device to provide the mean and the standard deviation (SD) of the MTF.

2.E. Resampling to square pixels

Current image displays use Cartesian coordinate system display data with equal pixel spacing along the two orthogonal directions. Hence, images acquired with the investigated detector require transformation to square pixels and Cartesian coordinate system for distortion-free display. The optimal square pixel size, p^* , was determined by minimizing the root-mean-squared-error (RMSE) between the hexagonal pixel aperture function, $T^H(u, v)$, and the square pixel aperture function, $T^S(u, v)$,

$$p^* = \arg \min_a \sqrt{\frac{1}{n_u n_v} \sum_u \sum_v [T^H(u, v) - T^S(u, v)]^2}. \quad (3)$$

In Eq. (3), n_u and n_v are the number of spatial frequency elements along the orthogonal axes, $T^H(u, v)$ is the hexagonal pixel aperture function reported by Barnard and Boreman,²⁸ and $T^S(u, v) = |\text{sinc}(au) \text{sinc}(av)|$ is the square pixel aperture function. Equation (3) was evaluated over the frequency ranges $u \in [-u_{\text{Nyq}}^H, u_{\text{Nyq}}^H]$ and $v \in [-v_{\text{Nyq}}^H, v_{\text{Nyq}}^H]$. Upon determination of p^* , the acquired images with the hexagonal SPC detector were resampled to square pixels using (tri)linear interpolation of the vertices of the triangle that correspond to the center of hexagonal pixels bounding each square pixel location. Referring to Fig. 4, for a generic pixel of the square matrix with coordinates (x, y) , the three closest hexagonal pixels are labeled #0, #1, and #2. The coordinates of these hexagonal

pixels are (x_0, y_0) , (x_1, y_1) , and (x_2, y_2) , respectively, and are the vertices of the triangle bounding (x, y) . The signal (counts) in these hexagonal pixels is represented as n_0 , n_1 , and n_2 , respectively. Then the counts corresponding to pixel coordinate (x, y) represented as n_{xy} are determined through linear interpolation from the equation of the plane passing through points (x_0, y_0, n_0) , (x_1, y_1, n_1) , and (x_2, y_2, n_2) ,

$$n_{xy} = n_0 + a(x - x_0) + b(y - y_0),$$

where

$$a = \frac{n_1 - n_0}{\Delta x},$$

$$b = \left(\frac{1}{\Delta y} \right) \left[n_2 - \left(\frac{n_0 + n_1}{2} \right) \right]. \quad (4)$$

Similar equations can be written for the five remaining triangles surrounding the pixel #0 shown in Fig. 4.

2.F. MTF of resampled images

Upon resampling the edge test device images to square pixels of pitch p^* , the presampled MTF was computed in a manner similar to that described in Sec. 2.D with determination of the sampling distances d_x and d_y for Cartesian coordinate system. The approach is identical to Algorithm A in Samei et al.,²⁷ with the exception of determining the angle of the edge test device using the Hough transform.²¹ The method was repeated for multiple nonoverlapping segments along the edge test device to provide the mean and the standard deviation of the MTF.

2.G. Qualitative evaluation

A diverging bar pattern with maximum resolution of 20 line-pairs (lp)/mm was imaged under RQA-5 conditions for visual analysis of the limiting resolution of the detector. Images of a finger phantom and the wrist of a hand phantom were acquired for qualitative evaluation. The ability to visualize the struts of a nickel–titanium (nitinol) stent is challenging in clinical neurointerventional C-arm systems. Hence, we imaged a nitinol stent (Enterprise™, Cordis Neurovascular) in a patient-specific neurovascular replica of the middle cerebral artery with iodinated contrast media. In order to provide x-ray attenuation conditions relevant to neurovascular applications, 15.7 cm of polymethyl methacrylate (PMMA) was added to the beam path. All images were resampled to square pixels of pitch, p^* .

3. RESULTS

3.A. Linearity and sensitivity

Figure 5 shows the linear response of the detector ($E_{\text{th}} \geq 10$ keV) over approximately five orders of magnitude of incident air kerma spanning 1 nGy to 100 μGy (0.15 μR to 10.4 mR of exposure). Linear regression (solid line) shows that the detector exhibits excellent linear response with $r > 0.9999$. The mean sensitivity of the detector, determined as the slope of the linear response curve, was $\bar{\Gamma} = 0.16335$ count/(pixel nGy).

FIG. 4. Illustration of the interpolation procedure for resampling to square pixel matrix. For a generic square pixel with coordinates (x, y) , the counts are determined by linear interpolation from three neighboring hexagonal pixels labeled #0, #1, and #2.

FIG. 5. Linear response characteristics of the detector ($E_{th} \geq 10$ keV). Sensitivity was calculated as slope of the linear response curve.

As noted earlier, the ASIC used in this detector version (PIXIE-1) was not equipped with charge-sharing compensation logic, and hence, the aforementioned sensitivity is inclusive of multiple counts due to charge sharing. It also includes multiple counts due to K-fluorescence emission and reabsorption within the CdTe as well as x-ray scattering interactions, predominantly Compton scattering, provided the energy deposited is greater the specified threshold of 10 keV.

3.B. MTF of hexagonal pixel array

Figure 6(a) shows the oversampled ESF(d_y) and with the fit (solid line) for the edge test device oriented at a small angle with respect to the apothem. Figure 6(b) shows the oversampled ESF(d_x) and the fit (solid line) for the edge test device oriented at a small angle along the direction perpendicular to the apothem. The oversampled ESF(d_y) and ESF(d_x) were constructed with 99 and 40 edge profiles, respectively.

Figures 7(a) and 7(b) show the presampled MTF determined from ESF(d_y) and ESF(d_x), respectively. Since the pixel spacing along apothem (x -axis) and orthogonal to

the apothem (y -axis) is 60 and 51.96 μm , the corresponding Nyquist limits are $u_{Nyq}^H = 8.33$ cycles/mm and $v_{Nyq}^H = 9.62$ cycles/mm. The error bars in each plot correspond to the SD from ten realizations of the oversampled ESFs obtained from nonoverlapping segments along the edge. In each plot, the pixel aperture function for the corresponding axis is plotted based on the work of Barnard and Boreman.²⁸ The $\text{MTF}(0, v_{Nyq}^H)$ and the $\text{MTF}(u_{Nyq}^H, 0)$ were 0.228 ± 0.021 and 0.397 ± 0.028 , respectively. The limiting resolution defined was 10% modulation factor occurred at $v = 12$ cycles/mm and $u = 13.3$ cycles/mm.

3.C. Resampling to square pixels

The optimal square pixel size was determined by minimizing the RMSE between the square and hexagonal 2-D aperture functions and is shown in Fig. 8. The optimal square pixel size was $p^* = 54$ μm . Evaluation of the total, and the maxima, of the absolute difference between square and hexagonal pixel aperture functions also yielded minima at identical $p^* = 54$ μm . Figure 9 shows the importance of resampling to square pixels. In Fig. 9(a), the grid lines of the phantom (CDMAM, St. Radboud, Nijmegen) are distorted with stretching along the y -axis. This is corrected for after resampling to square pixels [Fig. 9(b)]. Hereon, all images shown are after resampling to 54 μm square pixels.

After resampling the edge test device images to 54 μm square pixels using (tri)linear interpolation, the presampled MTF along the two orthogonal directions were determined (Fig. 10). In each plot, the presampled MTF of the hexagonal pixel array for the corresponding axis is also shown so that changes due to resampling can be readily visualized. The presampled MTF at Nyquist frequency of 9.26 cycles/mm corresponding to 54 μm square pixel was 0.29 and 0.24 along the u and v axes, respectively. The 10% MTF occurred at 12.3 and 12 cycles/mm along the u and v axes, respectively.

3.D. Qualitative imaging

Figure 11(a) shows the diverging bar pattern image acquired under IEC RQA-5 conditions and after resampling

FIG. 6. For the edge test device oriented at a small angle with respect to the apothem, the oversampled ESF(d_y) constructed from 99 edge profiles is shown in (a). The oversampled ESF(d_x) constructed from 40 edge profiles for the orthogonal direction is shown in (b). In each plot, the fit is shown as a solid line.

FIG. 7. The presampled MTF computed from (a) $\text{ESF}(d_y)$ and (b) $\text{ESF}(d_x)$. The error bars in each plot correspond to the SD from ten realizations of the oversampled ESFs obtained from nonoverlapping segments along the edge. The Nyquist limits are $v_{\text{Nyq}}^H = 9.62$ and $u_{\text{Nyq}}^H = 8.33$ in (a) and (b), respectively. In each plot, the pixel aperture function for the corresponding axis is plotted.

to $54 \mu\text{m}$ square pixels. In Fig. 11(b), a subregion of the diverging bar pattern image is shown, which indicates the ability to resolve between 10 and 12 lp/mm (arrow in figure) that is consistent with the limiting resolution (10% MTF) shown in Fig. 10. In Fig. 11(b), the bar pattern appears jagged as it is shown at $2\times$ zoom without interpolation. Figure 12 shows images of the (a) finger phantom and (b) wrist of the hand phantom. The technique factors used for acquisition are provided in figure caption and are representative of clinical practice. Figure 13 shows the images of the nitinol-stented neurovascular replica acquired with 15.7 cm of PMMA in the beam path. For the imaged stent, the strut width is $78 \mu\text{m}$ and the average strut thickness is $42 \mu\text{m}$ and contains radiopaque platinum markers at the proximal and distal aspects of the stent to guide treatment since the struts are not visualized with current C-arm systems.²⁹ Technique factors used for acquisition and the receptor entrance exposure (REE) are provided in the figure caption. For the single frame image [Fig. 13(b)], the REE of $30.5 \mu\text{R/frame}$ is approximately 40%

lower than the REE range of $50\text{--}100 \mu\text{R/frame}$ suggested for digital angiography.¹⁸ The REE of $426.6 \mu\text{R/frame}$ for the 14-frame average image [Fig. 13(c)] is approximately 15% lower than the $500\text{--}1000 \mu\text{R/image}$ suggested for digital subtraction angiography.¹⁸

4. DISCUSSION

The version of the ASIC (PIXIE-I) investigated in this study was not equipped with charge-sharing compensation circuit/logic. Two subsequent versions of the readout ASIC (PIXIE-II and PIXIE-III) have been fabricated, but were unavailable for this study. PIXIE-II improves on PIXIE-I by evaluating the charges at neighboring pixels surrounding each hexagonal pixel and allocating the count to the pixel with the highest charge. This improves spatial resolution but not the energy resolution. PIXIE-III improves on PIXIE-II by summing the charge in each pixel's neighborhood and assigning the count to the pixel with the highest charge and at the energy-bin equivalent to the summed charge. This is expected to improve both energy and spatial resolution. Evaluation³⁰ of PIXIE-III showed energy resolution (FWHM) of 3.9 keV for Am-241 (59.6 keV).

In the absence of charge-sharing compensation logic/circuitry, the effect of charge-sharing is discussed in the context of single-photon counting mode with one energy bin or channel, and during dual-energy imaging with two channels. For the hexagonal pixel matrix, excluding the pixels at the periphery of the detector, each pixel has six neighboring pixels. For simplicity, we discuss the effect of charge-sharing between two adjacent pixels. Let E_{in}^i represent the energy of the incident (and attenuated) x-ray photon on pixel i sampled from an x-ray spectrum with energy range $[E_{\text{min}}, E_{\text{max}}]$, and E_{sh}^j represent the energy corresponding to the charge shared with its neighboring pixel j . Since, $E_{\text{sh}}^j = 0$ corresponds to the absence of charge sharing, only $E_{\text{sh}}^j > 0$ is considered. In the single-photon counting mode, let E_{th} represent the energy-threshold above which the x-ray photons are counted and are selected such that $E_{\text{min}} > E_{\text{th}} > E_{\text{noise}}$, where E_{noise} corresponds to the energy-equivalent of the total system noise.

FIG. 8. The optimal square pixel size for resampling was determined from the minima of the RMSE between the square and hexagonal pixel aperture functions and yielded $p^* = 54 \mu\text{m}$.

FIG. 9. (a) The gridlines of a CDMAM phantom are stretched along the y -axis when the data from the hexagonal pixel array detector are displayed. (b) This is corrected for after resampling to square pixels.

Charge-injection based measurement for one version of this detector (PIXIE-III) showed $E_{\text{noise}} = 1.6$ keV.³¹ When $E_{\text{in}}^i \leq 2E_{\text{th}}$, then a count will be recorded in pixel i alone (true count). When $E_{\text{in}}^i > 2E_{\text{th}}$, then a count will be recorded in pixel i (true count) and a count will be recorded in pixel j (false count). Selecting $E_{\text{th}} = E_{\text{min}}/2$ could reduce the likelihood of false counts without degrading sensitivity. Typically, $E_{\text{min}} > 20$ keV during neurointerventional imaging and hence selection of $E_{\text{th}} = (E_{\text{min}}/2) > E_{\text{noise}}$ can be readily achieved. Thus charge-sharing in single-photon counting mode results in an apparent increase in sensitivity due to false counts and degradation of spatial resolution due to correlations between neighboring pixels.

In dual-energy imaging with two channels, let E_{low} and E_{high} represent the low- and high-energy thresholds, where $E_{\text{min}} > E_{\text{low}} > E_{\text{noise}}$ and $E_{\text{high}} > E_{\text{low}}$. Since $E_{\text{low}} > E_{\text{noise}}$, the low-energy bin will not contain counts due to electronic noise. The low- and high-energy bins correspond to $(E_{\text{low}}, E_{\text{high}}]$ and $> E_{\text{high}}$, respectively. When $E_{\text{high}} > E_{\text{in}}^i > E_{\text{low}}$, charge-sharing contributes to (1) a count in the low-energy bin in pixel i alone (true count), when $E_{\text{sh}}^j < E_{\text{low}}$, (2) a count in the low-energy bin in pixel i (true count) and a count in low-energy bin in pixel j (false count), when $E_{\text{sh}}^j > E_{\text{low}}$. It can be readily inferred

that in order for a false count to occur, $E_{\text{high}} > E_{\text{in}}^i > 2E_{\text{low}}$. When $E_{\text{in}}^i > E_{\text{high}}$, charge-sharing contributes to (1) a count in high-energy bin in pixel i alone (true count), when $E_{\text{sh}}^j < E_{\text{low}}$; (2) a count in high-energy bin in pixel i (true count) and a count in low-energy bin in pixel j (false count), when $E_{\text{high}} > E_{\text{sh}}^j > E_{\text{low}}$; and (3) a count in high-energy bin in pixel i (true count) and a count in high-energy bin in pixel j (false count), when $E_{\text{sh}}^j > E_{\text{high}}$. In order for a true count and a false count to occur in the high-energy bin, $E_{\text{in}}^i > 2E_{\text{high}}$. Although the selection of energy thresholds is predicated upon the materials of interest, choosing $E_{\text{low}} = E_{\text{min}}/2$, when possible, reduces the likelihood of false counts in the low-energy bin of pixel j , when $E_{\text{high}} > E_{\text{in}}^i > E_{\text{low}}$. Thus charge sharing in dual-energy mode results in an apparent increase in sensitivity due to false counts, degradation of spatial resolution, and reduction in energy resolution.

Regarding the sensitivity, the mean estimate of $\bar{\Gamma} = 0.16335$ count/(pixel nGy) is inclusive of multiple (“false” or “double”) counts due to charge sharing arising from photoelectron range and charge diffusion, x-ray scattering with CdTe, and K-fluorescence emission and reabsorption within CdTe. For the IEC RQA-5 spectrum and using the linear attenuation coefficient of CdTe from XCOM database,³² the

FIG. 10. Presampled MTF after resampling to square pixels showing minimal change from that for hexagonal pixels.

FIG. 11. (a) Image of a diverging bar pattern acquired under IEC RQA-5 spectral conditions and after resampling to $54\ \mu\text{m}$ square pixels ($E_{\text{th}} \geq 10\ \text{keV}$). (b) Subregion of the image ($2\times$ zoom without interpolation) shows the ability to resolve between 10 and 12 lp/mm (arrow).

estimated quantum efficiency ($\bar{g}_1$) of 0.65 mm CdTe is 95.48%, under the assumption of orthogonal x-ray beam incidence. Since $\bar{q}_0/K = 30.48\ \text{photon}/(\text{mm}^2\ \text{nGy})$, and defining the mean count sensitivity as measured counts per attenuated x-ray photon, $\bar{\Gamma}_c = \bar{\Gamma}/(\bar{q}_0/K) \times A_{\text{pix}} \times \bar{g}_1$ results in $\bar{\Gamma}_c = 1.8$. Since each attenuated x-ray photon should result in utmost one count, the estimated false count fraction is 80%. A prior study on charge-sharing with a similar detector using x-ray fluorescence lines from elemental media and using the SYRMEP beamline at Elettra synchrotron laboratory (Trieste, Italy) indicated that the charge-sharing fraction had a linear energy-dependence in the 3–10 keV range and a plateauing to $\sim 70\%$ in the 10–26 keV range.³³ Since these charge sharing measurements were conducted at energies less than the K-absorption edge of Cd (26.7 keV) and Te (31.8 keV), the observation of slightly higher false count fraction through the sensitivity analysis is reasonably consistent.

The effective atomic number ($Z = 50$) and the density ($\rho = 5.85\ \text{g}/\text{cm}^3$) of CdTe are substantially higher than that of Si ($Z = 14$, $\rho = 2.33\ \text{g}/\text{cm}^3$). Hence, a much thicker Si layer is needed to achieve the same quantum efficiency as CdTe. For the IEC RQA5 spectrum, approximately 3.5 cm

thick Si layer ($>50\times$ increase in thickness) would be needed to provide the same quantum efficiency (96%) as a 0.065 cm thick CdTe layer. The increased thickness of Si could contribute to worsening of MTF due to charge diffusion and in particular due to oblique x-ray incidence. The advantages of CdTe over CZT are reduced leakage current,³⁴ improved charge transport properties,³⁴ and improved energy resolution.³⁵

Currently, detectors with hexagonal pixel array are in clinical use for digital mammography (Aspire Cristalle, Fujifilm Medical Systems USA, Inc., Stamford, CT). A prototype version of this amorphous selenium (a-Se) based energy-integrating detector was reported³⁶ to have a pixel pitch of 75 and $62.5\ \mu\text{m}$ along the orthogonal directions. The study³⁶ characterizing this detector reported after resampling to square pixel matrix of $50\ \mu\text{m}$ pitch. The methodology described in this work would allow for determining the presampled MTF in the native pixel format so that the impact of resampling can be quantified. That study³⁶ also compared the detective quantum efficiency (DQE) after resampling to square pixel matrix of $50\ \mu\text{m}$ pitch. DQE estimation is the subject of ongoing investigation and will be reported in future. As noted earlier, the advantage of hexagonal pixel matrix is that it would

FIG. 12. (a) Image of a finger phantom (50 kVp, 1.94 mm of Al HVL, 0.8 mAs, ESE: 2.5 mR, $E_{\text{th}} \geq 10\ \text{keV}$). (b) Image of the wrist of a hand phantom (70 kVp, 2.69 mm of Al HVL, 0.8 mAs, ESE: 5.6 mR, $E_{\text{th}} \geq 15\ \text{keV}$). Both images are after resampling to $54\ \mu\text{m}$ square pixels. (ESE—entrance skin exposure).

FIG. 13. (a) Photograph of the neurovascular replica with nitinol stent. (b) Single frame from a digital cine sequence (70 kVp, 14.7 cm PMMA filtration, 5.25 mm Al HVL post-PMMA, 0.4 mAs/frame, REE: 30.5 μ R/frame, $E_{th} \geq 10$ keV). (c) Average of 14 frames (REE: 426.6 μ R). [REE—receptor (detector) entrance exposure.] For the imaged stent, the strut width is 78 μ m and the average strut thickness is 42 μ m. The struts of the stent and the kink [marked in (c)] are easily visualized.

require lesser number of pixels (by 13.4%) to achieve the same sampling density as square pixel matrix. With regards to the resampling procedure, there are alternative interpolation schemes and approaches.³⁷ However, the good agreement in presampled MTF prior to and after resampling (Fig. 10) suggests that the benefits of these alternative resampling schemes may be limited and need to be investigated.

Most clinical neurointerventional systems utilize a large field-of-view CsI:Tl coupled a-Si detector with pixel pitch in the range of 143–200 μ m. For these systems, the reported limiting resolution (10% MTF) is approximately 3.5 cycles/mm.^{5,6} Evaluations of direct-conversion a-Se based detector with 150 μ m pitch have shown a limiting resolution of approximately 6 cycles/mm.^{6,9} Characterization of a MAF employing a 300 μ m thick CsI:Tl scintillator coupled to a CCD with 35 μ m pitch via a light image intensifier and fiberoptic taper showed a limiting resolution of 4 cycles/mm.¹¹ The observed 10% MTF of ~ 12 cycles/mm with the investigated detector in the SPC mode compares favorably with the added advantage of effectively eliminating electronic noise through appropriate selection of the energy threshold. Further improvement in MTF is possible with appropriate selection of energy threshold, when there is prior knowledge of the x-ray spectrum. While magnification, which is unavoidable during clinical imaging would degrade the spatial resolution due to focal spot blur, current generation of neurointerventional systems has small focal spot sizes of 0.3–0.4 mm, compared to the 0.6 mm focal spot size used in the study.

The hexagonal pixel area of the investigated detector is equivalent to a square pixel with pitch of 55.8 μ m. This is similar to the 55 μ m square pixel pitch (Ref. 38) reported for Medipix3. In terms of the aperture function, the investigated detector is equivalent to 54 μ m square pixel (Fig. 8). Importantly, each module of the investigated detector provides a larger FOV (31 $\times$ 25 mm) than the 14 $\times$ 14 mm FOV of the Medipix3 detector. In addition to Medipix3, several CdTe-based single-photon counting detectors with sub-mm pixel pitch such as Pilatus 3 (Dectris, Ltd., Baden, Switzerland) with 172 μ m pixel pitch, XPAD3 with 130 μ m pixel pitch,³⁹ and DANA and HILDA with 500 μ m pixel pitch (Ref. 40) have been reported. Considering that image-guided interventions require the ability to visualize 67–78 μ m wide struts of nitinol stents,²⁹ there is a need for a high-resolution imaging platform. The investigated detector, in addition to high-resolution imaging, provides for dual-energy imaging that obviates the need for precontrast and postcontrast images during digital subtraction angiography. Quantification of the dependence of the presampled MTF on the energy threshold (E_{th}), noise properties and consequently the DQE, and the evaluation of the dual-energy capabilities are subjects of ongoing investigations and will be reported in future.

5. CONCLUSIONS

In summary, the study investigated the spatial resolution properties of a thin-layer (0.65 mm) CdTe-based photon-counting hexagonal pixel array detector operating in a single-photon counting mode. The study also addressed the methodology for quantifying the presampled MTF in hexagonal pixel arrays in its native format. The fine sampling pitch of 60 and 51.96 μ m along the orthogonal directions resulted in limiting resolution (10% MTF) of 12 cycles/mm or greater. Since hexagonal pixel arrays require to be resampled to square pixel matrix to match current image displays, the optimal square pixel pitch was determined to be 54 μ m. After resampling

to square pixels, the presampled MTF was similar to that of the native pixel format and provided a limiting resolution (10% MTF) of approximately 12 cycles/mm. Visual analysis of a diverging test pattern confirmed the high-resolution capabilities of the detector. The detector also showed the ability to resolve the struts of a nitinol-stent at clinically relevant conditions and at potentially lower radiation dose. Considering that the detector is capable of concurrently acquiring dual-energy images at frame rates of up to 200 image/s, it could allow for artifact-free subtraction angiography and basis material decomposition. The investigated high-resolution photon-counting detector with energy-resolving capability can be of importance for several image-guided interventional applications as well as for pediatric applications.

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^{a)} Author to whom correspondence should be addressed. Electronic mail: andrew.karellas@umassmed.edu; Telephone: (508) 856-1238; Fax: (508) 856-6363.

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