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Progress in electron-multiplying CCD (EMCCD) based, high-resolution, high-sensitivity x-ray detector for fluoroscopy and radiography

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Abstract

A new high-resolution, high-sensitivity, low-noise x-ray detector based on EMCCDs has been developed. The EMCCD detector module consists of a 1kx1k, 8 μ m pixel EMCCD camera coupled to a CsI(Tl) scintillating phosphor via a fiber optic taper (FOT). Multiple modules can be used to provide the desired field-of-view (FOV). The detector is capable of acquisitions over 30fps. The EMCCD's variable gain of up to 2000x for the pixel signal enables high sensitivity for fluoroscopic applications. With a 3:1 FOT, the detector can operate with a 144 μ m effective pixel size, comparable to current flat-panel detectors. Higher resolutions of 96 and 48 μ m pixel size can also be achieved with various binning modes. The detector MTFs and DQEs were calculated using a linear-systems analysis. The zero frequency DQE was calculated to be 59% at 74 kVp. The DQE for the 144 μ m pixel size was shown to exhibit quantum-noise limited behavior down to $\sim 0.1\mu$ R using a conservative 30x gain. At this low exposure, gains above 30x showed limited improvements in DQE suggesting such increased gains may not be necessary. For operation down to 48 μ m pixel sizes, the detector instrumentation noise equivalent exposure (INEE), defined as the exposure where the instrumentation noise equals the quantum-noise, was $< 0.1\mu$ R for a 20x gain. This new technology may provide improvements over current flat-panel detectors for applications such as fluoroscopy and angiography requiring high frame rates, resolution, dynamic range and sensitivity while maintaining essentially no lag and very low INEE. Initial images from a prototype detector are also presented.

Keywords

DX; DCT; METR; SIM; SYS

1. INTRODUCTION

Current demands on imaging devices require detectors to provide high spatial resolution, high sensitivity, low noise, wide dynamic range, negligible lag and high frame rates. Previous designs based on CCD imagers have indicated promise to fulfill these requisites, whereas flat-panel detectors are currently limited to $\sim 150\mu$ m pixel size and have noise issues at fluoroscopic exposures. To achieve the high sensitivity, light amplifiers have been used in

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the past to amplify the signal appropriately [1]. However, the new EMCCD detector [2] employs a charge multiplication register consisting of 400 multiplication elements prior to the A/D converter, providing the necessary system gain (up to 2000x) required for fluoroscopic exposures. EMCCD technology was first introduced in 2001 and has been met with much success, particularly in spectroscopy and microscopy, and is capable of detecting single photon events. Larger EMCCD sensors (currently as large as 13.3 mm) are being developed and used for night-vision applications. To achieve a larger FOV, we propose to design an array of modules in conjunction with FOTs of magnification of 3:1 or larger [2].

Models of the linear-system analytical quantities, such as MTF and DQE as functions of spatial frequency are provided as a means for objectively evaluating the new EMCCD detector system performance. Parallel cascade theory [4,5], which incorporates parallel branching within individual cascade steps, has been studied extensively and has been used to model many different types of detectors [3,7,9] and has shown good agreement with experimental results. The instrumentation noise equivalent exposure (INEE) [2,21], defined as the detector exposure at which the instrumentation noise and the quantum noise are equivalent, is also calculated for further performance measure.

2. METHODS AND MATERIALS

2.1 EMCCD Detector Module

The prototype EMCCD detector (Photonic Science LTD, East Sussex, UK) uses an Impactron sensor, model TC285SPD (Texas Instruments, Dallas, TX). EMCCD technology has been extensively reported on for use in other applications [22–29]. The Impactron sensor consists of 1004x1002, 8 μm pixels and is capable of imaging at over 30 frames per second (fps). The EMCCD sensor operates in a similar fashion as a standard CCD, with the addition of a unique multiplication register prior to the readout of the signal, which effectively reduces the readout noise to negligible levels at high gains. A variable gain of over 1000x is achievable when necessary, under fluoroscopic conditions in which the number of incident x-ray photons is limited. A Peltier cooler has been used to minimize the dark current, which is an important consideration because thermal electrons are propagated, along with the signal, through the gain register.

The EMCCD detector (Fig. 1) was designed to allow for maximum flexibility. Binning modes from 1x1 to 8x8 allow the EMCCD pixel size to be varied from 8 μm to 64 μm . A 2 cm 1:1 fiber optic was coupled directly to the sensor and protrudes slightly from the camera head casing, allowing for coupling of a fiber optic taper (FOT). Various taper ratios can then be studied. Different scintillating phosphors can then be coupled to the FOT. Previous experiments suggest a 250 μm CsI(Tl) phosphor would be optimal in terms of maximizing the area under the DQE curve [3]. It is proposed that initially a 4:1 taper be used which would provide a field of view (FOV) of 3.2 cm, which is sufficient for initial testing purposes. A modular array of such detectors can then be used in combination with a larger FOT to increase the FOV to the desired size [2]. A larger EMCCD sensor (1024x1024, 13 μm pixels) could also be used to increase the FOV even further.

2.2 Calculation of the detector MTF, DQE and INEE

Parallel cascade theory [4,5] was used to derive the frequency dependant DQE for the EMCCD detector and consisted of 13 stages (Fig. 2). At each stage, the quanta undergo one of three processes: amplification, blurring or addition of noise. Amplification processes may involve binary selection or conversion of quanta from one form to another. The signal at the output stage of an amplification process is given by:

$$\Phi_{out} = \bar{m} \Phi_{in}. \quad (1)$$

The NPS at the output stage of an amplification process is given by [6]:

$$NPS_{out}(f) = \bar{m}^2 NPS_{in}(f) + \sigma_m^2 \Phi_{in}, \quad (2)$$

where $\bar{m}$ is the average gain with a variance of σ_m^2 and Φ_{in} is the number of input quanta per pixel. It follows that consecutive binomial processes ($\bar{m} < 1$) can be combined into one binomial process with a probability equal to that of the product of the probabilities of each individual stage [7].

Blurring processes may be either stochastic, which is caused by the spatial spreading (scattering) of stochastic quanta, or deterministic (no addition of spatial noise) [7]. The signal at the output stage of a blurring process is given by:

$$\Phi_{out}(f) = MTF_{blur}(f) \Phi_{in}(f). \quad (3)$$

The output NPS after a stochastic blurring stage is given by [6]:

$$NPS_{out}(f) = [NPS_{in}(f) - \Phi_{in}] |MTF_{blur}(f)|^2 + \Phi_{in}; \quad (4)$$

and for a deterministic blurring stage:

$$NPS_{out}(f) = NPS_{in}(f) |MTF_{blur}(f)|^2. \quad (5)$$

Using equations 1–5 at each appropriate stage enables the calculation of the frequency dependent signal and noise at the output of the detector, denoted by $\Phi_{13}(f)$ and $S_{13}(f)$ respectively. Several authors [3,7,9] have performed this calculation on similar detectors, so the details have been omitted. The resulting DQE was obtained using the following formula (variables are described above in Table 1):

$$DQE(f) = \frac{\Phi_{13}^2(f)}{\Phi_0 S_{13}(f)} = \frac{\Phi_4^2(f) \kappa_1^2 \kappa_2^2 \tau^2 Q^2 G^2 T_8^2(f) T_{11}^2(f)}{\Phi_0 [\{\kappa_2^2 \kappa_1^2 \tau^2 Q^2 T_8^2(f) [S_4(f) - \Phi_4] + \kappa_2 \kappa_1 \tau Q \Phi_4 + \sigma_t^2\} G^2 T_{11}^2(f) + G \Phi_{11} + \sigma_r^2]}, \quad (6)$$

where $\Phi_4(f)$ is the calculated signal at the output of the phosphor (stage 4):

$$\Phi_4(f) = \eta \Phi_0 T_4(f) \left[\{1 - \xi \omega\} g_A + \xi \omega \sum_i I_{\alpha i} w_i \{g_{bi} + f_{ki} g_{ci} T_{Ki}(f)\} \right], \quad (7)$$

$\Phi_{11} = \Phi_{13}/G$ and $S_4(f)$ is the calculated NPS at the output of the phosphor (stage 4):

$$S_4(f) = \Phi_4 + \eta \Phi_0 T_4^2(f) \left[\{1 - \xi \omega\} g_A^2 + \xi \omega \sum_i I_{\alpha i} w_i \{g_{Bi}^2 + f_{ki} g_{Ci}^2 + 2f_{ki} g_{Bi} g_{Ci} T_{Ki}(f)\} \right]. \quad (8)$$

The various MTF components for the EMCCD detector system were obtained as follows. Sinc functions were used to approximate the MTF of the following: two fiber optic plates (FOP) with 3 μm fiber diameter; the fiber optic taper (FOT) with 6 μm fiber diameter; and the finite pixel size associated with charge collection and sampling in the EMCCD chip with 25–150 μm size (using a 3:1 taper and appropriate binning). The MTF associated with the spread of light in the 350 μm CsI(Tl) was obtained from experimental measurements [10] previously obtained using a similar phosphor as the one proposed for the EMCCD detector. The MTF associated with K-fluorescence re-absorption at a remote location has been calculated as described by Que et al [11].

Spectrum generating software [31] was used to determine the x ray photon spectrum of a 74 kVp x-ray beam with 20.8 mm Al filtration. This spectrum was selected based on the IEC 62220-1 RQA 5 standard with a 7.1 mm Al HVL. The photon fluence was calculated using the average energy of this spectrum [12], which was ~54 keV. The quantum detection efficiency of the 350 μm CsI(Tl) was obtained by using the relevant energy dependent mass attenuation coefficients [13]. Conversion gains of the CsI phosphor were calculated assuming 1 light photon is created for every 17 eV [14], using the appropriate absorbed energy, which is dependent on the x-ray spectral component and on the branch A, B, or C taken by the x-ray photon (stage 2 of Fig. 1). K-fluorescence was taken into account using the methods established by Chan and others [3, 15, 16, 17]. Energy dependent transmission efficiencies of the two FOPs and the FOT (assuming a taper size of 3:1) were obtained from manufacturer data sheets (Incom, Inc. Charlton, MD). The overall transmission efficiency was then calculated by integrating over the CsI output spectrum. Coupling losses were also taken into account. The average quantum efficiency, dark current and read noise of the EMCCD detector were determined using specifications from the chip manufacturer (Texas Instruments Inc., Dallas TX), camera manufacturer from whom the system has been purchased (Photonic Science LTD., East Sussex UK) and the CsI output spectrum. EMCCD gain was varied between 1x and 1000x. The calculated gains and the physical constants used in the parallel cascade model are shown in Table 1.

The INEE can be determined experimentally by fitting the $(S/N_{\text{exp}})^2$ to the equivalent form [2]:

$$(S/N_{\text{exp}})^2 = AE / (1 + B/E) \quad (9)$$

where S is the measured signal (average pixel intensity), N_{exp} is the measured total noise (σ_S), E is the detector entrance exposure, and A and B are fit parameters representing a proportionality constant and the INEE respectively [2]. The theoretical form of equation (9) follows, using the notation developed above in equations 6–8:

$$\frac{\Phi_{13}^2}{\int S_{13}(f)df} = \frac{AE}{1+B/E} \quad (10)$$

where Φ_{13} is the calculated output signal, and $\int S_{12}(f)df$ is the calculated variance of the output signal [30].

2.3 Evaluation Objects

Several images were taken of various ‘test’ objects to help determine detector health, which included a mammographic line pair phantom, an operational amplifier, and an asymmetric stent [18]. These test images were taken prior to the installation of the FOT. Two different phosphors were used: $\sim 5 \text{ mg/cm}^2$ of $\text{Gd}_2\text{O}_2\text{S(Tb)}$ and $\sim 140 \text{ mg/cm}^2$ of CsI(Tl) . The detector was operated in 2x2 binning mode (16 μm pixels) for the GADOX images and 1x1 binning mode (8 μm pixels) for the CsI image. All objects were imaged very close to the detector, in order to minimize image magnification.

3. RESULTS AND DISCUSSION

The calculated EMCCD detector MTF (Fig. 3) is shown up to the nyquist frequency, for 144, 96 and 48 μm pixel sizes. The resulting MTF is dominated by blurring in the CsI(Tl) . The curves suggest that we can expect about a 12%, 7%, and 4% contrast at the corresponding nyquist frequencies of 3.4, 5.2 and 10.4 cycles/mm respectively. Decreasing the phosphor thickness to 150 μm would increase these values by roughly a factor of 3 [19] which would also tend to decrease the QDE of the phosphor by a factor of ~ 1.6 .

Figure 4 shows DQE curves as a function of EMCCD gain at an exposure of 0.1 μR for 144, 96 and 48 μm pixel sizes. Significant improvements in the DQE are observed when operating at 50x gain as compared to the lower gains, for all three pixel sizes. Increases of up to 27%, 58% and 170% for the 144, 96 and 48 μm pixel sizes, respectively, were calculated by integrating the DQE over all frequencies and comparing the results for the 10x and 50x gain modes. This suggests that the additional gain provided by the EMCCD may provide substantial improvements at fluoroscopic exposures. Increasing the gain above 50x resulted in minimal to modest improvement, depending on the pixel size (2% to 13% total increase from 50x to 300x for the three pixel sizes). This indicates that for practical purposes, a gain greater than 50x will be unnecessary. Figure 4 also shows DQE curves as a function of detector entrance exposure using 20x gain for 144, 96 and 48 μm pixel sizes. An increase in exposure from 0.1 to 1 μR results in an increase in the DQE for all three pixel sizes (6%, 16%, and 51% total increase respectively). Above 1 μR , the DQE improves slightly for the 48 μm pixel size (8% from 1 to 100 μR) and remains constant ($< 2\%$ increase) for the larger pixel sizes.

This seems to indicate that the EMCCD detector module will be quantum limited (integral DQE > 95% of maximum) when using an EMCCD gain of ~80x and a 48 μm pixel size, at an exposure of 0.1 μR , which is significantly lower than median fluoroscopic exposure rates of ~2 μR [20]. The larger pixel sizes are expected to exhibit quantum limited behavior at a gain less than 50x. If the exposure is increased to 1 μR , a gain of only 12.5x, 18x, and 31x should provide quantum limited imaging for 144 μm , 96 μm , and 48 μm pixel sizes respectively.

At an EMCCD gain of 1x, the inherent INEE was calculated to be 1.4 μR , 4.0 μR , and 22.0 μR for 144 μm , 96 μm , and 48 μm pixel sizes respectively. Experimental INEE measurements have been made for a similar (CCD based) detector [21], operating with 50 μm pixels, of ~20 μR , which is in good agreement with the theoretically predicted INEE of the EMCCD detector. Because the effective read noise of the detector decreases inversely with EMCCD gain (equation 6), the INEE will decrease for increasing gains. At a gain of 20x, the INEE was calculated to be less than 0.1 μR for all three pixel sizes.

Improvements to the current detector design can be made by directly growing/depositing the phosphor onto the FOT. This would eliminate the need for FOP1, and increase the detector efficiency by a factor of 4. Back illuminated EMCCD sensors are also available on the market. These sensors can have a conversion efficiency of ~90% compared to that of the front-illuminated sensors which are around 60%. This would provide an additional 1.5x improvement in detector efficiency.

4. CONCLUSIONS

A new EMCCD x-ray detector is shown to provide substantial improvements in overall system DQE at 0.1 μR using signal amplification of 50x, in addition to a low INEE (<0.1 μR), indicating an ability to efficiently operate down to fluoroscopic exposures. Initial test images show the detector is capable of providing high resolution angiographic and radiographic images. A modular array of such detector elements could be incorporated to increase the FOV to the desired size. This new technology may provide improvements over current flat-panel detectors for applications such as fluoroscopy and angiography requiring high frame rates, resolution, dynamic range and sensitivity while maintaining essentially no lag and very low INEE.

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Figure 1.

(Left) EMCCD based prototype module. The camera head is shown without an additional FOT. The power supply unit is shown attached to the right. (Right) a schematic of the EMCCD detector module. To increase the transmission efficiency of the light photons, the Phosphor could be grown/deposited directly onto the FOT, eliminating the need of FOP1.

Figure 2.
Schematic of the imaging cascade in the EMCCD detector module.

Figure 3.

Theoretical MTF for the EMCCD detector module for 144 μm (solid line), 96 μm (dotted line), and 48 μm (dashed line) pixel sizes. Frequency dependent contributions from the finite fiber size of the fiberoptics, the pixilation of the sensor, CsI blurring, and K-fluorescence reabsorption have been included.

Figure 4. Calculated DQE curves as a function of EMCCD gain (left column) and detector entrance exposure (right column), plotted up to the corresponding Nyquist frequency for each binned mode (pixel size).

Figure 5.

Display of asymmetric stent, which consists of a stainless steel mesh laser welded onto a Guidant TriStar stent (Abbott Vascular, Abbott Park, IL) using a $5\text{mg}/\text{cm}^2$ GADOX phosphor for a 66kVp beam. The folds of the stainless steel mesh, resulting from the crimping process, and the puddling of molten steel at the edge of the mesh, resulting from the cutout process, are clearly visible.

Figure 6. (Left) An operational amplifier in which the 1 mil ($25\ \mu\text{m}$) gold wires are clearly visible (using the GADOX phosphor) for a 54kVp with 0.8mm Al additional filtration placed near the source. (Right) A mammographic line pair phantom (computer zoomed) taken using a CsI(Tl) phosphor clearly shows 15 lp/mm for a 50kVp beam with 0.8mm Al additional filtration.

Table 1

Values of constants and various gain/noise components used in the parallel cascade model (Fig. 2).

Parameter	Notation	Stage # (Fig. 2)	Estimated value	
			I (i=1)	Cs (i=2)
Fluence [xph/(mm ² mR)]	$\Phi_0/(X \text{ Area}_{\text{pixel}})$	0	2.85×10 ⁵	
CsI quantum detection efficiency	η	1	0.79	
Fractional weights of I and Cs	w_i	2	0.488	0.512
K-shell contribution to photoelectric (PE) effect	ζ	2	0.833 (weighted average)	
K-fluoro yield	ω	2	0.870 (weighted average)	
K-fluoro absorption probability	f_{Ki}	2C'	0.630 (weighted average)	
Frequency of K _α production	$I_{\alpha i}$	2C'	0.814	0.822
K-fluoro absorption MTF	$T_{\text{Ki}}(f)$	2C''	Reference #11	
CsI conversion gain [lph/absorbed xph]	$g_A g_{\text{Bi}} g_{\text{Ci}}$	3	$E_{\text{A,Bi,Ci}}/17\text{eV}$	
MTF caused by spread of light in CsI(Tl)	$T_4(f)$	4	Reference #10	
Trans. of lph through FOP1	κ_1	5	0.27	
Trans. of lph through FOT (3:1)	τ	6	0.05	
Trans. of lph through FOP2	κ_2	7	0.60	
Blurring in FOP/FOT, pixilation	$T_8(f), T_{11}(f)$	8,11	$\text{Sin}(\pi df)/(\pi df)$; d = element size (i.e. fiber/pixel size)	
Quantum efficiency of EMCCD [e ⁻ /lph]	Q	9	0.60	
Additive noise due to thermal e ⁻	σ_t	10	1 e ⁻ /pixel/sec	
EMCCD gain	G	12	1 to 1000	
Additive noise due to readout [e ⁻]	σ_r	13	20	