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Quantitative analysis of an enlarged area Solid State X-ray Image Intensifier (SSXII) detector based on Electron Multiplying Charge Coupled Device (EMCCD) technology

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Abstract

Present day treatment for neurovascular pathological conditions involves the use of devices with very small features such as stents, coils, and balloons; hence, these interventional procedures demand high resolution x-ray imaging under fluoroscopic conditions to provide the capability to guide the deployment of these fine endovascular devices. To address this issue, a high resolution x-ray detector based on EMCCD technology is being developed. The EMCCD field-of-view is enlarged using a fiber-optic taper so that the detector features an effective pixel size of 37 μm giving it a Nyquist frequency of 13.5 lp/mm, which is significantly higher than that of the state of the art Flat Panel Detectors (FPD). Quantitative analysis of the detector, including gain calibration, instrumentation noise equivalent exposure (INEE) and modulation transfer function (MTF) determination, are presented in this work. The gain of the detector is a function of the detector temperature; with the detector cooled to 5° C, the highest relative gain that could be achieved was calculated to be 116 times. At this gain setting, the lowest INEE was measured to be 0.6 $\mu\text{R}/\text{frame}$. The MTF, measured using the edge method, was over 2% up to 7 cycles/mm. To evaluate the performance of the detector under clinical conditions, an aneurysm model was placed over an anthropomorphic head phantom and a coil was guided into the aneurysm under fluoroscopic guidance using the detector. Image sequences from the procedure are presented demonstrating the high resolution of this SSXII.

DESCRIPTION OF PURPOSE

Treatments for neurovascular conditions such as aneurysms involve guiding a catheter to the region of treatment using x-ray fluoroscopic guidance. Once the catheter is deployed into the aneurysm region, treatment devices such as stents, coils, and balloons are then deployed into the aneurysm. These devices have small features which demand imaging devices with a high spatial resolution. In order to address the high resolution concerns a high resolution x-ray detector based on EMCCD technology from e2v Technologies Ltd, United Kingdom was developed (Figure 1)[1]. The detector features an effective pixel size of 37 μm giving it a

Nyquist frequency of 13.5 lp/mm which is significantly higher than the state of the art Flat Panel Detectors (FPD) having a Nyquist frequency of less than 2.6 lp/mm. The field-of-view for this SSXII detector is enlarged to around $3.5 \text{ cm} \times 3.5 \text{ cm}$ with the fiber-optic taper (FOT) and is dependent on the dimensions of the FOT. A different SSXII detector based on EMCCD from Texas Instruments, Japan was previously developed and reported [2][3], with a field of view of $2.4 \text{ cm} \times 2.4 \text{ cm}$ and a resolution of upto 20 lp/mm. The current detector is larger in field of view but lower in resolution as compared to [2][3].

This paper presents the quantitative analysis performed on the detector developed in [1]. For the purpose of this study, a circular fiber optic taper (2.88 ratio) was used. First the relative gains of the detector were derived; then using the relative gain data, the instrumentation noise equivalent exposure (INEE) of the detector was determined. In order to evaluate the spatial resolution of the detector, the Modulation Transfer Function (MTF) was determined and presented.

CONSTRUCTION

The schematic of a single module EMCCD camera is shown in Figure 1. The camera consists of the following components

1. Input Phosphor: 350 μm thick Cesium Iodide phosphor coupled to a fiber optic window. The input x-rays are absorbed by the phosphor and converted into light photons coming out of the fiber optic window.
2. Fiber optic taper: Magnification ratio of 2.88. The light from the fiber optic window of the phosphor is collected by the larger end of the taper. The smaller end of the taper is coupled to a fiber optic window on the image area of the sensor.
3. EMCCD sensor mounted on a headboard, collects the light from the small end of the fiber optic taper and converts it to an analog voltage.
4. The electronic boards supplying the power, clocks and readout electronics for the sensor. The image from the sensor is read pixel by pixel, the analog voltage from each pixel is converted to a digital value using a 12 bit Analog to Digital converter. This data is transferred to a frame grabber installed in an acquisition computer, using cameralink communication technology. The camera is triggered using an external trigger which is synchronized to the trigger pulse for the x-ray generator pulse.

Figure 2 shows the inside assembly of the camera with different components. Figure 3 shows the x-ray setup and the data transfer link from the camera to the acquisition computer.

QUANTITATIVE ANALYSIS

Gain calibration

EMCCD's are similar to CCD's, but with a charge multiplication stage before the readout amplifier as shown in Figure 4. A certain voltage applied across the multiplication registers, causes impact ionization, and many such registers (and hence impact ionization processes) in series leads to charge multiplication. The multiplication achieved is a function of two

parameters, the voltage applied to the register and the temperature of the chip (substrate). For the same voltage, the lower the temperature, the higher is the gain achieved. [1]

For the detector in question, a serial command consisting of a gain number is sent from the acquisition computer to the camera (using the serial communication channel of the CameraLink interface) to change the voltage across the multiplication register. The higher the gain number, the higher is the voltage and hence the higher is the gain achieved for a particular temperature settings. A thermoelectric cooling system is used to cool the sensor below room temperatures and achieve higher gains.

To calibrate the relative gain, the sensor was cooled to 5°C. The input exposure was varied from 5 $\mu\text{R}/\text{frame}$ to 40 $\mu\text{R}/\text{frame}$. The gain number was varied from 45000 to 55000 for each exposure. For each gain number the signal (in digital number)/ μR is calculated. The gains for the respective gain number are calculated relative to the gain number of 45000. Table 1 gives the non-linear relation between the gain number, respective signal/ μR and the resulting relative gain.

Instrumentation Noise Equivalent Exposure (INEE)

The equation for the output signal to noise ratio (SNR)[4] for a particular gain G is

$$SNR = \frac{G \times \text{Signal}}{\sqrt{(G^2 \times \text{Variance}_{(\text{Input quanta})} + G^2 \times \text{Variance}_{(\text{Dark Noise})} + \text{Variance}_{(\text{instrumentation noise})})}}$$

The instrumentation noise in the EMCCD chip is a function of gain. At higher gains the signal is amplified to a value above the instrumentation noise floor of the system, so that the noise at the output is dominated by the noise in the input quanta. Under this condition in the quantum limited region of operation, the output SNR of the camera is limited by the input SNR. INEE is the input exposure at which the input quantum noise is equal to the instrumentation noise of the detector [5]. The INEE is a function of the multiplication gain of the detector. In order to calculate the INEE, various flat fields for exposures varying from 2 $\mu\text{R}/\text{frame}$ to 90 $\mu\text{R}/\text{frame}$ at different gains varying from 48000 to 55000 were taken. The corresponding relative gain is given in table 1. After flat field correction a plot of variance vs. exposures for different gains were plotted as shown in Figure 5a–g. After a linear fit, the INEE was calculated by taking the ratio of the intercept and the slope, for different gains as shown in table 2.

By increasing the gain, the signal can be amplified above the noise floor, thus reducing the INEE. From the table it can be seen that INEE varies inversely as a function of gain. But gain for a given multiplication voltage is a function of operating temperature of the chip. For the measurements presented, the chip was cooled to 5° C, with the gain number and the corresponding relative gain given in table 1. The INEE for the highest gain used in this work was calculated to be 0.6 μR . By cooling the chip further, higher gains can be achieved and thus the INEE can be reduced further. Also, since the gain stage also amplifies the dark current and CIC noise in EMCCD's, care has to be taken to minimize these noise sources.

MTF Measurement

The MTF of the detector at various spatial frequencies was measured using the edge method [6]. The edge was placed up against the detector. The small focal spot was chosen on the x-ray machine and RQA 5 spectrum was used. The plot of MTF vs. spatial frequencies is shown in Figure 6. An x-ray image of a line pair phantom is shown in Figure 7, showing up to 9 line pairs/mm resolution.

PHANTOM STUDY

To evaluate the performance of the detector under clinical conditions, an aneurysm coiling procedure was performed on an anthropomorphic head phantom [7]. To simulate the aneurysms occurring near the Circle of Willis, an aneurysm model was placed on top of the head phantom in the region between the nose and the eyes. The coil was guided into the aneurysm under fluoroscopy using the new detector. The x-ray technique parameters used were 76 kVp, 50 mA tube current and 10 ms pulse width and the detector entrance exposure was measured to be 11 μ R/frame. The detector was cooled to 5° C, the gain was set to a gain number of 54000 (from table 1, corresponding to a relative gain of 60) so that the detector is operated at an exposure 8 times above the INEE as shown in table 2. Three images from the procedure are shown in Figures 8, 9, and 10.

DESIGN EXPANSION TO M $\times$ N ARRAY

To increase the field of view, the single module design with square FOT input faces can be expanded to a M $\times$ N array, which would have the same resolution [8], and INEE as the single module. Figure 11 shows a graphic model of a 3 $\times$ 3 array. A 2 $\times$ 2 array was designed and built, achieving a total field of view of 8.45cm $\times$ 8.45cm with a resolution of 11.2 line pair/mm. Figure 12 shows the cross section of the detector. Figure 13 shows a front view of the 2 $\times$ 2 fused array of the fiber optic taper of the detector.

CONCLUSION

The gain of the detector is a function of the detector temperature; with the detector cooled to 5° C the highest relative gain that could be achieved was calculated to be 116 times. At this gain, the lowest INEE was measured to be 0.6 μ R/frame. The MTF presented is significantly higher than the state of the art flat panel detectors. From the aneurysm coiling experiment it appears that the detector may be useful to satisfy the need for high resolution x-ray imaging for guiding clinical endovascular neuro-interventions. The field of view of the detector can be expanded by making an array of individual modules, but having the same high resolution as the single module.

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Figure 1.
Schematic diagram of a single module SSXII detector

Figure 2.
Internal assembly of the camera: a) Phosphor, b) Fiber Optic Taper, c) EMCCD, d) Driver Board, e) Clock Generator, f) Camera Link Interface

Figure 3.
X-ray Setup and data transfer link

Figure 4.
Functional Blocks of an EMCCD chip

a: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 2.91

b: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 7.03

c: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 11.42

d: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 18.78

e: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 34.24

f: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 59.78

g: Plot of Variance (in DN) vs. Exposure at a Relative Gain of 116.38

Figure 5.

Figure 6.
MTF of the detector

Figure 7.
X-ray image of a line pair phantom

Figure 8.
Beginning of the aneurysm coiling procedure

Figure 9.
Part of the coil guided into the aneurysm

Figure 10.
Entire coil guided into the aneurysm

Figure 11.
A graphic model of a 3×3 array of detectors

Figure 12.

Cross section of the 2×2 detector array, a) Fiber Optic Taper, b) EMCCDs, c) Electronic Boards, d) CameraLink interface, e) Cooling system

Figure 13.
Top View of a 2×2 fused array used in the detector

Table 1

Gain number vs. Relative gain

Gain number	Signal/ μ r	Relative gain
45000	1.65	1.00
46000	2.32	1.41
48000	4.80	2.91
49000	7.54	4.57
50000	11.60	7.03
51000	18.83	11.42
52000	30.98	18.78
53000	56.48	34.24
54000	98.60	59.78
55000	191.95	116.38

Table 2

INEE vs. Gain number

Gain number	Slope	Intercept	INEE (μ r)
48000	3.7039	54.234	14.6424
50000	17.117	69.409	4.054975
51000	36.886	119.52	3.240254
52000	86.823	264.32	3.044355
53000	249.7	499.29	1.999559
54000	811	1155	1.424168
55000	3106.9	1901.1	0.611896