Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2009 January 1; 7258: 725817-1–725817-10. doi:10.1117/12.813957.

Component analysis of a new Solid State X-ray Image Intensifier (SSXII) using photon transfer and Instrumentation Noise Equivalent Exposure (INEE) measurements

Andrew Kuhls-Gilcris^{*}, Daniel R. Bednarek, and Stephen Rudin

University at Buffalo (State University of New York), Toshiba Stroke Research Center, 3435 Main St., Buffalo, NY USA 14214

Abstract

The SSXII is a novel x-ray imager designed to improve upon the performance limitations of conventional dynamic radiographic/fluoroscopic imagers related to resolution, charge-trapping, frame-rate, and instrumentation-noise. The SSXII consists of a CsI:Tl phosphor coupled via a fiber-optic taper (FOT) to an electron-multiplying CCD (EMCCD). To facilitate investigational studies, initial designs enable interchangeability of such imaging components. Measurements of various component and configuration characteristics enable an optimization analysis with respect to overall detector performance. Photon transfer was used to characterize the EMCCD performance including ADC sensitivity, read-noise, full-well capacity and quantum efficiency. X-ray sensitivity was measured using RQA x-ray spectra. Imaging components were analyzed in terms of their MTF and transmission efficiency. The EMCCD was measured to have a very low effective read-noise of less than 1 electron rms at modest EMCCD gains, which is more than two orders-of-magnitude less than flat panel (FPD) and CMOS-based detectors. The variable signal amplification from 1 to 2000 times enables selectable sensitivities ranging from 8.5 (168) to over 15k (300k) electrons per incident x-ray photon with (without) a 4:1 FOT; these sensitivities could be readily increased with further component optimization. MTF and DQE measurements indicate the SSXII performance is comparable to current state-of-the-art detectors at low spatial frequencies and far exceeds them at higher spatial frequencies. The instrumentation noise equivalent exposure (INEE) was measured to be less than 0.3 μ R out to 10 cycles/mm, which is substantially better than FPDs. Component analysis suggests that these improvements can be substantially increased with further detector optimization.

Keywords

solid-state x-ray image intensifier; SSXII; EMCCD; detectors; high-resolution; MTF; DQE; INEE

1. INTRODUCTION

Improved visualization of fine anatomical morphology and structural details of interventional devices to enable accurate deployment requires increasingly higher resolution and higher quality images, obtainable in real time – all while keeping the patient dose to a

^{*}Corresponding author: Andrew Kuhls-Gilcris. Email: atkuhls@buffalo.edu; Dept. of Physics; Phone: 716 829 3595.

minimum^{1, 2}. Flat panel detectors (FPD) have been widely adopted as the technology of choice, replacing x-ray image intensifiers (XII). Much work is underway in an effort to improve upon the various limitations of FPDs, including clever new design concepts and computer algorithms, aimed at reducing the effect of instrumentation noise^{3,4} and temporal blur^{5,6}, while moving towards smaller pixel sizes⁷ (with reasonably high fill factors). Time will tell how far FPD technology will advance and how well it can adapt to fulfill the increasingly demanding requirements and expectations placed on medical x-ray imagers. However, a prototype solid state x-ray image intensifier (SSXII) module⁸, currently under development, has *already* demonstrated great promise to provide the improved imaging capabilities necessary to meet 21st century demands. The SSXII has been shown to provide excellent distortion-free, high-resolution images with negligible instrumentation noise and temporal blur⁹. The detector MTF and DQE have been measured to objectively quantify the vast improvements over current technologies^{10–12}.

The goal of this work was to measure the performance of each *individual* component in the imaging chain of the SSXII and to then study the resulting effect each component has on the overall detector performance. It is then possible to discern if the components are performing satisfactorily and to weigh the cost and benefits of each component. Such an optimization analysis is expected to lead to even further improvements beyond those currently realized. Extrapolated performance measures for various design configurations are used to quantify such potential improvements. Measurements of detector performance were also done on a Varian Paxscan 2020+ FPD (Palo Alto, CA) to put the results in perspective with current state-of-the-art detector technologies.

2. MATERIALS AND METHODS

2.1 The Solid-State X-Ray Image Intensifier

The SSXII consists of a 350 μm thick CsI:Tl phosphor coupled to an EMCCD camera (Photonic Science LTD, East Sussex, UK) via a 4:1 FOT. A schematic of the SSXII prototype module is shown in figure 1. The imaging components are packaged in a light-tight enclosure which can be mounted on a clinical c-arm gantry for combined large field-of-view (FOV) low-resolution imaging with a FPD and high-resolution region-of-interest (ROI) imaging with the SSXII^{14, 15}.

To facilitate investigational studies, the initial designs employ two fiber optic plates (FOPs) to enable interchangeability of the phosphor and FOT. The phosphor is grown on a separate FOP (commonly referred to as a fiber optic scintillating plate or FOS) and the EMCCD camera has a fiber optic input window (i.e. a FOP bonded to the EMCCD sensor). FOTs with varying magnification ratios and different phosphor materials/thicknesses can be installed or omitted in order to determine their effects on the overall detector performance, enabling an optimization analysis specific to a particular imaging task (i.e. neurovascular, chest, or breast imaging).

2.2 Photon Transfer

2.2.1 Theory—Photon transfer has been used over the past several decades for calibrating, characterizing, and optimizing CCD performance¹⁶ and more recently for characterizing

EMCCD performance¹⁷. The photon transfer technique exploits the signal and noise transfer properties of a CCD (or EMCCD) camera system. When exposed to a uniform Poisson distributed light source, the resulting image signal distribution can be used to obtain valuable information regarding sensor performance *without* knowledge of individual transfer functions. Perhaps the most useful result of the photon transfer technique is the camera gain constant, K , which provides the conversion factor from arbitrary units of digital numbers (DN) to absolute units of electrons (e^-), by simply measuring the signal and signal variance (image noise) of an image formed from a uniform Poissonian light source. K can be determined using the following equation¹⁸,

$$K = \frac{S(DN)}{\sigma_S^2(DN) - \sigma_R^2(DN)}, \quad (1)$$

where $S(DN)$ is the average output signal, $\sigma_S^2(DN)$ is the output signal variance and $\sigma_R^2(DN)$ is the read-noise floor variance (i.e. the output signal variance of a dark-field or zero-signal image), all in units of DN. Sensor performance can then be measured in absolute units of e^- : the read noise can be determined from the zero-signal variance; the full well capacity (S_{FW}) can be obtained by plotting the signal variance versus signal where the signal corresponding to the maximum variance corresponds to the point where the pixels begin to saturate, above which the noise will subsequently decrease; the dynamic range can be determined by taking a ratio of the full well capacity and the read noise and converting to decibels if desired.

2.2.2 Measurement—To measure the EMCCD sensor performance using the photon transfer technique, the phosphor was removed because the x-ray to optical photon conversion process provides a stochastic gain (i.e., has a gain greater than 1) and would result in a non-Poisson distribution of optical photons incident on the sensor. The EMCCD sensor was then exposed to a 2×4 array of LED light bars (Model KB2885SGD, Kingbright Corp., Taiwan), providing an approximately uniform light source with a peak wavelength of 565 nm, similar to that of CsI:Tl. The intensity of the light was varied by varying the voltage supplied to the diodes such that the measurements were done across the full range of the detector. Measurements were also done at several different EMCCD gains. Due to the probabilistic nature of the EMCCD gain process, an additional noise component was required to be taken into account as described by the excess-noise factor¹⁹, F , where

$$F = \sqrt{2(M - 1)M^{-\frac{N+1}{N}} + \frac{1}{M}}, \quad (2)$$

M is the EMCCD gain and N is the total number of multiplication stages in the multiplication register ($N = 400$ for the sensor used). Equation 1 was then modified to take into account this additional noise using the following equation,

$$K = \frac{S(DN)}{\frac{\sigma_S^2(DN)}{F^2} - \sigma_R^2(DN)} \quad (3)$$

for gain values greater than unity. It has been shown that this excess-noise term is negligible when imaging secondary (optical) quanta from an x-ray converting phosphor²⁰, and is applicable only when imaging optical photons with the phosphor removed.

2.3 X-Ray Sensitivity

The detector x-ray sensitivity (after re-coupling of the CsI:Tl phosphor) was measured in units of electrons at the EMCCD sensor per incident x-ray photon (e^-/xph). Two different standard x-ray spectra (RQA 3 and RQA 5) were used. A range of exposures, measured using a calibrated ionization chamber (Keithley 35050A) in accordance with the IEC guidelines²¹, covering the operating range of the detector were used. The slope of a plot of average signal versus detector entrance exposure was determined and provided the x-ray sensitivity in units of DN per unit exposure. This arbitrary sensitivity was then converted to absolute physical units ($e^-/\text{inc xph}$) using the camera gain constant, K, and the calculated x-ray photon fluence²².

2.4 MTF and Transmission Efficiency of Individual Optical Components

The MTF and transmission efficiency were measured for individual optical components using various configurations of the detector system. Measurements were done with both x-ray and optical photons, depending on whether or not the CsI:Tl phosphor was implemented. The x-ray exposure was measured using the calibrated ionization chamber and the optical photon flux was measured using a calibrated photodiode (Model FDS100-CAL, Thorlabs, Newton, New Jersey). MTF measurements were done using the standard edge response method²³, where a straight edge, slightly angulated at ~ 1.5 degrees, was placed directly on the input surface of the detector for each configuration studied.

The MTF of the phosphor was estimated by measuring the detector MTF with and without the CsI:Tl coupled directly to the FOP (i.e. without the FOT in both cases), using x-rays and visible light, respectively, according to

$$\text{MTF}_{\text{CsI}} = \frac{\text{MTF}_{\text{DETECTOR W/CsI}}}{\text{MTF}_{\text{DETECTOR W/O CsI}}} \quad (4)$$

The MTF and transmission efficiency of the 4:1 FOT was determined by measuring the MTF and x-ray sensitivity with and without the 4:1 FOT in the imaging chain, both with and without the $350 \mu\text{m}$ CsI:Tl phosphor, using x-rays (RQA 3 and RQA 5) and visible light, respectively as described by

$$\text{MTF}_{\text{FOT XRAY}} = \left(\frac{\text{MTF}_{\text{DETECTOR W/FOT}}}{\text{MTF}_{\text{DETECTOR W/O FOT}}} \right)_{\text{XRAY}} \quad \text{and} \quad \text{MTF}_{\text{FOT VISIBLE}} = \left(\frac{\text{MTF}_{\text{DETECTOR W/FOT}}}{\text{MTF}_{\text{DETECTOR W/O FOT}}} \right)_{\text{VISIBLE}} \quad (5)$$

The MTF of the FOT was determined from the visible light measurements and a ratio of the x-ray and visible light measurements was used to estimate the MTF associated with use of x-rays (e.g., due to Compton scattering or veiling glare), where

$$\text{MTF}_{\text{XRAY}} = \frac{\text{MTF}_{\text{FOT XRAY}}}{\text{MTF}_{\text{FOT VISIBLE}}}. \quad (6)$$

The combined MTF of the EMCCD sensor and FOP was measured by illuminating the edge, placed directly on the surface of the FOP, with the LED light bar array and the transmission efficiency was measured using a plot of the number of incident light photons per pixel versus the number of visible light photons absorbed at the sensor (i.e. the image signal in electrons).

For each measurement, the fiber optic components were removed and then re-coupled three times and the transmission efficiency/MTF was measured, each time, to minimize potential variability resulting from differences in the optical couplings. The standard deviation of the three measurements was used to provide an indication of the variability or precision of the measurements.

2.5 Spatial-Frequency-Dependent Instrumentation Noise Equivalent Exposure (INEE)

The performance of high-sensitivity x-ray imagers may be limited by additive electronic instrumentation noise rather than by quantum noise when operated at the low exposure rates used in fluoroscopic procedures. Instrumentation noise is often specified by vendors in units of electrons and medical physicists often use the normalized noise power spectrum in units of mm^2 , both of which are generally not as intuitive as would be a direct radiological representation (in terms of exposure). Previous attempts to gauge the quantum noise limited performance range, most common of which examines the constancy of the product of the NNPS and exposure (or similarly, the constancy of DQE) with exposure²⁴ to verify instrumentation-noise free operation over a particular range of exposures, fall short of providing a direct quantitative measure. The instrumentation noise equivalent exposure (INEE) addresses this need by providing a *direct* quantitative measure of the quantum noise limited exposure range of x-ray detectors by providing the exposure at which the additive instrumentation-noise equals the quantum-noise^{8, 25, 26}. The INEE can be directly determined through experimental measurements of image noise (or signal-variance) as a function of incident exposure.

The INEE methodology may be generalized to include spatial-frequency effects by using the two-dimensional NPS instead of the signal-variance²⁷, where the NPS can be considered to be the variance divided among the various frequency components of the image²². The NPS for each noise source is additive²⁸ and can be separated into a quantum noise term which scales with the exposure and an additive noise term which is independent of exposure

$$\text{NPS}(u, v, E) = \text{NPS}'_Q(u, v)E + \text{NPS}_{\text{ADD}}(u, v) \quad (7)$$

where $NPS(u, v, E)$ is the measured detector NPS at exposure E and spatial frequencies u and v . $NPS'_Q(u, v)$ is the quantum NPS per unit exposure, which characterizes the noise resulting from fluctuations in the number of quanta. The product of $NPS'_Q(u, v)$ and E is the actual quantum NPS. $NPS_{ADD}(u, v)$ is the additive NPS which characterizes all additive noise sources that are independent of E , largely resulting from the additive electronic instrumentation noise. The images are gain and offset corrected, so there is no structure term appearing in equation 7, which would scale proportionally with E^2 . The different noise contributions can then be separated using a straight line fit of the $NPS(u, v, E)$ plot as a function of exposure, at each spatial frequency u and v . The exposure at which the quantum-noise equals the additive electronic instrumentation noise [i.e., the instrumentation noise equivalent exposure (INEE)], for each spatial frequency, can then be determined by dividing the intercept by the slope of the fitted line or alternatively by taking a ratio of the additive NPS and the quantum NPS per unit exposure:

$$INEE(u, v) = \frac{NPS_{ADD}(u, v)}{NPS'_Q(u, v)}. \quad (8)$$

3. RESULTS AND DISCUSSION

3.1 EMCCD Performance Characterization

Images were acquired for several different LED intensities such that the average signal ranged from 10% to 70% of the maximum signal and the average signal and signal variance were calculated in a central 250×250 pixel region for the entire image sequence, which consisted of 150 images. The uniformity of the light across this central region was measured to be better than 99%. The photon transfer technique was then utilized (equation 3) to determine the camera gain constant for various EMCCD gain values ranging from 1 to 100 $\times$.

The camera gain constant, K , was measured to be $15.0 \pm 0.9 \text{ e}^-/\text{DN}$. An increase in EMCCD gain resulted in a proportional decrease in K , as illustrated by the constancy of the product of K and the EMCCD gain, as shown in figure 2. A decrease in K corresponds to an increase in detector sensitivity, which scales with the inverse of K . A summary of the various performance measures of the EMCCD are shown in table 1. The inherent read noise was measured to be $240 \pm 20 \text{ e}^-$ and was found to decrease proportionally with the EMCCD gain, as expected. The read noise was measured to be less than 1 e^- at a gain of 250 $\times$. The next-generation prototype, currently in production, is expected to have an order-of-magnitude less read noise than that currently measured due to improved electronic design, more in-line with current state-of-the-art CCDs which are limited by the amplifier noise (specified by the vendor to be 22 e^-). The full well capacity was measured to be $49,000 \pm 3,000 \text{ e}^-$, slightly higher than the manufacturer specified value of 40,000. The resulting dynamic range was 46 dB at a gain of 1, which is less than the expected 66 dB, attributable to the higher read noise. However, given the *variable* sensitivity of the sensor, the effective dynamic range of the detector is actually much larger. For instance, when considering the maximum and minimum detectable signals *across* different gains, the dynamic range is 93

dB (maximum detectable signal equals $49,000\text{ e}^-$ at a gain of $1\times$ and the minimum detectable signal equals 1 e^- at a gain of $250\times$).

3.2 X-Ray Sensitivity

A plot of signal versus x-ray photon flux (figure 3) was used to determine the x-ray sensitivity of the SSXII, at a gain of $1\times$, with and without the 4:1 FOT for two different x-ray spectra. For RQA 5 the x-ray sensitivity was measured to be $168 \pm 5\text{ e}^-/\text{xph}$ with no FOT and $8.56 \pm 0.06\text{ e}^-/\text{xph}$ with the 4:1 FOT. Similarly, for RQA 3 the sensitivities were 128 ± 1 and $6.8 \pm 0.4\text{ e}^-/\text{xph}$, respectively.

3.3 MTF and Transmission Efficiency of Individual Optical Components

The MTF of the various components that comprise the imaging chain of the SSXII prototype module are shown in figure 4. The total detector MTF was then compared with two state-of-the-art FPDs (figure 5), a Varian Paxscan 2020+ FPD (with $194\text{ }\mu\text{m}$ square pixels) measured in our laboratory and a Varian Paxscan 2520 (with $127\text{ }\mu\text{m}$ square pixels) digitized from published results³⁰. The MTF of the SSXII was 0.20 at 3 cycles/mm, comparable to standard detectors, and better than 0.05 up to 7 cycles/mm, well beyond current capabilities. With further component optimization (i.e. improving or eliminating the FOP and FOT and minimizing Compton scattering) the MTF may be significantly improved, as shown maximally with the “SSXII – Optimized” curve. This optimization could result in an MTF of 0.3 at 3 cycles/mm and more than 0.05 out to 14 cycles/mm with (and limited by) a $350\text{ }\mu\text{m}$ thick CsI phosphor – increasing the performance even further beyond current state-of-the-art detectors, for all spatial frequencies.

A ratio of the 4:1 FOT versus no FOT x-ray sensitivity measurements provided the transmission efficiency of the FOT including coupling losses, which was determined to be 0.050 ± 0.002 . A similar plot of signal versus light photon flux was used to determine the interacting quantum efficiency of the EMCCD and the transmission efficiency of the FOP, which was measured to be a combined 0.48 ± 0.02 for an approximately collimated light source. This is in agreement with the vendor specified quantum efficiency of the sensor²⁹ and the transmission efficiency of the FOP at 565 nm (60% and 80%, respectively) and suggests there are minimal losses at the coupling interface of the optical components.

Figure 6 shows a comparison plot of the detector DQE between the SSXII and the FPD, acquired at $2.7\text{ }\mu\text{R}/12$. Also shown is an “SSXII – optimized” curve which indicates the extrapolated performance after component optimization resulting from eliminating one of the two FOPs and by decreasing the FOT ratio from 4:1 to 2:1.

3.4 Spatial-Frequency-Dependent Instrumentation Noise Equivalent Exposure (INEE)

The 2D NPS was measured for 11 different exposures ranging from 0.3 to $25\text{ }\mu\text{R}$. The NPS was then plotted as a function of exposure at each spatial frequency u and v and a linear regression was fit to the data with an average r squared value of more than 0.997. The quantum NPS and the additive NPS were then determined using the slope and intercept of this fit, respectively. Figure 7 shows a representative example of the technique for INEE measurements. The frequency dependent INEE was then determined by locating the

exposure at which the additive electronic instrumentation noise and quantum noise were equivalent, using equation 8. The radially averaged (between 18 and 72 degrees) INEE is shown in figure 8. At 2.5 cycles/mm the INEE for the SSXII was 0.20 μR , significantly lower than that of the FPD, which was measured to be 4.6 μR . Furthermore, the SSXII maintains a low INEE of 0.3 μR or less out to 10 cycles/mm, indicating the ability to maintain quantum limited performance at even the lowest of exposures at *all* spatial frequencies of interest.

The flatness of the SSXII curve, at higher spatial frequencies, requires further investigation. The quantum NPS is comprised of contributions from three fundamental noise sources, all which scale proportionally with exposure: *primary quantum noise* which results from inherent variations in the incident number of x-ray quanta; *secondary quantum noise* which results from variations in the number of secondary quanta detected per primary quanta absorbed; and *Poisson excess noise* which results from variations in the x-ray energy absorbed in the phosphor. Subsequent development of the INEE31 will involve distinguishing the various quantum NPS components and determining the exposure at which the total instrumentation noise (which includes both additive electronic noise *and* noise resulting from the conversion and transmission of quanta) equals the primary quantum noise. A system dominated by secondary quantum noise may exhibit a relatively constant INEE in the current formalism. Decomposition of the quantum NPS should provide greater insight.

4. SUMMARY AND CONCLUSION

A new dynamic, high-resolution x-ray imager, the solid state x-ray image intensifier (SSXII), has been developed to improve upon current-state-of-the-art detectors. A component analysis has been performed for objective performance assessment. Using photon transfer, the camera gain constant was measured, enabling measurement of the EMCCD sensor performance. Very low effective read noise levels of less than 1 electron rms were realized when using the variable signal amplification, three orders of magnitude less than that of current FPDs. The variable EMCCD gain enables *selectable* x-ray sensitivities ranging from 8.5 to more than 15k electrons per x-ray photon. The transmission efficiency and MTF associated with each component in the imaging chain were measured to diagnose their performance and to assist in further improving upon the overall detector design. Detector MTF and DQE measurements indicate that performance of the SSXII is comparable to current state-of-the-art detectors at low spatial frequencies and far exceeds them at higher spatial frequencies. Component analysis suggests that these improvements may be substantially increased with further detector optimization, potentially resulting in even greater improvements relative to standard radiographic and fluoroscopic imagers. The frequency dependent instrumentation noise equivalent exposure (INEE) was measured to be maximally 0.3 μR at 10 cycles/mm, indicating the ability to maintain quantum limited performance well below current fluoroscopic exposure rates. These results indicate that this detector technology offers great promise to be the basis of next generation dynamic x-ray imager arrays.

Acknowledgments

This work was supported in part by the UB Interdisciplinary Research Development Fund (IRDF), NIH R01 Grants EB002873, EB008425 and NS43924 and an equipment grant from Toshiba Medical Systems Corp.

References

1. Rudin S, Bednarek DR, Hoffmann KR. Endovascular image-guided interventions (EIGIs). *Med Phys.* 2008; 35:301–309. [PubMed: 18293585]
2. Ionita CN, Rudin S, Hoffmann KR, Bednarek DR. Microangiographic image guided localization of a new asymmetric stent for treatment of cerebral aneurysms. *Proc of SPIE.* 2005; 5744:354–365. [PubMed: 21311733]
3. Zhao W, Li D, Rowlands JA, Egami N, Takiguchi Y, Nanba M, Honda Y, Ohkawa Y, Kubota M, Tanioka K, Kawai T. An indirect flat-panel detector with avalanche gain for low dose x-ray imaging: SAPHIRE (Scintillator Avalanche Photoconductor with High Resolution Emitter readout). *Proc of SPIE.* 2008; 6913:69130M.
4. Antonuk LE, El-Mohri Y, Zhao Q, Konieczek M, McDonald J, Yeakey M, Wang Y, Behravan M. Exploration of the potential performance of polycrystalline silicon-based active matrix flat-panel imagers incorporating active pixel sensor architectures. *Proc of SPIE.* 2008; 6913:69130I.
5. Mail N, Moseley DJ, Siewerdsen JH, Jaffray DA. An empirical method for lag correction in cone-beam CT. *Medical Physics.* 2008; 35:5187–5196. [PubMed: 19070253]
6. Mollov I, Tognina C, Colbeth R. Photodiode forward bias to reduce temporal effects in a-Si based flat panel detectors. *Proc of SPIE.* 2008; 6913:69133S–69131.
7. El-Mohri Y, Antonuk LE, Zhao Q, Wang Y, Li Y, Du H, Sawant A. Performance of a high fill factor, indirect detection prototype flat-panel imager for mammography. *Med Phys.* 2007; 34:315–327. [PubMed: 17278517]
8. Rudin S, Kuhls AT, Yadava GK, Josan GC, Wu Y, Chityala RN, Rangwala HS, Ionita C, Hoffmann KR, Bednarek DR. New light-amplifier-based detector designs for high spatial resolution and high sensitivity CBCT mammography and fluoroscopy. *Proc of SPIE.* 2006; 6142:61421R.
9. Kuhls A, Yadava G, Bednarek D, Rudin S. The new Solid State X-ray Image Intensifier (SSXII): A demonstration of operation over a range of angiographic and fluoroscopic exposure levels. *Med Phys.* 2007; 34:WE-C-L100J-103.
10. Kuhls A, Yadava G, Bednarek DR, Rudin S. Progress in electron-multiplying CCD (EMCCD) based high-resolution high-sensitivity x-ray detector for fluoroscopy and radiography. *Proc of SPIE.* 2007; 6510:6510C.
11. Kuhls A, Yadava G, Bednarek D, Rudin S. Linear systems analysis for a new Solid State X-ray Image Intensifier (SSXII) based on electron-multiplying charge-coupled devices (EMCCDs). *Med Phys.* 2007; 34:WE-C-L100J-106.
12. Kuhls-Gilcrst A, Yadava G, Patel V, Jain A, Bednarek DR, Rudin S. The Solid-State X-ray Image Intensifier (SSXII): An EMCCD-based x-ray detector. *Proc of SPIE.* 2008; 6913:69130K. <http://www.pubmedcentral.nih.gov/articlerender.fcgi?tool=pubmed&pubmedid=18836568>.
13. Rudin S, Kuhls A, Keleshis C, Kim D, Yadava G, Patel V, Ionita C, Hamwi H, Cartwright A, Verevkin A, Hoffmann K, Bednarek D. The Solid State X-ray Image Intensifier (SSXII): A Next-Generation High-Resolution Fluoroscopic Detector System. *Med Phys.* 2007; 34:WE-C-L100J-104.
14. Ionita CN, Keleshis C, Patel V, Yadava G, Hoffmann KR, Bednarek DR, Jain A, Rudin S. Implementation of a High-Sensitivity Micro-Angiographic Fluoroscop (HS-MAF) for in-vivo Endovascular Image Guided Interventions (EIGI) and Region-Of-Interest Computer Tomography (ROI-CT). *Proc of SPIE.* 2008; 6918:69181I. <http://www.pubmedcentral.nih.gov/articlerender.fcgi?tool=pubmed&pubmedid=18958294>.
15. Kuhls A, Patel V, Ionita C, Noel PB, Walczak AM, Rangwala HS, Hoffmann KR, Rudin S. New microangiography system development providing improved small vessel imaging, increased contrast to noise ratios, and multi-view 3D reconstructions. *Proc of SPIE.* 2006; 6142:61423M.
16. Janesick, JR. *Photon Transfer*. SPIE: Bellingham; 2007.

17. DeWeert MJ, Cole JB, Sparks AW, Acker A. Photon transfer methods and results for electron multiplying CCDs. *Proc of SPIE*. 2004; 5558:248–259.
18. Janesick JR, Elliot T, Collins S, Blouke MM, Freeman J. Scientific charge-coupled devices. *Optical Engineering*. 1987; 26:692–714.
19. Robbins MS, Hadwen BJ. The noise performance of electron multiplying charge coupled devices. *IEEE Trans on Electron Devices*. 2003; 50:1227–1232.
20. Kuhls-Gilcris A, Bednarek D, Rudin S. Noise evaluation of a new Solid State X-ray Image Intensifier (SSXII). *Med Phys*. 2008; 35:2879.
21. IEC 62220-1:2003, Medical electrical equipment - Characteristics of digital X-ray imaging devices - Part 1: Determination of the detective quantum efficiency
22. Cunningham, IA.; Dobbins, JT, III. *Handbook of Medical Imaging Physics and Psychophysics*. SPIE: Bellingham; 2000.
23. Samei E, Flynn MJ, Reimann DA. A method for measuring the presampled MTF of digital radiographic systems using an edge test device. *Med Phys*. 1998; 25:102–113. [PubMed: 9472832]
24. Dobbins JT III, Samei E, Ranger NT, Chen Y. Intercomparison of methods for image quality characterization. II. Noise power spectrum. *Med Phys*. 2006; 33:1466–1475. [PubMed: 16752581]
25. Yadava GK, Kuhls-Gilcris AT, Rudin S, Patel VK, Hoffmann KR, Bednarek DR. A practical exposure-equivalent metric for instrumentation noise in x-ray imaging systems. *Phys Med Biol*. 2008; 53:5107–5121. <http://www.pubmedcentral.nih.gov/articlerender.fcgi?tool=pubmed&pubmedid=18723932>. [PubMed: 18723932]
26. Szczykutowicz T, Kuhls-Gilcris A, Bednarek D, Rudin S. Instrumentation Noise Equivalent Exposure (INEE) for routine quality assurance: INEE measurements on a clinical flat panel detector. *Med Phys*. 2008; 35:3879.
27. Kuhls-Gilcris A, Jain A, Bednarek D, Rudin S. Instrumentation Noise Equivalent Exposure (INEE): An investigation of spatial frequency effects. *Med Phys*. 2008; 35:2636.
28. Mackenzie A, Honey ID. Characterization of noise sources for two generations of computed radiography systems using powder and crystalline photostimulable phosphors. *Med Phys*. 2007; 34:3345–3357. [PubMed: 17879798]
29. Texas Instruments, “TC285SPD-B0 1004 × 1002 Pixel Impactron CCD Image Sensor Manufacturer Data Sheet,” (2003).
30. Samei E. Image quality in two phosphor-based flat panel digital radiographic detectors. *Med Phys*. 2003; 30:1747–1757. [PubMed: 12906192]
31. Kuhls-Gilcris A, Jain A, Bednarek DR, Rudin S. Frequency Dependent Instrumentation Noise Equivalent Exposure (INEE). In Preparation.

Figure 1.

Schematic of the solid-state x-ray image intensifier (SSXII) module which includes a CsI(Tl) fiber-optic scintillating plate (FOS), a fiber-optic taper (FOT) and an EMCCD camera with fiber-optic input window. The design is extensible to an array of such modules to provide an increased field-of-view¹³. Note: the light tight enclosure has been omitted so that the inner components can be seen.

Figure 2.

A plot of the product of the measured camera gain constant and the EMCCD gain versus EMCCD gain, indicating a sensitivity of 15.0 ± 0.9 e-/DN at a gain of 1. The error bars represent the standard deviation between measurements taken at various LED intensities.

Figure 3.

X-ray sensitivity of two different system configurations (No FOT and 4:1 FOT), using the RQA 5 x-ray spectrum for three different trials, expressed in absolute units. The average fit given by the linear regressions is also presented. The slopes of the curves provide the x-ray sensitivity in units of electrons per incident x-ray photon. A similar plot provided the x-ray sensitivity for RQA 3.

Figure 4.

MTF of the various components that comprise the imaging chain of the SSXII prototype module. The MTF associated with the use of x-rays was plotted up to 5 cycles/mm, above which the measurement was unreliable. Such a detailed analysis of individual components could potentially lead to further design improvements as the imaging chain is better optimized. The error bars represent the standard deviation of three measurements.

Figure 5.

A comparison between the detector MTFs of the SSXII and two state-of-the-art FPDs, illustrating comparable performance at low spatial-frequencies (less than 4 cycles/mm) and the considerable improvement at high spatial-frequencies (i.e. beyond current Nyquist frequencies). The “SSXII – optimized” curve demonstrates, with further component optimization, substantial improvements may be realized at *all* spatial frequencies.

Figure 6.

A comparison plot of the detector DQEs of the SSXII and an FPD measured at a fluoroscopic exposure. The SSXII prototype is shown to substantially outperform the FPD at lower doses. With further component optimization, these improvements may be increased further.

Figure 7.

The 2D NPS measured with the Paxscan 2020+ is shown (left) measured at 22 μR . The NPS was then plot as a function of exposure at each spatial frequency u and v . Three examples are plot (right) at spatial frequencies (mm^{-1}) of (0.5, 0.5), (1, 1) and (2, 2) resulting in measured INEEs of 0.39 (0.138/0.356), 0.39 (0.098/0.254) and 4.1 μR (0.132/0.032) respectively, as described by equation 8.

Figure 8.
Comparison of the radially averaged frequency dependent INEE for the SSXII and the Paxscan 2020+.

Table 1

Various performance measures of the EMCCD camera. Sensitivity was measured as described in figure 3, read noise was determined from the dark field signal variance (16 DN rms at $1\times$ EMCCD gain) and the full well capacity was determined from the saturation signal (3300 DN at $1\times$ EMCCD gain).

Sensitivity (Camera Gain Constant, K) [e ⁻ /DN]	Read Noise [e ⁻ rms]	Read Noise @ 250× EMCCD Gain [e ⁻ rms]	Full Well Capacity [e ⁻]
15.0 ± 0.9	240 ± 20	0.96 ± 0.08	$49,000 \pm 3,000$