Published in final edited form as:

Proc SPIE Int Soc Opt Eng. 2013 March 6; 8668: 86680K-. doi:10.1117/12.2006451.

Intrinsic and total system performance evaluation for a newly developed Solid State X-ray Image Intensifier (SSXII) detector

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Abstract

The new Solid State X-ray Image Intensifier (SSXII) is a high-resolution, high-sensitivity, real-time region-of-interest (ROI) x-ray imaging detector. Evaluations were made of both standard linear systems metrics (MTF, DQE) and total system performance with generalized linear systems metrics (GMTF, GDQE) including scatter and geometric un-sharpness for simulated clinical conditions.

The SSXII is based on a $1k \times 1k$ EMCCD sensor coupled to a $300 \mu\text{m}$ thick CsI(Tl) phosphor through a 2.88:1 fiber optic taper resulting in a $37 \mu\text{m}$ effective pixel size and an effective $3.7 \text{ cm} \times 3.7 \text{ cm}$ square field-of-view (FOV). Standard methods were used to calculate MTF, NNPS and DQE. Generalized metrics were calculated and compared for three different magnifications (1.03, 1.11 and 1.2) and three different focal spots (0.3 mm, 0.5 mm and 0.8 mm) for a scatter fraction of 0.28.

For an RQA5 spectrum, at 5 cycles/mm the MTF was found to be 0.06 and DQE was 0.04, while the DQE(0) was 0.60. Focal spot un-sharpness and scatter significantly degrades the GMTF and GDQE performance of the detector. A low frequency drop is caused by scatter and is almost independent of focal spot size and magnification. The degradation for middle range frequencies is caused by geometric un-sharpness and increases with focal spot size and magnification. This degradation was least in the case of the small focal spot and almost independent of magnification.

In spite of this degradation, the high resolution SSXII with a small FOV may have a significant impact on ROI image-guided neuro-interventions since it demonstrates far better performance than standard current detectors.

Keywords

MTF; DQE; geometric un-sharpness; generalized metrics; image quality; relative detectability; system optimization

1. INTRODUCTION

Improved image-guided interventions for minimally invasive treatments of vascular diseases requires detectors with high resolution, high sensitivity and low noise. Current state-of-art

medical x-ray detector technologies including x-ray image intensifiers (XIIIs) and flat panel detectors (FPDs) have low resolution (approximately 3 line pairs per mm) and have inherent problems which limit their imaging performances. XIIIs suffer from veiling glare and distortion¹, while FPDs suffer from lag, ghosting and excessive instrumentation noise^{2, 3}. In the past, our group originated the idea of region-of-interest (ROI) imaging^{4–6} which is essentially getting a very high quality image where the intervention is taking place in a part of the full field of view (FOV) of larger detectors (XIIIs, FPDs). With a reduction in x-ray intensity in the periphery, it also reduces radiation dose to patients. A detector having a small FOV with excellent spatial resolution can serve the purpose of ROI imaging. Higher resolution and lower noise detectors for real time x-ray imaging are essential for endovascular image guided interventions (EIGIs) for accurate guidance of devices through the vascular system⁷. Detectors with such capabilities improve the effectiveness of minimally invasive interventional procedures while minimizing radiation dose to patients^{8–10}. Our group has developed a new high resolution, high sensitivity region-of-interest (ROI) Solid State X-ray Image Intensifier (SSXII) from the component level which can provide real-time, high spatial-resolution, and low noise (hence, high dynamic range) images even at low exposures because of the high variable on-chip gain of the electron-multiplying CCD (EMCCD)¹¹.

The linear system metrics modulation transfer function (MTF), normalized noise power spectrum (NNPS), noise equivalent quanta (NEQ) and detective quantum efficiency (DQE) were experimentally determined for intrinsic detector performance evaluation. The linear metrics are measured as a function of the spatial frequency at the input image plane of the detector. To evaluate imaging systems in real clinical environments, linear metrics must include the effects of focal spot un-sharpness, magnification and scatter and be referenced to the object plane^{12–15}. The Generalized Linear metrics designated as Generalized Modulation transfer Function (GMTF), Generalized Normalized Noise Power Spectrum (GNNPS), Generalized Noise Equivalent Quanta (GNEQ) and Generalized Detection Quantum Efficiency (GDQE) take these factors into account to evaluate the total system performance. The Generalized Metrics are measured as a function of the spatial frequency at the object plane and include the effects of focal spot un-sharpness, magnification and scatter, hence give total system performance. These generalized metrics were calculated for the new SSXII detectors by taking system parameters into account which mimic the real clinical situation. The generalized metrics show that there are tradeoffs in choice of different system parameters. More importantly, generalized metrics can be used to determine detectability of various interventional devices^{16, 17} which can guide optimization of the performance of the total imaging system¹⁸.

2. METHODS AND MATERIALS

The SSXII is based on a 13 μ m pixel, 1k $\times$ 1k EMCCD sensor (model CCD201-20, e2v Technologies Ltd, UK) connected through a fiber optic taper (FOT) to a 300 μ m thick CsI(Tl) (HL) structured input phosphor grown on a fiber optic plate (FOP). The FOT with taper ratio 2.88:1 gives an effective pixel size of 37 μ m and an effective field-of-view (FOV) of 3.7 cm diameter. All images were taken at RQA5 spectrum to calculate linear system metrics for the SSXII. The individual frames were corrected for flat field and offset, and the

Fourier transform method was used to calculate detector normalized noise power spectrum (NNPS_D)¹⁹. The detector modulation transfer function (MTFD) was measured by the slanted edge method²⁰. The detector detective quantum efficiency (DQE_D) was computed using MTF_D and NNPS_D.

A formulation [shown in equations (1), (2), (3) and (4)] developed by Kyprianou et. al¹⁸ was used to carry out the generalized analysis.

$$GMTF(f, p, m) = [(1 - \rho)MTF_F((\frac{m-1}{m})f) + \rho MTF_S(\frac{f}{m})]MTF_D(\frac{f}{m}) \quad (1)$$

$$GNNPS(f, X, m) = \frac{NNPS_D(\frac{f}{m}, X)}{m^2} \quad (2)$$

$$GNEQ(f, \rho, X, m) = \frac{GMTF^2(f, \rho, m)}{GNNPS(f, X, m)} \quad (3)$$

$$GDQE(f, \rho, X, m) = \frac{GNEQ(f, \rho, X, m)}{m^2 \phi_{in}} = \frac{GMTF^2(f, \rho, m)}{m^2 \phi_{in} GNNPS(f, X, m)} \quad (4)$$

where MTF_F and MTF_S are focal spot and scatter MTF ϕ_{in} respectively. is the fluence at the detector entrance, ρ is the scatter fraction, X is the exposure, m is the magnification for the plane of the object and f is the spatial frequency in that plane.

The MTF_S was introduced to account for scatter blur and defined as the frequency content of the spatial distribution of scatter entering the detector^{14, 15}. The MTF_F accounts for the geometric un-sharpness (focal spot blur) due to the x-ray intensity distribution of the focal spot (focal spot distribution) and is defined as the modulus of the Fourier transformation of a slit image of the focal spot¹³. The x-ray beam entering the detector is composed of both primary and scatter x-rays whose distribution is determined by focal spot and scatter blur respectively. Further, both components of the x-ray beam entering the detector are filtered by the detector MTF_D. Finally, the magnification is used to rescale the frequencies in the image plane to define them in the object plane¹⁷.

To calculate the generalized linear metrics for the SSXII, data for MTF_F for three different focal spots [small (0.3 mm), medium (0.5 mm) and large(0.8 mm)], MTF_S and scatter fraction ($\rho = 0.28$ for 5.0 cm air gap and $\rho = 0.33$ for 2.5 cm air gap between the phantom and the detector) were used. These values are obtained from a previous analysis²¹ done for the micro-angiographic fluoroscope (MAF)²² which has a similar front end including phosphor, FOV and effective pixel size to the SSXII. The generalized metrics were calculated and compared for three different magnifications (1.03, 1.11 and 1.2 which represents three different planes in the phantom) and three different focal spots for a fixed value of scatter fraction.

3. RESULTS

The MTF_D and DQE_D of the intrinsic detector are shown up to 8 cycles/mm in figures 1 and 2. The MTF_D was found to be 0.06 at 5 cycles/mm and the DQE_D was 0.04. The MTF_D is phosphor limited since the Nyquist frequency resulting from the effective pixel size ($37\mu\text{m}$) is 13.5 cycles/mm. The measured zero frequency $DQE_D(0)$ was 0.57 for an RQA 5 spectrum. The calculated NNPS and their product with exposure for several exposures corresponding to low fluoroscopic range are shown in figures 3(a) and 3(b).

The GMTF and GDQE calculated for medium focal spot for a magnification value of 1.11 and scatter fraction 0.28 are compared with the MTF_D and DQE_D of the intrinsic detector in (figures 4(a) and 4(b)). These figures show how the linear metrics of the intrinsic detector degrade when scatter and geometric un-sharpness are taken into account.

The $GMTF_S$ and $GDQE_S$ were calculated for different focal spots for a magnification value of 1.11 and scatter fraction of 0.28 to demonstrate the effect of focal spot sizes for a given magnification and are shown in (figures 5(a) and 5(b)). To demonstrate the effect of magnification for a particular focal spot, the GMTF and GDQE were calculated for three different magnifications for each focal spot for a scatter fraction of 0.28 and are shown in figures 6 7 and 8.

The GMTF and GDQE were also calculated for the small focal spot for magnification of 1.2 for a scatter fraction of 0.33. The GMTF and GDQE for two different scatter fractions were compared for a magnification of 1.2 for the small focal spot (in figures 9(a) and 9(b)) to demonstrate the effect of scatter fraction for the same magnification.

4. DISCUSSION

The same form of the NNPS for a range of low exposures (figure 3(a)) demonstrates that the SSXII appears to remain quantum limited even at exposures as low as $1.32\mu\text{R}$ due to its large on-chip variable gain. Figure 3(b) indicates that the DEQ is independent of exposure as the product of NNPS with respective exposures remains almost the same even at such a low exposure. The large dynamic range of the SSXII owing to lower noise level and on-chip variable gain together with its high resolution capability makes it a high resolution dynamic x-ray detector to be used both for fluoroscopic and angiographic applications¹¹. Moreover, it may also be possible with the SSXII to provide superior quality images for a FOV comparable to current state-of-the-art detectors (FPD_S and XII_S) because the SSXII can be extended in two dimensions to any arbitrary size to provide a larger FOV²³.

The MTF_D of the intrinsic detector simply gives resolution at the detector plane while the GMTF gives the system resolution relative to the object plane by taking the effect of scatter and geometric un-sharpness into account and degrades intrinsic detector metrics (MTF_D and DQE_D) (figure 4(a)). The results also show that the effect of scatter and geometric un-sharpness on the GDQE is similar to that on the GMTF (figure 4(b)). The initial drop at low frequencies is caused by scatter and is not much affected either by the focal spot sizes or by the magnification values (figures 5(a) to 8(b)). An increase in scatter for the same magnification worsens the degradation for low frequencies while affecting middle range

frequencies and high range frequencies a little (figures 9(a) and 9(b)). The degradation in middle frequencies and the high range frequencies is mainly due to geometric un-sharpness, which is affected both by the focal spot size and by the magnification value. This degradation becomes worse by increasing the focal spot size for a fixed magnification shown in figures 5(a) and 5(b), therefore, the choice of focal spot size plays a crucial role in determining the system performance especially at high frequencies. Similarly, the degradation is enhanced by increasing magnification for middle and large focal spots (figures 7(a) to 8(b)).

The generalized metrics help us understand the dependence of system performance on various parameters which affect image quality as there are always tradeoffs in selecting system parameters to optimize the system performance. For instance, the air gap between the patient and the detector decides what compromise should be made between geometrical un-sharpness (affects high frequencies) and scatter (affects low frequencies).

The geometric magnification may be used to improve image quality to see finer details for the proper selection of the combination of focal spot size and magnification. The GMTF provides quantitative information to evaluate this improvement by referencing the spatial resolution to the object plane. Figure 10 shows the GMTF of selected frequencies with magnification for medium focal spot which provides an alternative way to look at the GMTF. The GMTF improves for lower spatial frequency (1 to 3 cycles/mm) up to a particular value of magnification before it starts degrading due to increasing geometrical un-sharpness (caused either by large focal spot size or higher magnification values) which overshadows the advantage of getting better GMTF (better spatial resolution) due to magnification. The improvement in the GMTF can be seen further up into the middle range frequencies (5 cycles/mm) in the case of the small focal spot (figure 11).

The GNEQ is the absolute measure of the image quality (higher GNEQ means images with higher signal to noise ratios) and the GDQE gives comprehensive performance of the system by combining the GMTF, GNNPS and number of x-ray photons referenced to the object plane. The interpretation of generalized metrics is not trivial and requires caution and this is illustrated by an example. Let the air gap between the phantom and the detector be reduced to 2.5 cm (position 'B') from 5 cm (position 'A') (changing the scatter fraction to 0.33 from 0.28) changing the magnification to 1.16 from 1.2 of the plane containing an object of interest. The GMTF and the GDQE corresponding to these two situations (A and B) are shown in figure 12. The situation 'A' would be more favorable for a low frequency content object compared to 'B' which would be more favorable for a high frequency object regarding its detectability or both would give the same detectability if the object has such a mix of both frequency content.

Although generalized metrics take the clinically relevant parameters into account, still they cannot serve as perfect figures of merit for the imaging system on their own because they do not indicate the detectability of objects of interest in clinical tasks. The GNEQ can be used to predict the detectability¹⁶ of the objects of interests in clinical tasks and can also be used to optimize¹⁸ the performance of the imaging systems.

5. CONCLUSIONS

The high spatial resolution (>3 cycles/mm) and the small FOV make the SSXII capable of playing a critical role for ROI endovascular image guided interventions when used in conjunction with large FOV detectors (FPDs and XRIIs). It can be concluded from the overlapping of graphs in figures 6(a) and 6(b) that the degradation suffered by system metrics is almost independent of magnification in the case of the small focal spot for those magnifications considered; therefore, the SSXII can be used to overcome the issue of image degradation due to unequal blurring caused by differential magnification when a somewhat thick object is imaged in the clinical situation.

The MTF_D and DQE_D of the intrinsic detector are used as metrics to evaluate the performance of the detectors but they lose their relevance when it comes to evaluate the performance of the total imaging system. The GMTF and GDQE take into account the clinically relevant parameters (scatter, focal spot un-sharpness and magnification) and can give a more comprehensive description of the system and can be useful to evaluate and compare different systems for the same sets of parameters, in a similar manner that the MTF_D and DQE_D are used to compare different detector performances²¹. However they do need careful interpretation because the two sets of values of magnification and scatter may provide the same relative detectability for an object for a particular focal spot while the corresponding generalized plots would be different.

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Figure 1.
The detector MTF_D

Figure 2.
The detector DQE_D

Figure 3.

(a) The detector NNPS_D at different input exposures. (b) the product of NNPS_D with the exposure at different exposures.

Figure 4.

(a) the GMTF for the medium focal spot for a magnification of 1.11 and scatter fraction of 0.28 compared to the detector MTF_D . (b) the GDQE for the medium focal spot for the same magnification and scatter fraction as in (a) compared to the detector DQE_D .

Figure 5.

(a) GMTFs for different focal spots for a magnification of 1.11 and scatter fraction of 0.28.

(b) GDQEs for different focal spots for the same magnification and scatter fraction as in (a).

Figure 6.

(a) GMTFs for different magnifications for the small focal spot and scatter fraction of 0.28.

(b) GDQEs for different magnifications for the same focal spot and scatter fraction as in (a).

Figure 7.

(a) GMTFs for different magnifications for the medium focal spot and scatter fraction of 0.28. (b) GDQEs for different magnifications for the same focal spot and scatter fraction as in (a).

Figure 8.

(a) GMTFs for different magnifications for the large focal spot and scatter fraction of 0.28.

(b) GDQEs for different magnifications for the same focal spot and scatter fraction as in (a).

Figure 9.

- (a) GMTF for different scatter fractions for a magnification of 1.2 for the small focal spot.
 (b) GDQE for different scatter fractions for the same magnification and focal spot as in (a).

Figure 10.
GMTF of selected frequencies for three different magnification values for the medium focal spot.

Figure 11.
GMTF of selected frequencies for three different magnification values for the small focal spot.

Figure 12.

The GMTF and the GDQE corresponding to different sets of magnifications and scatter fractions for the same plane in the phantom for the medium focal spot.