Evaluation of the microangiographic fluoroscope (MAF) using generalized system performance metrics

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(Received 4 September 2012; revised 1 February 2013; accepted for publication 1 February 2013; published 28 February 2013)

Purpose: The performance of a newly developed, high resolution, microangiographic fluoroscope (MAF) (35 μm pixel pitch and 300 μm thick CsI phosphor) was evaluated using a generalized linear system analysis and compared with that of a standard amorphous Si thin film transistor flat panel detector (FPD) (194 μm pixel pitch and 600 μm thick CsI phosphor). The linear system metrics such as modulation transfer function (MTF), noise power spectrum, and detection quantum efficiency (DQE) are commonly used to gauge the intrinsic detector performance in the detector plane. However, these linear system metrics do not provide information about the image receptor performance in a real system since they do not include the effects of other parameters such as focal spot distribution, scatter radiation, and geometric unsharpness, which may compromise detector performance characteristics. Use of generalized linear system metrics [generalized modulation transfer function (GMTF), generalized normalized noise power spectrum (GNNPS), and generalized detection quantum efficiency (GDQE)] that include these effects gives a more meaningful, complete, and appropriate evaluation of detector performance as part of the imaging system.

Methods: A uniform head equivalent phantom was used to simulate realistic clinical parameters and x-ray spectra. The detector MTFs were measured using the slanted edge method and the focal spot MTFs were measured using a pinhole assembly. The scatter MTF was simulated and the scatter fraction was measured for a head-equivalent phantom. The generalized system metrics were calculated for different combinations of three choices of focal spots and three different magnifications with two different air-gaps. The performance of the MAF was also illustrated using stent images obtained with different focal spots under similar conditions.

Results: Results for the generalized metrics provide a quantitative description of the performance of the imaging system for both detectors. This generalized analysis demonstrated that both detectors have similar imaging capabilities at lower spatial frequencies, but that the MAF has superior performance over the FPD at higher frequencies even when considering focal spot blurring and scatter.

Conclusions: This generalized performance analysis demonstrates the significance of focal spot size, magnification, and scatter on the system performance metrics (GMTF, GNNPS, and GDQE). Although the ideal detector performance characteristics of the MAF are not fully realized due to these other system factors, it still retains an advantage in DQE at high spatial frequencies over the FPD. Similar studies based on the generalized linear system metrics can serve as an efficient tool to evaluate total system capabilities under different realistic conditions to enable optimal design for specific imaging tasks. © 2013 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4792460>]

Key words: MAF, microangiography, CCD detector, MTF, DQE, GMTF, GDQE

I. INTRODUCTION

The performance of any linear x-ray detector can be characterized by the linear system analysis metrics¹ of modulation

transfer function (MTF), normalized noise power spectrum (NNPS), noise equivalent quanta (NEQ), and detective quantum efficiency (DQE). MTF describes the detector's spatial frequency response, NPS is a representation of image noise

in the frequency domain, NEQ gives the effective number of quanta that were used for the image formation, and DQE demonstrates the fractional transfer of signal to noise squared from input stage to final stage, which is also the fraction of input quanta used for image formation. Each of these metrics has its own significance in characterizing detector performance. These linear system metrics do not generally account for other system parameters and hence they gauge only the intrinsic detector performance. The method of determining such metrics is defined in such a way that the effect of other system parameters is eliminated.²⁻⁴ For example, when the MTF of a digital detector is measured, the slanted edge or slit is placed directly against the detector face to minimize the effect of magnification and focal spot.^{2,4} However, in patient imaging, this does not happen. There is always a nonzero gap between the detector and the object of interest being imaged and that introduces geometric unsharpness because of the finite focal spot size. The true measure of the imaging capability of any detector depends on its performance in realistic conditions. The total system performance is the combined effect of all factors which affect image quality including the detector, the finite focal spot size, the magnification used, and scatter from the object so that the total system performance may be very different from the intrinsic detector performance.

The effect on the detector performance of system parameters has been studied extensively by others⁵⁻⁸ including Kyprianou *et al.*^{9,10} who introduced generalized linear system 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). In this study, we will evaluate the performance of our newly developed microangiographic fluoroscope (MAF) (Ref. 11) within a system similar to the clinical situation and compare it to the performance of a standard flat panel detector (FPD) using the generalized linear system metrics. During initial clinical tests, the MAF has demonstrated the potential to fulfill a need in neurovascular imaging for substantially higher resolution.^{12,13} It is therefore, important to evaluate and objectively compare both detectors as part of a total imaging system to determine quantitatively how much image quality improvement might be realized with the MAF as well as what compromises might be required.

II. METHODS AND MATERIALS

II.A. Formulation of generalized linear system metrics

The formalism of Kyprianou *et al.*^{9,10} is used for the calculation of generalized metrics for the system comparisons given here. In this formalism, the generalized modulation transfer function is defined in the object plane as^{9,10}

$$\text{GMTF}(u, \rho, m) = [(1 - \rho) \text{MTF}_F\left(\frac{m-1}{m}u\right) + \rho \text{MTF}_S\left(\frac{u}{m}\right)] \text{MTF}_D\left(\frac{u}{m}\right), \quad (1)$$

where u is spatial frequency in the object plane and m is the magnification of a feature in the object plane onto the detector plane, ρ is the scatter fraction, and $\text{MTF}_D(u)$, $\text{MTF}_F(u)$ and $\text{MTF}_S(u)$ are the modulation transfer functions of the detector, the focal spot and scatter, respectively. The GNNPS is defined in the object plane as^{9,10}

$$\text{GNNPS}(u, X, m) = \frac{\text{NNPS}_D\left(\frac{u}{m}, X\right)}{m^2}, \quad (2)$$

where $\text{NNPS}_D(u, X)$ is the normalized noise power spectrum for the detector at the detector plane and X is the incident detector exposure. The total exposure that reaches the detector and contributes to its signal is the sum of the scatter and the primary and, for the NNPS calculation, there is no distinction between primary and the scatter in terms of noise contribution. Using the GMTF and the GNNPS, the GDQE is defined as^{9,10}

$$\text{GDQE}(u, \rho, X, m) = \frac{\text{GMTF}^2(u, \rho, m)}{m^2 \phi_{in}(X, m) \text{GNNPS}(u, \rho, X, m)}, \quad (3)$$

where $\phi_{in}(X, m)$ is the number of quanta per unit area at the detector entrance increased by m^2 to the object plane.

II.B. Experimental setup

Our group developed the concept of region of interest (ROI) imaging¹⁴ to be used when high quality images are needed in only a part of the full field-of-view and we have been developing CCD-based region of interest detectors for over a decade. The MAF detector used for this study is the latest and most advanced outcome of our endeavor. The MAF (Ref. 11) has a CCD camera coupled to a GEN 2 dual MCP light image intensifier (LII) (Ref. 15) through a fiber-optic taper resulting in 35 μm pixels at the detector input plane. A CsI scintillator (300 μm thick HL type) is attached to the front end of the LII with fiber-optic coupling (Fig. 1). The key component in the design of the MAF is the LII which provides very high gain that enables the MAF to image even at very low exposures per frame. The MAF was developed as a high resolution region of interest x-ray detector and it has a field of view (FOV) of 3.5 $\times$ 3.5 cm. No antiscatter grid was used for the MAF because of the small scatter fraction resulting from its small FOV. In a previous work, we reported on a custom gold septa based grid, which we built specifically for the MAF, and found that the reduction in primary did not justify the small reduction in the already small amount of scatter resulting from the small FOV of this ROI detector.¹⁶ More details about the MAF can be found in Ref. 11. The other detector used for the comparison study is a standard amorphous Si thin film transistor (TFT) fluoroscopic FPD (Varian PaxScan 2020 with 194 μm pixel size and 600 μm thick CsI phosphor). The detector size of the FPD is 20 $\times$ 20 cm and, to provide a balanced comparison of the detector as it might be used for ROI imaging, the field of view was collimated to the size of the MAF and the antiscatter grid was removed in this study. The MAF was deployed on the C-arm gantry of an interventional fluoroscopic system (Infinix, Toshiba Medical Systems

FIG. 1. Schematic of MAF showing locations of fiber-optic plates (FOP), fiber-optic taper, light image intensifier (LII), and CsI(Tl) input phosphor.

Corporation) using a custom detector changer (Fig. 2), which allows interchangeable use of the MAF or FPD. The x-ray tube has three different focal spots: small (0.3 mm), medium (0.5 mm), and large (0.8 mm). The high resolution detector was deployed using the detector changer on the C-arm gantry at 100 cm from the x-ray tube. A uniform head phantom was used for the experiment to simulate clinical conditions of beam hardening and scatter. A thickness of 150 mm PMMA and 3 mm Al was used to construct the head equivalent phantom (AAPM Report No. 60) (Ref. 17) and it was placed on the patient table as shown in Fig. 3 for measurements of the scatter fraction. Different magnifications define different object plane locations. We used three different magnifications (1.03, 1.11, and 1.20) and two different air-gaps between the phantom and the detector (2.5 and 5 cm). For the 2.5 cm air-gap and a SID of 100 cm, the magnification values of 1.03, 1.11, and 1.20 represent upper, middle, and bottom planes of the simulated head phantom, respectively.

II.C. Measurements

The scatter fraction for the experimental setup was measured using the standard lead beam stop technique.¹⁸ Lead discs with 2 mm thickness and different diameters were placed between the table and head equivalent phantom blocking the primary radiation. For each disc diameter, 20 images were acquired with the MAF. The images were corrected for

offset and flat field and averaged. The gray values at the center of the disc images and its nearest pixels were averaged to obtain the scatter behind the disc. The gray values at the same pixels without a lead disc correspond to primary and scatter. The scatter fraction was calculated for each diameter and plotted against the disc diameter. The scatter fraction was estimated by extrapolating the curve to zero-diameter.

The detector NNPS was calculated using a standard Fourier transform method³ from flat field images obtained with a RQA5 spectrum (74 kVp with 21 mm of Al added filtration, HVL 7.1 mm of Al). The NNPS for both the MAF and the FPD was calculated from 60 offset and flat-field corrected images. Six different exposures were used to measure the NNPS for both detectors. In our clinical evaluations,¹⁹ the MAF was used in what we term high definition (HD) mode²⁰ (where low DA mode exposures were used to acquire images at fluoro rates) which resulted in an input exposure in the range of 40 micro-R per frame at 15 frames per second. This was assessed to provide the proper level of image quality for the neurointerventions, although no study was performed to find the optimal level (which may be task specific). This same level was used for the GDQE determination for the MAF and the FPD to evaluate the performance of the MAF as used and to provide a balanced comparison with the FPD. The antiscatter grid was removed from the FPD for all the experimental data acquisitions since the same small collimated beam was used.

The slanted edge method² was used to measure the detector MTFs with the RQA5 spectrum. To obtain the focal spot MTF, a 10 μ m pinhole was used to take the focal spot images for a magnification value of 3.12 with the MAF. The focal spot images were rescaled for unity focal spot magnification and the Fourier transforms of those focal spot images were taken to get the focal spot MTFs.

The scatter MTF used for this study was simulated using the method given by Boone *et al.*²¹ in which the scatter PSF was assumed to be Gaussian with a radial expansion term of $1/r$. The scatter PSF was simulated for the 15 cm PMMA head equivalent phantom and the Hankel transform was used to obtain the scatter MTF.

III. RESULTS

A plot of scatter fraction vs disc diameter is shown in Fig. 4 for both air-gaps. The scatter fractions extrapolated to zero disc diameter were 0.33 and 0.29 for the 2.5 and 5.0 cm air-gaps, respectively.

The measured MTFs for the three available focal spots are shown in Fig. 5. The Fourier transform of the focal spot image obtained with the pinhole gave the 2D MTF and the radial average of that 2D MTF is shown in Fig. 5. The scatter MTF with reference to the detector plane is also shown in Fig. 5. The scatter MTF has only low frequency content and falls rapidly as spatial frequency increases.

The measured MTFs for both detectors are shown in Fig. 6. The measured NNPSs for the MAF and the FPD are shown in Figs. 7(a) and 7(b), respectively. The DQEs calculated using the MTFs and NNPSs for both detectors are shown in Fig. 8.

FIG. 2. MAF in deployed position in front of a FPD with a custom-made detector changer on a C-arm gantry to facilitate switching between detectors.

The DQE for the MAF and the FPD was calculated using a detector exposure of 39.3 and 44.6 μR , respectively.

The GMTFs and GDQEs were calculated for various combinations of focal spot size, magnification, and scatter fractions. Figures 9 and 10 show the effect of focal spot choice on the GMTF and GDQE of the MAF for a fixed magnification (1.20) without scatter. The GMTF and GDQE of the two detectors are compared for small and large focal spots with the worst case selected scatter fraction of 0.33 and a magnification value of 1.20 in Figs. 11 and 12, respectively. For the effect of magnification on the system performance, the GMTF and GDE were calculated with the three magnifications as

specified in Sec. II and results showed that increasing magnification degraded system performance for the medium and large focal spot but provided a small improvement when the small focal spot was used.³² In general, the small focal spot is needed to at least retain the performance advantages of the MAF at typical magnification factors. Note that in this comparison the same scatter fraction was used for both detectors assuming that the FOV of the FPD was collimated to the size of the FOV of the MAF. Figures 13 and 14 show a comparison of the MAF performance with different focal spot choices. A stainless steel coronary stent with 100 μm struts was imaged with the MAF in similar conditions (magnification 1.20 and no scatter) for three different focal spots with the same x-ray technique parameters using the RQA5 spectrum ($\sim 40 \mu\text{R}$ per

FIG. 3. Experimental setup showing location of head phantom for scatter measurements.

FIG. 4. Measured scatter fraction as a function of lead disc size for two different air-gaps. The intercept on the y-axis represents the scatter fraction without the perturbation of the disc.

FIG. 5. Scatter and focal spot MTFs.

frame detector entrance exposure). Ten images for each focal spot were averaged to reduce the noise and highlight the resolution differences. The averaged images of the stent are shown in Figs. 13 and 14 shows the line profiles through the stent images normalized to background values for each focal spot (the locations of the line profile data are shown by lines with arrows in Fig. 13). Seven different peaks were identified in the line profiles shown in Fig. 14 and the signals of individual peaks and corresponding troughs were used to calculate contrast $[(\text{Peak} - \text{Trough})/(\text{Peak} + \text{Trough})]$. The average improvement in contrast of the struts for the small focal spot over the large focal spot was 42%.

IV. DISCUSSION

The modeled scatter MTF contains only very low frequency components and hence provides a steep drop at low frequency for the GMTF and GDQE as shown in the figures. The larger air-gap reduces the scatter and hence the low frequency drop is reduced. However, a larger air-gap provides greater magnification at the same SID and the effect of the

FIG. 6. Measured MTF for the MAF and the FPD.

FIG. 7. Measured NNPS for the MAF (a) and the FPD (b) for different detector exposures.

focal spot size increases with increasing magnification. It is clear that the choice of the air-gap (scatter fraction) and the magnification (focal spot effect) are connected to each other in such a way that one factor affects the other. The choice of the air-gap and magnification for the optimum system performance depends on the focal spot size and amount of scatter present. The MTFs for different focal spots demonstrate a strong dependence on the focal spot size such that the MTF degrades as size increases as shown in Fig. 5.

The resolution for both detectors is compared in Fig. 6, where the MTFs for the MAF and the FPD are plotted together. The detectors have different pixel size and thus they have different Nyquist frequency. The FPD has a Nyquist frequency of about 2.5 cycles/mm (194 μm pixel) while that of the MAF is 14.2 cycles/mm (35 μm pixels); however, we have plotted the MTF curve for the MAF only up to 10 cycles/mm. Both detectors have similar MTFs up to 1.5 cycles/mm, while the MAF shows better MTF at frequencies above. The MAF has a smaller pixel size and thinner phosphor and is capable of imaging higher spatial frequencies. Figures 7(a) and 7(b) show the measured NNPS for six different entrance exposures over approximately the same exposure range for the MAF and the FPD. The MAF is seen to have improved NNPS over the

FIG. 8. Calculated DQE for the MAF and the FPD for exposures of 39.3 and 44.5 μR , respectively.

FIG. 11. GMTF comparisons for the MAF and the FPD. Small and large focal spot with a scatter fraction of 0.33 and a magnification factor of 1.20.

FIG. 9. GMTFs for the MAF showing the effect of the choice of the focal spot size. Three different focal spots were used with a magnification factor of 1.20 and zero scatter fraction.

FIG. 10. GDQEs for the MAF showing the effect of the choice of the focal spot size. Three different focal spots with a magnification factor of 1.20 and zero scatter fraction.

FIG. 12. GDQE comparisons for the MAF and the FPD. Small and large focal spot with a scatter fraction of 0.33 and a magnification factor of 1.20.

FPD at corresponding exposures. The DQE comparison for the MAF and the FPD in Fig. 8 shows similar trends as the MTF comparison. The only noticeable difference is that at lower spatial frequencies, the FPD has a slightly better DQE because of a thicker CsI phosphor (600 μm for the FPD and 300 μm for the MAF). We could improve the low frequency MAF DQE by going to a thicker phosphor for increased QDE and preserving the high frequency response by using HR type CsI phosphor which does not use a front reflective surface as used in the HL type.

The generalized evaluation for the MAF starts from Fig. 11. Figures 11 and 12 show the effect of focal spot choice on the MAF system performance (in terms of GMTF and GDQE) for a fixed magnification (1.20) and scatter fraction (0.33). The performance of the MAF is adversely affected with increasing focal spot as indicated by the GMTF and the GDQE. The focal spot choice has a greater effect on the GDQE since, for the small focal spot, we have nonzero GDQE

at least up to 7.0 cycles/mm, but for the large focal spot, it barely reaches 3.5 cycles/mm.

The effect of magnification on detector performance will be similar to the effect of focal spot size. With increasing magnification the focal spot blur will increase and will affect the GMTF significantly. The GDQE is affected similarly but the effect of magnification is more pronounced because of the presence of the GMTF squared term in the GDQE.

The presence of scatter affects detector performance at low spatial frequencies as shown in Figs. 11 and 12. Because of the shape of the scatter MTF, the scatter fraction causes a sudden, proportional drop in GMTF and GDQE at low spatial frequencies. The magnitude of the low frequency drop depends on the value of the scatter fraction. A larger scatter fraction will give rise to a greater drop. For this study, we collimated the beam to the field of view of the MAF (3.5×3.5 cm) as used in region-of-interest imaging and we used the same scatter fractions for both detectors. If the full 20×20 cm field of view of the FPD were used, the scatter fraction would be much larger. Because of this larger scatter fraction, an antiscatter grid is usually an integral part of the system. Use of an antiscatter grid decreases the scatter fraction at the expense of increased patient dose and reduced primary detection.

An interesting comparison is shown in Fig. 11 between the GMTF of the FPD and the MAF for small and large focal spots (magnification value 1.20 and scatter fraction 0.33). Clearly, both the MAF and FPD suffer a low frequency drop because of scatter but the GMTF values of the MAF which are higher at much higher spatial frequencies show that the MAF will exhibit improved resolution under the same clinical conditions. It is interesting to note that the choice of the focal spot has a smaller effect on the GMTF of the FPD system in comparison to the MAF system. The GDQE comparison follows similar trends as the GMTF comparison.

For this study, the generalized performance of two detectors with different pixel size and different CsI thickness were compared. Although, it may be intuitive that the detector with a smaller pixel size would have better spatial resolution and the detector with the thicker CsI phosphor would have greater zero frequency DQE, a closer look at the DQE comparison indicates some not so obvious features. Both detectors were compared in terms of frequency dependent DQE at nearly the same exposure and both detectors showed comparable DQE up until the Nyquist of the FPD. The FPD does not show a very much larger DQE because of the very high additive noise in comparison to the MAF. Here, because of a larger pixel size, the FPD collects a lot more x-ray quanta per pixel than does the MAF and hence the relative quantum noise associated with the x-ray quanta is lower for the FPD but the high additive noise severely degrades its performance and would preclude having a FPD with as small a pixel size and thin a phosphor as the MAF. If we were to have 6×6 binning for the MAF, the pixel size would be similar to the FPDs and both detectors would collect the same number of x-ray quanta and would have the same quantum noise. Both detectors would perform similarly when quantum noise is dominant but the MAF would be far superior at low exposures because of its very small additive noise. This point has a bearing in choos-

FIG. 13. Images of a coronary stent with 100 μm struts were taken with the MAF using identical techniques parameters (RQA 5 spectrum with about 40 μR per frame detector entrance exposure and geometric magnification of 1.2) with three different focal spots. The arrow in each image shows the location for the line profiles shown in Fig. 14.

ing the clinical technique parameters for the MAF for clinical testing. When we compare DQE's (i.e., the ratio of the output and input SNR^2) for all frequencies up to the FPD's Nyquist frequency, they appear similar up to approximately 1.5 cycles/mm and after that the MAF has a better DQE; however, when we consider the classical SNR^2 [i.e., the ratio of

FIG. 14. Comparison of line profiles taken from the same locations in the stent images for the small and large focal spots shown in Fig. 13. (The location of the line profiles is shown by the lines with arrows in Fig. 13.) A total of seven different peaks indicated by the arrows in this figure were identified to calculate contrast.

the square of the zero frequency signal (mean) divided by the noise summed over all the frequencies (variance)] the FPD will indicate a higher SNR^2 because of the larger pixel size. As a result, we would expect to have to increase the exposure for the MAF or use temporal image filtering so as to match the final output SNR^2 values.^{19,22} Although we may need to increase the exposure for the MAF, the patient's effective dose should be less than that for the full FOV FPD because of the much smaller MAF field of view.²³ If we have a detector with pixel size and CsI similar to the MAF and additive noise similar to the FPD, its performance would be drastically affected by the high additive noise present in the imaging chain. Although direct flat panel imagers have better spatial resolution than indirect, they would not be an ideal choice for a microangiographic fluoroscope since they are often limited by high additive noise, ghosting, and lack of real time imaging capability.²⁴

Although important in many instances, we did not include the effect of motion blur in our generalized metrics because the MAF has been designed for neurovascular applications where motion blur is usually not significant. However, for other imaging tasks, the adverse effect of motion blur may have to be included in a generalized system performance evaluation and such blur might negate the resolution advantage of the MAF. Furthermore, high resolution images such as obtained with the MAF could present a structured noise problem if a radiographic grid were used. However, the MAF has been currently designed with a small FOV for ROI imaging and, fortunately, high quality clinical images have been obtained without the need for a grid. If a larger x-ray beam size were to be used, a grid might become necessary and careful consideration would then have to be given to structured noise. Grid structure noise is generally not an issue for the large detector elements of the FPD.²⁵

The MAF and FPD have previously been compared in imaging a stent and an aneurysm coil under the same con-

ditions and the much better quality of the MAF images was clearly evident.^{11,26} In this study, we demonstrate the effect of the focal spot size on image quality by imaging a stent with the MAF with identical techniques for three different focal spots. The effect of focal spot size can be appreciated visually in the stent images of Fig. 13. To evaluate the effect quantitatively, a comparison of line profiles taken at the same locations in the stent images for the small and large focal spots is shown in Fig. 14. The peaks in the line profile for the small focal spot are more prominent and clearer. The line profile for the large focal spot shows a significant loss of contrast and resolution over that of the small focal spot (See the supplementary material).³²

V. SUMMARY AND CONCLUSION

In this study, we evaluated the total system performance of the MAF and compared it with that of a TFT FPD using generalized metrics. The GMTF and the GDQE provide a characterization of image quality that includes the effect of the patient (magnification and scatter) and x-ray tube (focal spot distribution) as well as that of the image receptor. As shown in the results, the scatter MTF has only a low frequency component and it is clearly demonstrated that the presence of scatter produces a sharp low spatial frequency drop in both the GMTF and GDQE. The focal spot unsharpness affects the GMTF and GDQE at higher spatial frequencies and the effect of focal spot blurring increases with magnification. The comparison of the system performance between the MAF and the FPD showed that both systems have similar capabilities up to 1.5 cycles/mm but that the MAF has substantially better performance at higher spatial frequencies because of the higher detector MTF, which extends beyond the Nyquist of the FPD (2.5 cycles/mm). The effect of magnification and the focal spot choice on the MAF performance as realized in a realistic imaging setup is shown clearly with generalized metrics and the stent images. Furthermore, the FPD performance is seen to be less sensitive to the choice of the focal spot in comparison to the MAF because of the FPD's relatively poorer spatial resolution. The MAF stent images clearly showed the focal spot effect on the stent strut visibility and provided a strong argument for the use of a small focal spot with a high resolution detector.

In recent years, other types of FPDs based on CMOS technology have been investigated.^{27–30} These developing CMOS detectors have lower noise and enable smaller pixel sizes than TFT-based FPDs and thus may well have performance characteristics competitive with the MAF. Furthermore, FPDs using avalanche-mode amorphous selenium (a-Se) are being investigated and may be able to provide high spatial resolution and low dose quantum noise limited response.³¹ The conclusions from this study regarding total system performance and the effect of scatter, focal spot, and magnification will be equally applicable to those detectors.

In the present study, we evaluate the total system performance for a realistic situation where scatter, geometric unsharpness and finite focal spot size have effects on the overall image quality. The generalized approach shows total system

performance to be substantially degraded in comparison to the intrinsic detector performance and our results showed that the GDQE and GMTF both suffer a significant loss because of the finite size focal spot and scatter. The generalized linear system analysis identifies the weak link in the imaging system and the wise choice of different components that could provide the best use of available resources. The generalized linear system analysis also provides important insight about the total system performance that can be beneficial for the optimum design of task specific detectors.

ACKNOWLEDGMENTS

This work has been supported in part by the National Institutes of Health (NIH) Grant Nos. R01EB002873 and R01EB008425 and an equipment grant from Toshiba Medical Systems Corporation.

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¹I. A. Cunningham, in *Handbook of Medical Imaging: Physics and Psychophysics*, edited by J. Beutel, H. L. Kundel, and R. L. V. Metter (SPIE, Bellingham, 2000), Vol. 1, pp. 82–156.

²E. Samei, M. J. Flynn, and D. A. Reimann, “A method for measuring the presampled MTF of digital radiographic systems using an edge test device,” *Med. Phys.* **25**(1), 102–113 (1998).

³J. T. Dobbins, III, D. L. Ergun, L. Rutz, D. A. Hinshaw, H. Blume, and D. C. Clark, “DQE(f) of four generations of computed radiography acquisition devices,” *Med. Phys.* **22**(10), 1581–1593 (1995).

⁴H. Fujita, D. Y. Tsai, T. Itoh, K. Doi, J. Morishita, K. Ueda, and A. Ohtsuka, “A simple method for determining the modulation transfer function in digital radiography,” *IEEE Trans. Med. Imaging* **11**(1), 34–39 (1992).

⁵E. P. Muntz, “Analysis of the significance of scattered radiation in reduced dose mammography including magnification effects scatter suppression and focal spot and detector blurring,” *Med. Phys.* **6**, 110–117 (1979).

⁶J. A. Krol, D. A. Bassano, C. C. Chamberlain, and S. C. Prasad, “Scatter reduction in mammography with air gap,” *Med. Phys.* **23**, 1263–1270 (1996).

⁷K. Doi, H. K. Genant, and K. Rossmann, “Effect of geometric unsharpness upon image quality in fine-detail skeletal radiography,” *Radiology* **113**(3), 723–725 (1974).

⁸C. C. Shaw, X. Liu, M. Lemacks, J. X. Rong, and G. J. Whitman, “Optimization of MTF and DQE in magnification radiography—A theoretical analysis,” *Proc. SPIE* **3977**, 466–475 (2000).

⁹I. S. Kyprianou, S. Rudin, D. R. Bednarek, and K. R. Hoffmann, “Study of generalized MTF and DQE for a new micro-angiographic system,” *Proc. SPIE* **5368**, 349–360 (2004).

¹⁰I. Kyprianou, S. Rudin, D. R. Bednarek, and K. R. Hoffmann, “Generalizing the MTF and DQE to include x-ray scatter and focal spot unsharpness: Application to a new micro-angiographic system,” *Med. Phys.* **32**, 613–626 (2005).

¹¹A. Jain, D. R. Bednarek, C. Ionita, and S. Rudin, “A theoretical and experimental evaluation of the microangiographic fluoroscope: A high-resolution region-of-interest x-ray imager,” *Med. Phys.* **38**(7), 4112–4126 (2011).

¹²M. J. Binning, D. Orion, P. Yashar, S. Webb, C. N. Ionita, A. Jain, S. Rudin, L. N. Hopkins, A. H. Siddiqui, and E. I. Levy, “Use of the microangiographic fluoroscope for coiling of intracranial aneurysms,” *Neurosurgery* **69**(5), 1131–1138 (2011).

¹³P. Kan, P. Yashar, C. N. Ionita, A. Jain, S. Rudin, E. I. Levy, and A. H. Siddiqui, “Endovascular coil embolization of a very small ruptured aneurysm using a novel microangiographic technique: Technical note,” *J. Neurointerv. Surg.* (2012). [Available online <http://jn.is.bmj.com/content/5/2/e2.full>].

¹⁴S. Rudin and D. R. Bednarek, “Region of interest fluoroscopy,” *Med. Phys.* **19**(5), 1183–1189 (1992).

¹⁵Delt Electronic Products B. V., “Image Intensifier Tube brochure,” NL-9300 AB Roden, The Netherlands, Dwaziewegen 2, Roden.

¹⁶A. Jain, D. R. Bednarek, and S. Rudin, “Performance evaluation of a custom-made anti-scatter grid used for the high-resolution microangiographic fluoroscope (MAF),” *Med. Phys.* **36**, 2765 (2009).

¹⁷AAPM, “Instrumentation requirements of diagnostic radiological physicists (generic listing),” AAPM Report No. 60, Task Group 4, Diagnostic x-ray imaging committee (AAPM, Medical Physics Publishing, Madison, WI, 1998).

¹⁸I. A. Brizovich and G. T. Barnes, “A new type of grid,” *Med. Phys.* **4**, 451–453 (1977).

¹⁹W. Wang, C. Ionita, Y. Huang, B. Qu, A. Panse, A. Jain, D. R. Bednarek, and S. Rudin, “Region-of-interest micro-angiographic fluoroscope detector used in aneurysm and artery stenosis diagnoses and treatment,” *Proc. SPIE* **8313**, 831317 (2012).

²⁰A. Panse, C. N. Ionita, W. Wang, S. K. Natarajan, A. Jain, D. R. Bednarek, and S. Rudin, “The micro-angiographic fluoroscope (MAF) in high definition (HD) mode for improved contrast-to-noise ratio and resolution in fluoroscopy and roadmapping,” in *Proceedings of IEEE Nuclear Science Symposium Conference Record, 1997* (NSS/MIC Knoxville, TN, 2010), pp. 3217–3220.

²¹J. M. Boone, B. A. Arnold, and J. A. Seibert, “Characterization of the point spread function and modulation transfer function of scattered radiation using a digital imaging system,” *Med. Phys.* **13**(2), 254–256 (1986).

²²A. Jain, A. Kulhs-Gilchrist, D. R. Bednarek, and S. Rudin, “An analysis of signal-to-noise ratio differences between the new high-sensitivity, microangiographic fluoroscope (HSMF) and a standard flat-panel detector,” *Med. Phys.* **35**, 2636 (2008).

²³K. Gill, C. Ionita, D. Bednarek, and S. Rudin, “Effective dose rate comparison between the micro-angiographic fluoroscope (MAF) and the x-ray image intensifier (XII) used during neuro-endovascular device deployment procedures,” *Med. Phys.* **38**(6), 3440 (2011).

²⁴D. C. Hunt, O. Tousignant, and J. A. Rowlands, “Evaluation of the imaging properties of an amorphous selenium-based flat panel detector for digital fluoroscopy,” *Med. Phys.* **31**(5), 1166–1175 (2004).

²⁵C.-Y. Lin, W.-J. Lee, S.-J. Chen, C.-H. Tsai, J.-H. Lee, C.-H. Chang, and Y.-T. Ching, “A study of grid artifacts formation and elimination in computed radiographic images,” *J. Digit. Imaging* **19**(4), 351–361 (2006).

²⁶A. Jain, A. Panse, C. Ionita, V. Singh, S. Rudin, and D. Bednarek, “Improved high-resolution imaging through an aneurysm coil mass using the MAF compared with a flat panel detector,” *Med. Phys.* **38**(6), 3440 (2011).

²⁷M. Esposito, T. Anaxagoras, A. Fant, K. Wells, A. Konstantinidis, J. P. F. Osmond, P. M. Evans, R. D. Speller, and N. M. Allinson, “DynAMITE: A wafer scale sensor for biomedical applications,” *J. Instrum.* **6**(12), C12064 (2011).

²⁸A. Jain, D. R. Bednarek, and S. Rudin, “Theoretical performance analysis for CMOS based high resolution detectors,” *Proc. SPIE* **8668**, (2013).

²⁹B. Loughran, S. N. S. Vasan, V. Singh, C. Ionita, A. Jain, D. R. Bednarek, A. H. Titus, and S. Rudin, “Design considerations for a new, high resolution micro-angiographic fluoroscope based on a CMOS sensor (MAF-CMOS),” *Proc. SPIE* **8668** (2013).

³⁰A. C. Konstantinidis, “Evaluation of digital X-ray detectors for medical imaging applications,” Ph.D. thesis, University College London, London, UK, 2011.

³¹M. M. Wronski and J. A. Rowlands, “Direct-conversion flat-panel imager with avalanche gain: Feasibility investigation for HARP-AMFPI,” *Med. Phys.* **35**(12), 5207–5218 (2008).

³²See supplementary material at <http://dx.doi.org/10.1118/1.4792460> for effect of magnification for different focal spot sizes.