Journal of Medical Imaging

MedicalImaging.SPIEDigitalLibrary.org

Back-irradiated and dual-screen sandwich detector configurations for radiography

Anthony R. Lubinsky
Adrian Howansky
Hao Zheng
Wei Zhao

Back-irradiated and dual-screen sandwich detector configurations for radiography

Anthony R. Lubinsky,* Adrian Howansky, Hao Zheng, and Wei Zhao

Stony Brook University, Department of Radiology, Health Sciences Center, Stony Brook, New York, United States

Abstract. Recent advances in thin film transistor array technology have enabled the possibility of “back-irradiated” (BI) indirect active-matrix flat-panel imagers (AMFPIs), in which x-rays first expose the scintillator through the optical sensor, and “dual-screen” AMFPIs, in which two scintillating screens are sandwiched around a bidirectional active matrix. We developed a theoretical treatment of such detectors. The theory is used to investigate possible imaging performance improvements over conventional “front-irradiation” (FI) AMFPIs, where the active matrix is opposite the x-ray entrance surface. Simple expressions for the modulation transfer function, normalized noise power spectrum, Swank factor (A_s), Lubberts function $L(f)$, and spatial frequency-dependent detective quantum efficiency $DQE(f)$ are derived and used to compute these quantities for a variety of FI, BI, and dual-screen detector configurations. $DQE(f)$ is used as the figure of merit for optimizing and comparing the performance of the various configurations. Large performance improvements over FI single-screen systems are found possible with BI. Further improvements are found possible with dual-screen configurations. The ratio of the thicknesses of the two screens that optimizes DQE is generally asymmetric, with the thinner screen facing the incident flux. The optimum ratio depends on the x-ray attenuation length in the screen. © 2019 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.6.3.033501]

Keywords: flat-panel detector; digital radiography; indirect x-ray detection; detective quantum efficiency; Swank factor; Lubberts effect.

Paper 19064R received Mar. 11, 2019; accepted for publication Jun. 18, 2019; published online Jul. 9, 2019; corrected Jul. 25, 2019.

1 Introduction

Indirect active-matrix flat-panel imagers (AMFPIs) have become common in clinical applications of digital radiography. Large area thin film transistor (TFT) array technology has reached an impressive level of cost-versus-performance after much investment by the display industry, and considerable effort has been expended to facilitate their use in detectors suitable for medical x-ray imaging.¹ The arrays typically consist of pixels containing a hydrogenated amorphous silicon (a-Si:H) TFT coupled with an a-Si:H photodiode sensor. The array is placed in intimate contact with a scintillating screen that absorbs x-rays and produces light, which is detected by the photodiodes to generate electrical charge signals. The charges are stored on the pixel capacitances until they are read out by sequentially turning on rows of TFTs to transfer the charges to an array of charge-sensitive amplifiers. The amplifiers' output signals are then sent to an analog-digital-converter for digitization and subsequent storage in memory. An illustration of such a detector is shown in Fig. 1.

Indirect AMFPIs are conventionally fabricated and operated in a “front irradiation” (FI) mode, in which the sensor array is placed opposite the x-ray entrance surface of the intensifying screen, as drawn in Fig. 1. More recently, some work including the development of commercial devices^{2,3} has been done on “back irradiation” (BI) AMFPIs, in which x-rays first enter the screen through the active matrix. The predominance of FI single-screen AMFPI systems contrasts with prior history in screen/film (S/F) radiography, where “dual-screen” systems comprising a dual-emulsion film “sandwiched” between two intensifying screens were a standard operating mode, since their introduction in 1918.⁴

Elementary considerations suggest that BI and dual-screen AMFPI configurations have performance advantages over FI systems. The thickness of a single x-ray screen used for a particular application is typically chosen based on a tradeoff between x-ray absorption and image spatial resolution. Increasing screen thickness improves x-ray absorption and thus sensitivity (i.e., dose-efficiency) but also decreases spatial resolution due to light spreading, necessitating a compromise. BI and dual-screen configurations may improve this compromise, as they reduce the average distance between x-ray absorption events and the light-sensitive layer. This offers the possibility to improve spatial resolution and detector sensitivity by minimizing light spreading and absorption before detection, without sacrificing the total screen thickness.

Despite their nominal performance advantages, BI and dual-screen AMFPI configurations have not yet been widely implemented. This may be partially due to challenges presented by the active-matrix. For example, x-rays may be absorbed or scatter off the TFT array's substrate, which is typically 0.7-mm-thick borosilicate glass.⁵ These processes would respectively degrade the detector's x-ray quantum efficiency and image contrast. This limitation makes such configurations nonideal at radiographic energies; we calculate that 3% to 15% of x-rays are attenuated in such layers from 20 to 80 keV.

Recent advances in TFT array technology⁶ are now enabling their manufacture on thin, flexible, and possibly transparent substrates. These developments address the practical challenges outlined above and can potentially be leveraged in BI and dual-screen AMFPI detector configurations like the examples shown in Fig. 2. This possibility motivates investigating the theoretical imaging performance of such configurations. Relevant previous

*Address all correspondence to Anthony R. Lubinsky, Email: Anthony.Lubinsky@stonybrookmedicine.edu

Fig. 1 Illustration of a typical indirect AMFPI, using a single screen exposed in FI mode.

work has studied stacked screen-and-light-detector configurations for application to dual-energy radiography.^{7,8} Other investigators have also described⁹ and characterized¹⁰ the imaging performance of dual-S/F systems, and the possibility of dual-screen AMFPI systems has been mentioned in the patent literature.¹¹ However, to our knowledge, no prior attempt has been made to theoretically optimize the performance of systems comprising light-scattering screens placed on both sides of a bidirectional light detecting layer. The present work studies potential improvements in the detective quantum efficiency (DQE) of such dual-screen detector configurations, using analytical methods describing the x-ray and optical interactions in intensifying screens. The expressions derived for the dual-screen detector are also used to compare the DQE of FI and BI configurations, by taking the limit where one of the two screens has zero thickness.

2 Methods

2.1 Overview

The DQE is used as a figure of merit for comparing the performance of digital x-ray detector systems. Rabbani et al.¹² and Nishikawa et al.¹³ have modeled the interactions in a screen

as a series of cascaded stochastic events, leading to an approximate expression for the DQE as:

$$DQE(f) = A_q A_s L(f) R_N(f), \quad (1)$$

where A_q is the x-ray quantum absorption efficiency, A_s is the Swank factor, $L(f)$ is the Lubberts function, and $R_N(f)$ is the ratio of the quantum noise power to the total system noise power, which may include other effects such as secondary quantum noise, structure noise, detector noise, or electronic noise. The Swank factor¹⁴ quantifies noise due to random variations in the magnitude of the response to absorption events. The Lubberts function¹⁵ quantifies noise due to random variations in the spatial response to absorption events. Here we consider the case of prompt-emitting screens with high conversion gain and assume that structure noise, electronic noise, and secondary quantum noise are small compared to the x-ray quantum noise. Under these conditions, $DQE(0)$ is reduced by A_q and A_s , and $L(f)$ describes the “roll-off” of $DQE(f)$ with increasing spatial frequency.

The following derivations use a depth-dependent model to calculate A_s and $L(f)$ for proposed detector configurations under monoenergetic x-ray beam conditions. We consider the simplest case where x-ray energy is deposited locally at the point of each interaction (i.e., K-fluorescence effects and Compton scatter are neglected). The calculated product $A_q A_s L(f)$ is taken to measure $DQE(f)$ prior to losses in the light detector, which is used to optimize detector design in terms of screen mass loadings (thicknesses) and backing reflectances.

2.2 Single Screen Configuration

For a given x-ray exposure, the average number of photons emitted by a screen (i.e., its zero-frequency DC signal) is

$$S = \frac{\mu E}{W} \int_0^T w(z) \epsilon(z) dz, \quad (2)$$

where E denotes x-ray energy and W is the average energy required to generate one scintillation photon. Here, z is depth in the screen as measured from the x-ray entrance side, $w(z)$ is a weighting function accounting for exponential x-ray attenuation in the screen given as

Fig. 2 Cross-sectional schematic of a possible (a) BI and (b) dual-screen AMFPI. X-rays first pass through the active matrix before exposing the scintillator in BI geometry. The dual-screen AMFPI comprises an FI “front” screen, and a BI “back” screen in intimate contact with a bidirectional light detector layer and thin transparent substrate.

$$w(z) = e^{-\mu z}, \quad (3)$$

where $\epsilon(z)$ is the screen's depth-dependent light escape efficiency,^{16,17} and μ is the photoelectric linear attenuation coefficient. We consider one screen material in our derivations such that the $\mu E/W$ term in Eq. (2) is an arbitrary constant, which we hereafter fix to unity for simplicity.

Assuming K -fluorescence and other random variations in x-ray conversion gain can be neglected, the Swank factor of a single screen is determined by the depth-dependence in its light escape efficiency, and can be calculated as¹⁶

$$A_s = \frac{\{\int_0^T w(z)\epsilon(z)dz\}^2}{C \int_0^T w(z)\epsilon(z)^2 dz}, \quad (4)$$

where C is a normalizing factor given as

$$C = \frac{(1 - e^{-\mu T})}{\mu}. \quad (5)$$

The Lubberts function $L(f)$ is due to depth dependence in the screen's modulation transfer function (MTF) and is given as^{15,18}

$$L(f) = \frac{\text{MTF}^2(f)}{\text{NNPS}(f)}, \quad (6)$$

where normalized noise power spectrum (NNPS) (f) is the quantum noise power spectrum, normalized by dividing by the mean signal squared. MTF and NNPS are expressed as integrals over the depth of the screen:

$$\text{MTF}(f) = \frac{\int_0^T w(z)\epsilon(z)\text{MTF}(z, f)dz}{N_m}, \quad (7)$$

$$\text{NNPS}(f) = \frac{\int_0^T w(z)\epsilon^2(z)\text{MTF}^2(z, f)dz}{N_n}. \quad (8)$$

MTF(z, f) denotes the screen MTF applicable to light emitted at depth z , T is the screen thickness, and N_m and N_n are normalizing integrals for MTF and NNPS, respectively:⁷

$$A_s(f) = \frac{[\int_0^{z_d} e^{-\mu z} \epsilon_1(z)dz + \int_0^{T-z_d} e^{-\mu(T-z)} \epsilon_2(z)dz]^2}{C[\int_0^{z_d} e^{-\mu z} \epsilon_1^2(z)dz + \int_0^{T-z_d} e^{-\mu(T-z)} \epsilon_2^2(z)dz]}, \quad (14)$$

$$\text{MTF}(f) = \frac{\int_0^{z_d} e^{-\mu z} \epsilon_1(z)\text{MTF}_1(z, f)dz + \int_0^{T-z_d} e^{-\mu(T-z)} \epsilon_2(z)\text{MTF}_2(z, f)dz}{N_{2m}}, \quad (15)$$

$$\text{NNPS}(f) = \frac{\int_0^{z_d} e^{-\mu z} \epsilon_1^2(z)\text{MTF}_1^2(z, f)dz + \int_0^{T-z_d} e^{-\mu(T-z)} \epsilon_2^2(z)\text{MTF}_2^2(z, f)dz}{N_{2n}}, \quad (16)$$

where $\text{MTF}_1(z, f)$ is the depth-dependent MTF of screen 1, $\epsilon_1(z)$ is its depth-dependent light escape efficiency, and similarly for screen 2. Note that N_{2m} and N_{2n} are normalizing integrals equal to the corresponding numerators evaluated at $f = 0$. The two screens are assumed to be of the same material, with the light detector located at depth z_d .

$$N_m = \int_0^T w(z)\epsilon(z)\text{MTF}(z, 0)dz, \quad (9)$$

$$N_n = \int_0^T w(z)\epsilon^2(z)\text{MTF}^2(z, 0)dz. \quad (10)$$

The Lubberts function can be expressed by combining Eqs. (6)–(8):

$$L(f) = \left(\frac{N_n}{N_m^2}\right) \frac{[\int_0^T w(z)\epsilon(z)\text{MTF}(z, f)dz]^2}{\int_0^T w(z)\epsilon^2(z)\text{MTF}^2(z, f)dz}. \quad (11)$$

2.3 Front versus Back X-Ray Irradiation

When a screen's response to x-rays depends on the depth of x-ray interaction, the above metrics [S , A_s , MTF, $L(f)$] will depend on the direction of x-ray irradiation (i.e., FI versus BI). This is due to exponential x-ray absorption within the screen, which weights its depth-dependent light escape efficiency and MTF differently in each geometry. The x-ray weighting function $w(z)$ for an FI configuration is defined by Eq. (3), while for a BI configuration it is

$$w^{\text{BI}}(z) = e^{-\mu(T-z)}. \quad (12)$$

It is of interest to compare the zero-frequency (DC) signals between BI and FI. Their comparison can be expressed as ratio of the BI to FI signal using

$$\frac{S_{\text{BI}}}{S_{\text{FI}}} = \frac{\int_0^T e^{-\mu(T-z)} \epsilon(z)dz}{\int_0^T e^{-\mu z} \epsilon(z)dz}. \quad (13)$$

2.4 Dual-Screen Configuration

As shown in Fig. 2(b), a dual-screen detector comprises an FI “front screen” and BI “back screen.” The detector's A_s , MTF, and NNPS can therefore be described by splitting Eqs. (4), (7), and (8) into two parts (corresponding to each screen), and changing the variable of integration in the second integral

2.5 Swank Model

Up to this point, the derived expressions have not considered a particular form of $\epsilon(z)$ and $\text{MTF}(z, f)$. Here, the Swank model¹⁹ is used to provide simple expressions for these terms in the “diffusion limit,” i.e., when the light scattering coefficient in

the screen is large and the optical absorption coefficient is small. This assumption represents an optically turbid granular phosphor screen.¹⁸ In the model, the spatial spreading depends strongly on the depth of x-ray interaction within the screen. These depth effects have recently been measured experimentally and found to be substantial for typical granular screens such as Gd₂O₂S:Tb (GOS).²⁰

2.5.1 Dual screens with optically reflective (white) backing

In the diffusion limit, a screen with a perfectly reflective (i.e., white) backing has unit escape efficiency at all depths. This makes its Swank factor unity as well, given our current assumption that other random variations in x-ray conversion gain can be neglected. Its depth-dependent MTF is¹⁹

$$\text{MTF}^W(z, f) = \frac{\cosh(2\pi f z)}{\cosh(2\pi f T)}. \quad (17)$$

The overall MTF of a dual-screen configuration comprising two screens with “white” backings is then

$$\begin{aligned} \text{MTF}^W(f) &= \frac{\int_0^{z_d} e^{-\mu z} \left[\frac{\cosh(2\pi f z)}{\cosh(2\pi f z_d)} \right] dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left\{ \frac{\cosh(2\pi f z)}{\cosh[2\pi f(T-z_d)]} \right\} dz}{N_{2m}}. \end{aligned} \quad (18)$$

Casting the NNPS into sums of integrals in the same way as above results in

$$\text{MTF}^B(f) = \frac{\int_0^{z_d} e^{-\mu z} \left(\frac{z}{z_d} \right) \left[\frac{\sinh(2\pi f z)}{\sinh(2\pi f z_d)} \right] dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left(\frac{z}{T-z_d} \right) \left\{ \frac{\sinh(2\pi f z)}{\sinh[2\pi f(T-z_d)]} \right\} dz}{N_{2m}}, \quad (23)$$

$$\text{NNPS}^B(f) = \frac{\int_0^{z_d} e^{-\mu z} \left(\frac{z}{z_d} \right)^2 \left[\frac{\sinh(2\pi f z)}{\sinh(2\pi f z_d)} \right]^2 dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left(\frac{z}{T-z_d} \right)^2 \left\{ \frac{\sinh(2\pi f z)}{\sinh[2\pi f(T-z_d)]} \right\}^2 dz}{N_{2n}}. \quad (24)$$

2.6 Dual-Screen Calculations Using Swank Model

We proceeded with calculations of A_S , MTF, and NNPS for various dual-screen configurations using Eqs. (18) and (19) and (21), (23), and (24). The integrals were evaluated analytically using the symbolic math kernel of Maple (Waterloo Maple Inc., Waterloo, Ontario) implemented in MATLAB (The MathWorks Inc., Natick, Massachusetts). This resulted in algebraic expressions for the three quantities in terms of x-ray attenuation coefficient, detector position, and the thicknesses of the two screens. For clarity, we omit the lengthy algebraic expressions and only provide that for MTF^W as an example below:

$$\begin{aligned} \text{MTF}^W(f) &= \frac{2\mu - e^{-(\mu+\nu)z_d} [\mu - \nu + (\mu + \nu)e^{2\nu z_d}]}{2C(\mu^2 - \nu^2) \cosh(\nu z_d)} \\ &\quad - \frac{2\mu - e^{(\mu-\nu)T_2} [\mu + \nu + (\mu - \nu)e^{2\nu T_2}]}{2C(\mu^2 - \nu^2) e^{\mu T} \cosh(\nu T_2)}, \end{aligned} \quad (25)$$

where

$$\begin{aligned} \text{NNPS}^W(f) &= \frac{\int_0^{z_d} e^{-\mu z} \left[\frac{\cosh(2\pi f z)}{\cosh(2\pi f z_d)} \right]^2 dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left(\frac{\cosh(2\pi f z)}{\cosh[2\pi f(T-z_d)]} \right)^2 dz}{N_{2n}}. \end{aligned} \quad (19)$$

2.5.2 Dual screens with optically-absorptive (black) backing

In the diffusion limit, a screen with a perfectly optically absorptive (black) backing has depth-dependent light escape efficiency

$$e^B(z) = \frac{z}{T}. \quad (20)$$

Its Swank factor is then

$$A_s^B(f) = \frac{\left[\int_0^{z_d} e^{-\mu z} \left(\frac{z}{z_d} \right) dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left(\frac{z}{T-z_d} \right) dz \right]^2}{C \left[\int_0^{z_d} e^{-\mu z} \left(\frac{z}{z_d} \right)^2 dz + \int_0^{T-z_d} e^{-\mu(T-z)} \left(\frac{z}{T-z_d} \right)^2 dz \right]}. \quad (21)$$

Its depth-dependent MTF is given as¹⁹

$$\text{MTF}^B(z, f) = \frac{\sinh(2\pi f z)}{\sinh(2\pi f T)}, \quad (22)$$

therefore, the MTF and NNPS of a dual-screen configuration comprising two screens with “black” backings are

$$\nu = 2\pi f, \quad (26)$$

$$T_2 = T - z_d. \quad (27)$$

The expressions were coded into functions used to evaluate the imaging performance parameters of various detector configurations and screen types. Single-screen FI and BI cases are included in these analyses as limiting cases where $z_d = T$ and $z_d = 0$, respectively.

3 Results

3.1 Front versus Back Irradiation

Figure 3(a) compares the DC output signal for a 300- μm screen with either a perfectly reflective “white” backing or perfectly absorptive “black” backing, in both FI and BI. Figure 3(b) shows the ratio of the BI to FI signals. The abscissae represent the x-ray absorption probability, which is determined by the linear attenuation coefficient μ . It thus serves as a measure of the average x-ray attenuation length in the screen (either due to the screen’s x-ray attenuation properties or x-ray energy).

Fig. 3 Comparison of (a) DC signal for a single screen ($T = 300 \mu\text{m}$) with white or black backing, in FI and BI, for a given exposure. (b) The corresponding ratio of BI:FI signal.

In the diffusion limit, a screen with a white backing has no difference between FI and BI signal, because the light escape efficiency is unity in both cases. With an absorptive backing, however, the two cases differ. When the x-ray absorption probability is low (i.e., low μ), both FI and BI give a signal which is reduced by half due to light absorption in the backing. As the x-ray absorption probability increases (i.e., high μ), BI is increasingly favored over FI as more x-rays are absorbed closer to the detector. For FI, the signal first increases with x-ray absorption, but then decreases at higher values. This is because the reduced x-ray path length increases light absorption in the backing. When x-ray absorption is highest, BI is much advantaged over FI since the light photons are produced in a thin layer near the detector in BI, whereas in FI the photons are produced in a thin layer far from the detector.

Figure 4 shows the Swank factor for a $300\text{-}\mu\text{m}$ screen with either white or black backing, in both FI and BI. The Swank factor is unity for a screen with white backing in both FI and BI, again because the depth-dependent light escape efficiency is unity at all depths and because we consider no other random variations in x-ray conversion gain. The Swank factor is reduced with absorptive backings, however. When x-ray absorption probability is low (i.e., low μ), both FI and BI give $A_s = 0.75$, which is the value predicted by Swank for this limiting case.¹⁹ As the x-ray absorption probability is increased, BI is increasingly

Fig. 4 Comparison of the optical Swank factor for a single screen ($T = 300 \mu\text{m}$) with white and black backing, in FI and BI.

favored over FI. This result is because the screen's light escape efficiency is depth-dependent. As shown in Eq. (4), A_s is higher in BI because the squared integral of $\epsilon(z)$ weighted over depth (i.e., the numerator) is always less than or equal to the integrated square of $\epsilon(z)$ weighted over depth (i.e., the denominator). This inequality is greater in FI than in BI because $\epsilon(z)$ is weighted more heavily when it is smaller.

Figures 5 and 6 show the impact of irradiation geometry on the MTF(f) and DQE(f) of the same $300 \mu\text{m}$ screen, for a linear x-ray absorption coefficient of $\mu = 3.33 \text{ mm}^{-1}$. Figure 5 shows the case where the scintillator has a white backing layer, while Fig. 6 shows a black backing layer.

Figure 5(a) shows that BI improves the overall MTF(f) of a single screen with white backing. This is because in BI, x-ray absorption is weighted more heavily toward the part of the screen that is closest to the light detector, where the z -dependent MTF(z, f) is higher. Figure 5(b) shows that the DQE(f) is also improved in BI at higher frequencies, due to improvement in the Lubberts factor $L(f)$. This improvement is for similar reasons as the improvement in A_s described above. That is, the “squared-sums” in the numerator of Eq. (11) more closely match the “summed-squares” in its denominator, because the product of $\epsilon(z)$ MTF(z, f) is weighted more heavily when it is higher in BI.

Figure 6(a) shows that MTF is generally improved with the use of a light-absorbing backing instead of a reflective one, and that BI is still advantaged over FI for the reasons explained above. Figure 6(b) shows that the DQE(f) is improved at all frequencies, due to improvements in both A_s and $L(f)$ as described above.

3.2 Improving High-Frequency DQE Using Dual Screens

The dual-screen sandwich concept can be used to improve high spatial frequency response while keeping DQE(0) constant. Figure 7(a) compares the MTF of a dual-screen configuration comprising two $80\text{-}\mu\text{m}$ -thick granular screens sandwiched around a bidirectional light detector, and an FI single-screen configuration with the same total thickness. The ratio of the MTFs is also plotted as a function of spatial frequency. For simplicity, x-ray absorption was considered uniform with depth, and both systems assumed to comprise screens with white backings. The MTF at higher frequencies (near 5 mm^{-1}) is roughly

Fig. 5 Comparison of (a) $MTF(f)$ and (b) $DQE(f)$ for a single screen with an optically reflective backing in FI and BI.

Fig. 6 Comparison of (a) $MTF(f)$ and (b) $DQE(f)$ for a single screen with an optically absorptive backing in FI and BI.

Fig. 7 (a) Comparison between the MTF for a single 160 μm screen in FI mode and the MTF of a dual-screen system comprising two 80- μm screens sandwiched around a thin light detector layer. (b) Comparison of $L(f)$, which is proportional to $DQE(f)$, of the same dual-screen and FI single screen systems. Uniform x-ray absorption versus depth and white screen backings were assumed in calculations.

doubled in the dual-screen case. Further note that in this limiting case, the theory gives $MTF^{dual}(2f) = MTF^{single}(f)$, meaning that the dual screen MTF at 4 mm^{-1} , for example, is equal to the single screen MTF at 2 mm^{-1} . Figure 6(b) shows that this behavior is similar with their Lubberts functions $L(f)$, which are proportional to $DQE(f)$.

3.3 Improving Low-Frequency DQE Using Dual Screens

The dual-screen advantage can also be used to improve the DQE at low spatial frequencies while keeping it constant at higher frequencies. This is shown in Fig. 8, which compares the $DQE(f)$ of a dual-screen configuration where both screens have 260 or 400 μm thickness to that of an FI single screen (300 μm thick) for two different linear absorption coefficients, corresponding to lower and higher x-ray beam kVp settings. Both systems assume white screen backings. Comparing Figs. 8(a) and 8(b) shows that the dual-screen configuration improves

$DQE(0)$ more at higher energies, where there is greater difference in the x-ray quantum efficiency A_q of each system. This improvement may be especially useful for applications requiring harder x-ray beams, such as cone-beam computed tomography.

3.4 Optimizing the Dual-Screen Sandwich Configuration

3.4.1 Screens with optically reflective (white) backing

Figure 9(a) shows the results of a series of calculations in which a single screen (300 μm thickness) is subdivided into two parts of different relative thicknesses and sandwiched around a bidirectional light detector with negligible thickness at the position shown on the x-axis. The linear x-ray attenuation coefficient was assumed to be $\mu = 3.33 \text{ mm}^{-1}$ for consistency with previous calculations. Both screens have white backings, and the performance parameters are evaluated at $f = 5 \text{ mm}^{-1}$. Under these conditions, $DQE(0)$ will be the same for all configurations but

Fig. 8 Comparison between the $DQE(f)$ of a single-screen system and a dual-screen system with the same high frequency DQE, but greater total thickness, for (a) softer and (b) harder x-ray beams.

Fig. 9 (a) The MTF and NNPS of many dual-screen systems at 5 mm^{-1} , where the front screen thickness is varied and total thickness is held constant at 300 μm . Both screens have perfectly reflective backings. X-ray beam attenuation versus depth is also shown for reference. Zero front screen thickness corresponds to a BI single screen, and 300 μm thickness corresponds to an FI single screen. (b) The corresponding Lubberts function and DQE of the dual-screen systems at 5 mm^{-1} , as a function of front screen thickness.

the higher-frequency MTF and DQE will vary. Note that the rightmost point on the abscissa ($x = 300 \mu\text{m}$) corresponds to an FI single-screen system, while the leftmost point ($x = 0 \mu\text{m}$) corresponds to a BI single screen system.

Figure 9(a) shows the MTF and NNPS for the various detector configurations. The depth-dependent x-ray absorption in the screens is also shown for reference. The MTF of the dual-screen system is optimal when the front screen is $\sim 75 \mu\text{m}$, or 25% of the total thickness. It is remarkably improved over that in the FI system, by a factor of more than 3. Figure 9 (b) shows the corresponding $L(f)$ and $\text{DQE}(f)$ for each detector configuration. The DQE is optimal when the front screen is $\sim 110 \mu\text{m}$, or $\sim 37\%$ of the total thickness. The dual-screen DQE is also much improved over that in the FI system, by a factor of more than 2.

The relative thickness configuration that maximizes $\text{DQE}(f)$ depends on the depth-dependent x-ray attenuation profile through the detector stack, which itself depends on the ratio of x-ray attenuation length to screen thickness. The calculations shown in Fig. 9 were repeated for different x-ray linear attenuation coefficients to show how the optimal configuration maximizing DQE varies as a function of x-ray energy. The results of these calculations are presented in Fig. 10, which shows the front screen fractional thickness that maximizes DQE (5 mm^{-1}) as a function of the total screen thickness, which is specified in terms of half-value layers (HVLs) for the incident beam. The thinner screen always faces the incident x-ray flux. For a high-energy x-ray beam, the total screen thickness is a small number of HVLs, and the optimal configuration approaches 50% front screen thickness ratio. For a lower-energy x-ray beam, the total screen thickness is a greater number of HVLs, and a thinner front screen is optimal.

3.4.2 Screens with optically absorptive (black) backing

The optical component of the Swank factor becomes significant when screens with black backings are used in a dual-screen detector. The effect is demonstrated in Fig. 11, which shows the (optical) Swank factor calculated for a variety of dual-screen detector configurations with different x-ray linear attenuation coefficients. The single screen BI case always results in a higher Swank factor compared to FI, as was noted in Sec. 3.1. In

Fig. 10 Ratio of front screen thickness to total thickness which maximizes DQE at 5 cy/mm versus total thickness in units of HVL. Thinner screen always faces incident x-ray flux.

Fig. 11 Optical Swank factor A_s versus front screen thickness for dual-screen systems with absorptive (black) backings for various values of x-ray linear absorption coefficient.

addition, A_s is reduced by 25% to 35% for the configurations which were found to optimize the dual-screen performance with white backing screens.

MTF, $L(f)$ and $\text{DQE}(f)$ have also been calculated as a function of detector position in various dual-screen configurations for the case where the screens have absorptive (black) backings. Both the MTF and the Lubberts factor are found to improve with absorptive (black) backing screens, but the lower Swank factor causes the $\text{DQE}(f)$ in the optimal configuration to be lower than that for dual-reflective (white) backing screens. The low-frequency DQE is also lower for absorptive (black) backings because of the Swank factor loss.

4 Discussion

4.1 Impact of Transparent Substrate Thickness

The calculations in this work considered negligible detector substrate between the two screens, which means that the MTF is dominated by light scattering in the screens. However, if light from the back screen must pass through a transparent substrate (e.g., of a TFT array) to reach the light detector, then there will be additional blur and loss of MTF due to light diffusion in the substrate.²¹ We model the impact of this effect using expressions by Koch, assuming the substrate has an index of refraction of $n = 1.65$. Figure 12 shows the expected MTF improvement ratio at $f = 1, 2$, and 4 cy/mm with dual screens as a function of the thickness of the glass substrate. The result shows that a substrate thickness of $< 35 \mu\text{m}$ is required to maintain a substantial improvement factor (e.g., 50% to 60% or more at higher spatial frequencies).

4.2 Practical Considerations and Advantages of Dual-Screen FPDs

In order to realize the imaging performance advantages offered by a dual-screen approach, the light detector array must be made capable of efficiently sensing and recording this information. This presents several challenges for fabrication. As noted above, if a glass substrate is used it must be thin ($< 35 \mu\text{m}$) in order to maintain a substantial improvement factor in MTF and Lubberts factor. Such a thin substrate might be achieved by bonding and debonding a thin polyimide substrate from a glass carrier.⁶ If the thin layer were removed before contact with the screen, then a

Fig. 12 The effect of an optical substrate layer with $n = 1.65$ on the MTF of the dual-screen system, as a function of substrate thickness, for spatial frequencies of 1, 2, and 4 cy/mm.

substrateless configuration would result. A second method would be to manufacture the photosensor array directly on the surface of the thicker (bottom) screen (with a possible thin transparent protective layer) by, for example, ink jet printing means.²² A third method would be to develop a scintillating glass which could be used as the back screen, and onto which the light detector array could be deposited.

In early S/F technology, the potential performance improvements with dual screens were not fully realized due to “cross-over,” i.e., the penetration of light photons through the film emulsion, followed by reflection by the screen on the opposite side. In the case of a dual-screen digital detector, light must not penetrate the photodiode layer and reflect back from the opposite screen. Also, care must be taken to minimize light scatter within the pixels, to prevent light from reaching the photosensor along any path other than directly from the adjacent screens. Finally, the sensing element must also have high sensitivity to light incident from both directions.

The simple calculations presented in this work demonstrate how depth dependence in the light escape efficiency and MTF of granular screens (e.g., GOS) can be leveraged to improve I-FPD imaging performance. Other potential advantages of dual-screen FPD include low cost because of the lack of expensive vacuum deposited layers of scintillator or photoconductor, the possibility of a flexible detector to minimize geometric blur, the absence of x-ray scattering from heavy elements in a thick glass substrate,²³ and the additional design freedom (relative thicknesses) to optimize for different applications (x-ray energies).

4.3 Limitations of the Current Model

The presented model for dual-screen FPDs is not exhaustive, as phenomena such as “Poisson-excess” in x-ray conversion gain,¹² polychromatic x-ray absorption, and “cross-talk” between layers (e.g., due to K-fluorescence or Compton scatter) were not considered for simplicity. Poisson-excess and polychromatic x-ray absorption are expected to degrade A_s from our current estimates, while “cross-talk” may degrade both A_s and $L(f)$. K-fluorescence or Compton scatter effects can be quantitatively studied using Monte Carlo techniques. Although the effect on MTF may not be large compared to optical scattering

effects, the DQE (10 cy/mm) may be reduced by 20% to 30%, and the DQE(0) reduced by 5% to 10% at energies above the K edge (60 keV) due to x-ray and electron transport in a phosphor screen material such as CsI.²⁴ Further, expressing the DQE as a multiplicative product as in Eq. (1) assumes Poisson-distributed cascaded processes.²⁵ Furthermore, the presented calculations using the Swank model considered only cases in the diffusion limit, with ideal boundary conditions. Swank¹⁹ has derived expressions describing $\epsilon(z)$ and $MTF(z, f)$ in screens with finite optical scattering and absorption coefficients, and nonideal boundary conditions. These expressions can be easily incorporated into the framework presented here to model systems where these parameters are known.

5 Conclusions

This work presented an analytical theory for the imaging performance of granular x-ray screens sandwiched around a bidirectional light detecting layer. The theory was used to study the DQE of BI and dual-screen indirect AMFPI detector configurations. Optimized dual-screen configurations were found to give large potential improvements in higher-frequency $MTF(f)$ and $DQE(f)$ over the standard FI single-screen configuration with the same total phosphor thickness and x-ray absorption. Alternatively, higher values of $DQE(0)$ can be achieved using a dual-screen configuration, while maintaining the same higher-frequency DQE response as a standard FI single screen.

Disclosures

The authors have no financial interests to disclose.

Acknowledgments

The authors gratefully acknowledge financial support from the NIH (R01EB026267). Research reported in this publication was also supported by the National Heart, Lung, and Blood Institute of the National Institutes of Health under Award Number U01HL127522 and the NYS Department of Economic Development under Award Number C140151. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health. Additional support was provided by the Center for Biotechnology, a New York State Center for Advanced Technology, Stony Brook University, Cold Spring Harbor Laboratory, Brookhaven National Laboratory, and Northwell Health. An abbreviated version of some of the results appeared in SPIE Medical Imaging 2018 Proceedings Volume 10573, *Physics of Medical Imaging*: A. R. Lubinsky, Adrian Howansky, Hao Zheng, and Wei Zhao “Dual screen sandwich configurations for digital radiography” 105735U (9 March 2018). The authors are grateful to John Rowlands for help and advice.

References

1. J. A. Rowlands and J. Yorkston, “Flat panel detectors for digital radiography,” in *Handbook of Medical Imaging, Volume 1*, J. Beutel, H. L. Kundel, and R. L. Van Metter, Eds., pp. 223–328, SPIE Publications, Bellingham, Washington, (2009).
2. K. Sato et al., “Effect of x-ray incident direction and scintillator layer design on image quality of indirect-conversion flat-panel detector with GOS phosphor,” *Proc. SPIE* **7961**, 79614I (2011).
3. S. Rivetti et al., “Characterization of a clinical unit for digital radiography based on irradiation side sampling technology,” *Med. Phys.* **40**(10), 101902 (2013).

4. A. G. Haus and J. E. Cullinan, "Screen film processing systems for medical radiography: a historical review," *RadioGraphics* **9**(6), 1203–1224 (1989).
5. T. Kloss, G. Lautenschläger, and K. Schneider, "Advances in the process of floating borosilicate glasses and some recent applications for specialty borosilicate float glasses," *Glass Technol.* **41**(6), 177–181 (2000).
6. Y. Fujisaki et al., "Transparent nanopaper-based flexible organic thin-film transistor array," *Adv. Funct. Mater.* **24**(12), 1657–1663 (2014).
7. D. M. Gauntt and G. T. Barnes, "X-ray tube potential, filtration, and detector considerations in dual-energy chest radiography," *Med. Phys.* **21**(2), 203–218 (1994).
8. S. Yun et al., "Feasibility of active sandwich detectors for single-shot dual-energy imaging," *Proc. SPIE* **9033**, 90335T (2014).
9. R. Van Metter, "Describing the signal-transfer characteristics of asymmetrical radiographic screen-film systems," *Med. Phys.* **19**(1), 53–58 (1992).
10. R. Van Metter and R. Dickerson, "Objective performance characteristics of a new asymmetric screen-film system," *Med. Phys.* **21**(9), 1483–1490 (1994).
11. J. Yorkston, K.-L. Yip, and T. Wojcik, "Inventor. Apparatus for asymmetric dual-screen digital radiography," US Patent No. 7,834,321 (2006).
12. M. Rabbani, R. Shaw, and R. Van Metter, "Detective quantum efficiency of imaging systems with amplifying and scattering mechanisms," *J. Opt. Soc. Am. A, Opt. Image Sci.* **4**(5), 895–901 (1987).
13. R. M. Nishikawa, M. J. Yaffe, and R. B. Holmes, "Effect of finite phosphor thickness on detective quantum efficiency," *Med. Phys.* **16**(5), 773–780 (1989).
14. R. K. Swank, "Absorption and noise in x-ray phosphors," *J. Appl. Phys.* **44**(9), 4199–4203 (1973).
15. G. Lubberts, "Random noise produced by x-ray fluorescent screens," *J. Opt. Soc. Am.* **58**(11), 1475–1483 (1968).
16. A. R. Lubinsky et al., "Screen optics effects on detective quantum efficiency in digital radiography: zero-frequency effects," *Med. Phys.* **33**(5), 1499–1509 (2006).
17. A. Howansky et al., "Deriving depth-dependent light escape efficiency and optical Swank factor from measured pulse height spectra of scintillators," *Med. Phys.* **44**(3), 847–860 (2017).
18. A. Badano et al., "Lubberts effect in columnar phosphors," *Med. Phys.* **31**(11), 3122–3131 (2004).
19. R. K. Swank, "Calculation of modulation transfer functions of x-ray fluorescent screens," *Appl. Opt.* **12**(8), 1865–1870 (1973).
20. A. Howansky et al., "An apparatus and method for directly measuring the depth-dependent gain and spatial resolution of turbid scintillators," *Med. Phys.* **45**(11), 4927–4941 (2018).
21. A. Koch, "Influence of optical gaps on signal and noise properties of luminescent screen x-ray detectors," in *Paper presented at: Medical Imaging 2004* (2004).
22. K. Suzuki et al., "Fabrication of all-printed organic TFT array on flexible substrate," *J. Photopolym. Sci. Technol.* **24**(5), 565–570 (2011).
23. A. R. Lubinsky et al., "Substrate effect on indirect digital radiography system performance," *Proc. SPIE* **6142**, 61423J (2006).
24. G. Hajdok, J. J. Battista, and I. A. Cunningham, "Fundamental x-ray interaction limits in diagnostic imaging detectors: frequency-dependent Swank noise," *Med. Phys.* **35**(7), 3194–3204 (2008).
25. I. A. Cunningham, M. S. Westmore, and A. Fenster, "A spatial-frequency dependent quantum accounting diagram and detective quantum efficiency model of signal and noise propagation in cascaded imaging systems," *Med. Phys.* **21**(3), 417–427 (1994).

Anthony R. Lubinsky is a principal investigator at the State University of New York at Stony Brook. He received his PhD in physics from Northwestern University. His research interests are physics, imaging technology, and materials science for medical imaging systems.

Adrian Howansky has been a graduate student in DRIL Laboratory at Stony Brook University from 2013 to 2019, and is a medical physics resident at Stony Brook University Hospital Department of Radiology.

Hao Zheng was a graduate student of the Department of Biomedical Engineering at Stony Brook University. He received his master's degree in biomedical engineering on 2014. He is working as a lead optical systems engineer at Magic Leap for augmented reality.

Wei Zhao is a professor of radiology at the State University of New York at Stony Brook. She received her PhD in medical biophysics from the University of Toronto. Her research interests are medical image sensors and digital breast tomosynthesis.