Signal and noise characteristics induced by unattenuated x rays from a scintillator in indirect-conversion CMOS photodiode array detectors

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We report the measurement results of signal and noise characteristics induced by the direct x-rays in an indirect-conversion CMOS photodiode array detector. In order to isolate the signal and noise due to the direct x-rays from those due to the optical photons, we inserted a light-absorbing blackout material between a phosphor screen and the photodiode array. From the images irradiated when with and without the blackout paper, the signal and noise characteristics due to the optical photons emitted from a phosphor screen were also estimated. For the analysis of the measurements, we have developed a model describing the signal and noise transfers based on the cascaded linear-systems approach. The measured results show the direct x-ray is very harmful to the detector performances, such noise power spectrum (NPS) and signal-to-noise ratio (SNR). However, from the theoretical estimation, the degradation of NPS and SNR would not be due to the directly absorbed x-ray photons, but we believe that other sources, such as Compton scattered x-ray photons from a scintillator, are main causes.

Index Terms – Radiography, Image quality, Noise power spectrum, Cascaded system analysis
I. Introduction

As long as the additive electronic noise is not exceptionally large, the noise characteristic of an imaging detector system is mainly governed by stochastic variation of information-carrying quanta introduced in various image-forming states, such as, at a scintillator, if it is used, the interaction of x-ray, the absorption of its energy, the conversion of x-ray to optical photons, the spread of an optical photon burst, and the escape of optical photons, and whereunder a photodiode, the collection of optical photons, the conversion of optical photons to electronic charges, and the spread of electronic charges and so on. This stochastic variation increases the noise at the output, and their effect can be seen by examining a noise-power spectrum (NPS). In the previous study [1], we have measured the NPS of an indirect-conversion CMOS (complementary metal-oxide-semiconductor) photodiode array detector as reproduced in Fig. 1. According to Rossmann [2], as the spatial-frequency dependence of signal transfer is described by a modulation-transfer function or MTF\(f\), where \(f\) indicates the spatial frequency, NPS\(f\) would behave as \(MTF^2(f)\). However, as shown in Fig. 1, the normalized NPS (NNPS) of our detector system shows a behavior close to white spectrum except at the very low frequency band, which is due to the imperfect removal of fixed-pattern noise (FPN) at the gain-offset correction procedure. For a comparison, a fit function, directly proportional to \(MTF^2(f)\), is also plotted in Fig. 1, and it is observed that there is large discrepancy in high spatial-frequency region. From Lubberts [3], the NPS generally falls off more slowly at high frequency than the \(MTF^2(f)\) because of the stochastic variation caused by the depth-dependent interaction of x-rays within the scintillator and the fluctuation of processes at the corresponding depth. Badano et al. [4] demonstrated this Lubberts’ effect on a phosphor screen by estimating the Lubberts fraction, \(L(f) = MTF^2(f)/\text{NNPS}(f)\) based on Monte Carlo simulation. However, their result on the phosphor screen having a similar thickness of that employed in our detector is quite different. Our measurement and analysis shows much severer property than the expected Lubberts' effect,
which means that there is another factor affecting on the NPS degradation. In the previous study [1], it was pointed out that in a detector employing a thin phosphor the extra noise due to the direct absorption of x-ray photons in a photodiode is white. The direct x-rays are those are unattenuated from the phosphor screen and that directly interact with the photodiode. This additional white noise can further raise the spectral density in the high spatial-frequency band. The proof that the NNPS due to the direct x-rays is white has been shown in [5]. Considering this additional white noise, we analyzed the measured NNPS with a least-squares fit of the form $a\text{MTF}^2(f) + b$, where $a$ and $b$ are constants obtained from the numerical fit, and plotted the resultant fit in Fig. 1. The fit curve reasonably agrees to the measured NNPS.

As described, the signal and noise characteristics of the absorption of direct x-rays can affect on the detector performance, such as NPS, hence detective quantum efficiency (DQE). However, there has been no sufficient attention paid to this kind study. In this study, we report the measurements of signal and noise characteristics induced by the direct x-rays in the indirect-conversion CMOS photodiode array detector. The detailed measurement results are reported. And we discuss the measured results with the model, which describes the magnitudes of signal and noise at a photodiode due both to the absorption of indirect-conversion optical photons emitted from a scintillator and the absorption of direct x-ray photons, developed based on the cascaded linear-systems approach [5], [6].

II. Materials and Methods

A. Detector Preparation

The indirect-conversion CMOS photodiode array detector, simply we called CMOS detector, mainly consists of a two-dimensional (2D) array of CMOS photodiodes and an overlying scintillation phosphor screen. The CMOS photodiode array (RadEye™, Rad-icon Imaging Corp., CA, USA) has a format of 512
× 1024 pixels with a pitch of 48 μm [7]. Therefore, the active area is about 25 × 50 mm². As a scintillator, we employed two commercial phosphor screens, Min-R2000™ and Lanex Medium™, from Eastman Kodak Co. (Rochester, NY, USA). The phosphor screens are mainly made up of terbium-doped gadolinium oxysulfide (Gd₂O₂S:Tb) phosphors and their mass thicknesses (ρM) are 33.91 and 59.20 mg/cm², respectively [8]. The approximate thicknesses are 84 and 160 μm, respectively. For the complete CMOS detector, the Gd₂O₂S:Tb screen was directly overlaid onto the active area of the CMOS photodiode array and held by using a thin polyurethane foam layer for compression between the screen and a 1-mm-thick graphite cover.

When we measure the direct signal and noise, in order to isolate the signal and noise due to the direct x-rays from those due to the optical photons, we inserted a light-absorbing blackout material (Flock Paper #40, Edmond Optics Inc., Barrington, NJ, USA) between a phosphor screen and the photodiode array as described in Fig. 2. The thickness of the blackout paper is about 100 μm. The ability of light absorption of the blackout paper was confirmed by measuring light transparency as a function of wavelength in the range from 300 to 800 nm with a spectrometer (Spectrophotometer U-2010, Hitachi, Japan). Slight light transparency between 300 to 450 nm was observed, but the maximum light transparency was about 1% around the wavelength of 343 nm. For the wavelengths above 450 nm, the light was completely blocked.

It is noted that the main peak of an emission light spectrum of the Gd₂O₂S:Tb phosphor is about 545 nm.

B. Imaging Conditions

X-ray irradiation on the CMOS detectors has been carried out with respect to two imaging conditions; mammography and general diagnostic radiography. For the radiation qualities of each imaging condition, we followed the standard radiation qualities suggested by the IEC (International Electrotechnical Commission) [9], [10]. For the mammographic imaging condition, we took the W/Al characterization among various standard radiation quality characterizations [9]. The x-ray tube (Series
5000 Apogee, Oxford Instruments, USA), used for the W/Al mammographic imaging condition, is a sealed tube with a 125-µm-thick beryllium exit window. A tungsten target produces x-rays with energies ranging from 4 to 50 kVp at 50 W.

For the radiographic imaging condition, we took RQA 5 among various RQA (the radiation quality based on addition of a certain amount of aluminum filtration) series [10]. Among the various RQA series from 2 to 10, IEC 62220-1 [10] specifies four spectra such as RQA 3, 5, 7, and 9. If only one spectrum would be used, it should be RQA 5. For making RQA 5 spectrum, an x-ray tube having a rotating tungsten anode (E7239X, Toshiba, Japan) was used. The inherent filtration due to the internal tube structures and the glass of the beam collimator is 2.4 mm in aluminum equivalent.

In order to get both radiation qualities, we set up tube voltages and inserted the additional filters as suggested by IEC. Mismatch between the measured and suggested half-value layers (HVLs) was corrected by tuning the tube voltage slightly. The detailed radiation qualities used in this study are summarized in Table I.

Although IEC reports recommend that the source-to-detector distance should be 600–700 mm and at least 1500 mm for mammography and radiography, respectively, the distance was set to 640 mm in this study, which was inevitable in the given experimental constraints. The exposure rate at the entrance surface of the CMOS detector was measured by replacing it with a calibrated ion chamber (Victoreen 6000-528, Inovision, USA) while keeping the same distance. Since the readout time of the CMOS detector was fixed at 550 ms during the measurements, the exposure can be easily calculated. To take into account the dose, the measured exposure has been converted to the kerma in air with the exposure-to-kerma conversion factor [11]. The incident photon fluence \( \phi \) was calculated using the experimentally measured exposures and HVLs, and the computational program for x-ray spectral analysis [12] and the values corresponding to each imaging condition are summarized in Table I.
C. Analysis of the Measured Images

Since the signals due to the absorption of optical photons or direct x-rays at a photodiode are uncorrelated in each other, the mean and variance of an x-ray white image $I_{\text{white}}$ may be described as

\begin{align}
\text{Mean}[I_{\text{white}}] &= S_{\text{opt}} + S_{\text{direct}} + S_{\text{dark}}, \\
\text{Var}[I_{\text{white}}] &= \sigma^2_{\text{opt}} + \sigma^2_{\text{direct}} + \sigma^2_{\text{dark}} + \sigma^2_{\text{add}},
\end{align}

where $\text{Mean}[\cdot]$ and $\text{Var}[\cdot]$ indicate the ROI (region of interest) mean and variance for a 2D image data, respectively. The image signal in a given frame is a function of the optical signal from a phosphor screen, the direct-absorbed x-ray signal, and the dark current of the photodiode array. The image noise, on the other hand, is made up of optical photon shot noise, direct absorption noise, dark current noise, and additive read electronic noise. The subscripts $\text{opt}$, $\text{direct}$, $\text{dark}$, and $\text{add}$ denote each corresponding component contributing to the signal or noise. If we know each mean number of charge carriers $\bar{n}$ contributing to the corresponding signal and the mean conversion gain of the detector $G$ in ADU per electrons or $\text{ADU/e}^-$, the signal magnitude can be calculated by

\begin{equation}
S = \bar{G}\bar{n}.
\end{equation}

Similarly, if $\delta$ is the fluctuation in the number of generated charge carriers or the shot noise following the Poisson statistics ($\delta = \sqrt{\bar{n}}$), the magnitude of noise is then,

\begin{equation}
\sigma^2 = \bar{G}^2\delta^2.
\end{equation}

In this study, $\bar{G} = 0.002 \text{ ADU/e}^-$ from the given gain factors such as the digitization gain of 250
μV/ADU and the sensor gain of 0.5 μV/e⁻, which was obtained from the mean-variance method [7].

From the measurements of other images, such as an x-ray image when the blackout paper is used, simply we called a black image in this study, \( I_{\text{black}} \), and a dark image without any irradiation of x-ray, called an offset image, \( I_{\text{offset}} \), we can identify the magnitude of each signal and noise by the simple arithmetic operations of the calculated ROI means and variances:

\[
S_{\text{opt}} = \text{Mean}\{I_{\text{white}}\} - \text{Mean}\{I_{\text{black}}\}, \tag{3a}
\]
\[
\sigma_{\text{opt}}^2 = \text{Var}\{I_{\text{white}}\} - \text{Var}\{I_{\text{black}}\}, \tag{3b}
\]
\[
S_{\text{direct}} = \text{Mean}\{I_{\text{black}}\} - \text{Mean}\{I_{\text{offset}}\}, \tag{4a}
\]
\[
\sigma_{\text{direct}}^2 = \text{Var}\{I_{\text{black}}\} - \text{Var}\{I_{\text{offset}}\}. \tag{4b}
\]

Since we are not interested in systematic (deterministic), but in stochastic random noise, the analysis has actually been performed on the difference of two images obtained under identical conditions. Each difference image variance was corrected by dividing by 2 because the two images have independent noise samples with equal variance and the variance value for the difference image would be twice that of a single image [12].

Besides the analysis of the magnitude of signals and noises, we also investigated the noise characteristics in the spatial-frequency domain by calculating NPS from the measured image data. All processes associated with our NPS estimation algorithm were verified by observing if the total sum of noise power densities in the 2D NPS is equal to the variance of analyzing image data. From the radially symmetric feature of 2D NPS, we extracted a one-dimensional (1D) NPS in radial direction.

\textbf{D. Signal and Noise Characteristics Induced by Optical and Direct X-ray Photons}
In the configuration of indirect-conversion x-ray imaging detectors, a scintillator normally has a finite thickness so that the quantum absorption efficiency is not always perfect. Instead, direct interactions of unattenuated x-rays passing through a scintillator within the subsequently located photosensitive elements, for example, a photodiode array, may occur. Scattered x-rays from a scintillator directing to the photosensitive elements are another possible interactions. Figure 1 describes some of interactions to produce electronic signals in an indirect-conversion imaging detector, in which the contribution of high energy electrons from a photodiode passivation layer or bulk substrates is also described. X-rays produced in a scintillator and a bottom glass or ceramic substrate are also possible interactions, but not depicted. All the interactions depend upon the energy of the incident x-ray photons, and the thickness and material compositions of an interacting layer.

In this study, we only consider two interactions; the absorption of indirect-conversion optical photons emitted from a scintillator and the absorption of direct x-ray photons within a photodiode. For an impinging x-ray photon having an energy \( E \) to a scintillator having a thickness of \( L_{\text{scn}} \), the probability that the photon survives up to a depth of \( z \) without any interaction and then interacts in the additional distance \( dz \) and thus deposits its energy \( E \) can be described as

\[
e^{-\mu(E)z} \times \left[ \mu_{\text{ab}}(E) \right]_{\text{scn}} \times E \times dz.
\]

Here \( \mu \) and \( \mu_{\text{ab}} \) denote the total and energy-absorption linear attenuation coefficients, respectively. Assuming that the energy of the generated optical photons are monochromatic, hence accounting for the average energy required to generate an optical photon \( \Phi_{\text{scn}} \) and at a depth of \( z \) the escape probability of the generated optical photons \( p_{\text{esc}}(z) \), we can analytically estimate the mean number of electrons or charge carriers collected in a photodiode having an aperture size \( a \) in one direction such that

\[
\bar{n}_{\text{opt}} = a^2 \gamma_{\text{pd}} \int_0^{L_{\text{scn}}} \Phi(E) e^{-\mu_{\text{ab}}(E)z} \left[ \mu_{\text{ab}}(E) \right]_{\text{scn}} \frac{E}{\Phi_{\text{scn}}} p_{\text{esc}}(z) dEdz.
\]

(5)
Here, $\gamma_{pd}$ is the quantum efficiency of the photodiode, the peak applied x-ray tube potential is $E_{max}$, and $\Phi(E)$ is the differential x-ray photon fluence with energy between $E$ and $E + dE$. In a similar manner, the mean number of electrons induced by the absorption of direct x-rays within a photodiode can be given by

$$
\bar{n}_{direct} = a^2 \int_0^{E_{max}} \Phi(E) e^{-\frac{\mu(E)}{\mu_{pd}}} \left[ 1 - e^{-\frac{\mu(E)}{\mu_{pd}} \frac{E}{\bar{W}_{pd}}} \right] \frac{E}{\bar{W}_{pd}} dE \, ,
$$

where $\bar{W}_{pd}$ is the average ionization energy required to collect an electron-hole pair in a photodiode. However, these approaches are somewhat tedious to calculate and difficult to incorporate Monte Carlo simulation, which is a very excellent tool for estimating the radiation and optical transports in a complex geometry and sometimes replaceable to an experiment. In these regards, we applied the cascaded linear-systems theory instead, and thus (5) can be more simply expressed by [5]

$$
\bar{n}_{opt} = \bar{q} a^2 \alpha_{scn} \beta_{scn} \gamma_{pd} \, ,
$$

where $\alpha_{scn}$ describes the quantum absorption efficiency of a scintillator and $\beta_{scn}$ is the mean gain of optical photons, and those values can be estimated from the Monte Carlo simulations. $\bar{q}$ is the average incident photon fluence. The variance related to (7) can be expressed by [5]

$$
\delta_{opt}^2 = \bar{q} a^2 \alpha_{scn} \beta_{scn} \gamma_{pd} \left[ 1 + \gamma_{pd} \left( \frac{\beta_{scn}}{I_{scn}} - 1 \right) \right] \, ,
$$

where $I_{scn}$ is the Swank noise factor.

Similarly, the magnitudes of signal and noise due to the direct x-rays are given by [5]
\[ \pi_{\text{direct}} = \pi \alpha^2 (1 - \alpha_{\text{esc}}) \alpha_{\text{pd}} \beta_{\text{pd}}, \]  
\[ \delta_{\text{direct}}^2 = \frac{\pi \alpha^2 (1 - \alpha_{\text{esc}}) \alpha_{\text{pd}} \beta_{\text{pd}}^2}{I_{\text{pd}}}. \]

Here, \( \alpha_{\text{pd}} \) is the quantum absorption efficiency in a photodiode due to the direct x-ray and \( \beta_{\text{pd}} \) is the mean gain of secondary quanta, and \( I_{\text{pd}} \) is the Swank factor due to the charge generation by the absorption of direct x-ray photons within the photodiode.

**E. Monte Carlo Simulations**

In order to analyze the measured signal and noise characteristics of the CMOS detector, we have calculated the magnitude of signal and noise in theoretical approach based on the cascaded models. Although the physical parameters involved in the cascaded models can be analytically calculated, we estimated them basically based on the absorbed energy distributions (AEDs) and the optical pulse-height distributions (PHDs) obtained from the Monte Carlo simulations. We employed two Monte Carlo codes, MCNPX\textsuperscript{TM} (Version 2.5.0., ORNL, USA) and DETECT2000\textsuperscript{TM} (Laval University, Quebec, Canada) for x-ray and optical photon transports, respectively, in the detector.

In the x-ray simulation using the MCNPX\textsuperscript{TM} code, we considered a pencil beam incident perpendicularly onto the detector and we modeled the detector as cylindrical thin-slab geometries with infinite lateral dimension. Since the phosphor screens have usually small thickness, a diameter of 20 cm may be enough. Despite of the complexity of phosphor screens [8], we simply modeled into Gd\textsubscript{2}O\textsubscript{2}S monolayer with a calculated density based on the given thickness and mass thickness. The photodiode was modeled as a combination of a 2.5-\( \mu \)m-thick SiO\textsubscript{2} passivation layer plus a 700-\( \mu \)m-thick Si layer. In the Si layer, a thickness of 2 \( \mu \)m in the top region was defined as an active region considering \( p-n \) junction depth of the photodiode. In order to avoid the contamination of AED at the photodiode from the other sources such as Compton scattered x-rays from a phosphor screen, we separately simulated considering...
the attenuated x-ray spectra through a phosphor screen. It is noted that the source was sampled based on the spectra obtained from the computational program for x-ray spectral analysis [13] accounting for the radiation qualities.

For the optical model of a phosphor screen, it was regarded as a weakly absorbing medium in which scattering is due to Frensel reflection and refraction at boundaries between the phosphor grains and polyurethane polymer binders [14], and the average refractive index of 2.4 for two compositions was taken in the optical transport simulation with the DETECT2000™ code. Mean free paths of absorption and scattering were taken to be 1 and $4 \times 10^{-3}$ cm, respectively [4]. The side surface of a phosphor screen was defined as a metal surface with a reflection coefficient of zero, which implies that the optical photon escaping through the side surface never comes back and its simulation is terminated. The top surface was, on the other hand, defined as a polished surface with a reflection coefficient of 0.88 accounting for the reflectance of TiO₂, which is typically used in the phosphor screen design. The bottom surface was treated as a ground surface and the escaped optical photons were tallied at a virtual detection plane located 1 μm apart from the bottom surface to consider the real escaped photons. For various depth positions in the modeled screen, more than $10^8$ optical photons were generated isotropically and the escape probability as a function of depth position was estimated.

In order to estimate the quantum absorption efficiencies and the Swank factors both in a phosphor screen and the photodiode, the AEDs within each layer were obtained by using pulse-height tally. The quantum absorption efficiency was calculated by the zero-th moment of AED within a phosphor screen or the photodiode. In order to determine the mean gain of optical photons and Swank factor of a phosphor screen, the AEDs for various depth positions in the modeled screen were obtained. Each energy bin of the AED scaled to its corresponding spectral probability is then divided by the effective $W$-value of the screen, and which becomes the generated optical photons for a given depth position. If we weights on the generated optical photons with the escape efficiency for the corresponding depth position, we can obtain
the detected optical photons. This number is then converted to the PHD based on the approximation that
the detected optical photons are Poisson distributed. The Poisson distribution can be modeled using
normal Gaussian distributions with the mean and variance. The superposition of all the PHDs for the
entire energy bins of the AED with respect to overall depth positions creates a single composite PHD.

Taking the first moment of this final PHD, we determine the mean gain of optical photons. The Swank
factor of a phosphor screen is, on the other hand, determined by

\[ I_{\text{scn}} = \frac{\left[ \sum_n \text{PHD}(n) n \right]^2}{\sum_n \text{PHD}(n) \sum_n \text{PHD}(n) n^2} \]  \hspace{1cm} (11)

where \( n \) is the pulse-height bin of PHD. The Swank factor due to the direct x-ray absorption within the
photodiode was determined with the same approach, but the AED obtained at the photodiode was used
instead of PHD in (11).

III. Results

Figure 4 shows the measured signal variance due to the direct x-ray photon absorption as a function
of the corresponding mean signal value. In spite of the different screen usages, the mean-variance
tendency at the RQA 5 radiographic imaging condition is well described by a single first-order least-
squares fit curve. Since the number of data is not enough to be analyzed, the data for the mammography
imaging condition are excluded from the fitting analysis. The incremental slope of the mean-variance
curve is 7.4 ADU. From the equations (9) and (10), and considering the mean conversion gain of the
detector, the theoretical slope is given by \( \frac{\beta_{\text{scn}}}{I_{\text{scn}}} \), and the estimated slopes of the Min-R2000\textsuperscript{TM} and
Lanex Medium™ screens are 8.4 and 8.3 ADU.

As shown in Fig. 5, the dependency of the estimated mean-variance plot for the optical photon absorption on what kind of a phosphor screen is used is apparent. If the ratio of $\beta_{\text{esc}}$ and $I_{\text{esc}}$ is much larger than unity, the incremental slope of a mean-variance curve is approximately proportional to

$$\frac{\eta_{\text{pd}}}{\beta_{\text{esc}}} \frac{\beta_{\text{esc}}}{I_{\text{esc}}}$$

from (7) and (8). From the Monte Carlo simulations, the slope of the Min-R2000™ screen is larger than that of the Lanex Medium™ one by a factor of 1.16, which is mainly due to the better escape efficiency of a thinner phosphor screen, hence the mean gain of optical photons. According to Fig. 5, however, the difference between the incremental slopes of the mean-variance curves is much larger; the slope of the Min-R2000™ screen is more than 8.5 times larger than that of the Lanex Medium™ one. This result probably implies that the measured variance is not due to the pure optical photon absorption.

For the overall measurements, it should be noted that while the optical signals are much larger than the direct signals, the impact of the related variance is much server in the direct signals.

Figure 6 shows an example of the calculation results compared with the measured data. Although the discrepancy between the calculations and measurements gradually increases as the air kerma increases, the calculation results of signals reasonably describes the measured data as shown in Fig. 6(a). In the case of variances, however, the componential analysis is quite different. While the measurements show that the variances due to the direct x-rays are larger than those due to the optical photons, the calculations show a reversed result against the measurements. In addition, it is interesting to note the results of signal-to-noise (SNR). Both the measurements and calculations show that the total SNRs are less than those only due to the optical photons.

For the use of Min-R2000™ screen, noise power spectra normalized by the air kerma are plotted in Fig. 7. In order to remove the FPN, the NPSs were calculated for the difference images (divided by 2). It is noted that the NPSs due to the direct x-rays are white with respect to the spatial frequency as reported in [5]. As already described, the effect of the energy of irradiation x-ray is apparent. Figure 8 shows the
dependency of the NPSs on the phosphor screen thickness. The NPSs due to both the optical and direct x-ray photons well reflect the typical performances of the screen type. If the phosphor screen is thinner, the contribution of the direct-absorbed x-rays becomes larger.

IV. Discussion

As described in Fig. 3, there are a number of interactions contributing to the detector signal. We tried to measure the signal and noise due to the direct x-rays by inserting the blackout paper. We could probably block the optical photons emitted from a phosphor screen, but we could not block other sources, for example, the Compton scattered x-rays, and which may be the cause why the signal and variance due to the direct x-ray photons are larger than those expected by the theoretical model. The difference between the measurements and the calculations of the signal and variance due to the optical photon absorption can also be understood by the same manner. Figure 9 shows the AEDs within a p-n junction layer obtained from the Monte Carlo simulations for three different Monte Carlo geometries; a phosphor screen plus a 2-μm-thick p-n junction photodiode made on a 700-μm-thick Si bulk substrate, a p-n junction photodiode made on the Si substrate, and only a 2-μm-thick p-n junction photodiode. Compared with the p-n junction photodiode geometry, if we consider a combination of a phosphor screen and the silicon photodiode in the Monte Carlo geometry, much larger energy-absorption events are recorded (about 13.5 times larger), which implies that there are a large amount of contributions from the overlying phosphor screen except the direct x-ray absorption events. When considering a p-n junction photodiode made on the Si substrate geometry, the overall distribution is much reduced, but still larger energy-absorption events are recorded (about 1.7 times larger) compared with only the p-n junction photodiode geometry, which implies that there are signal contributions from the overlying SiO₂ layer and the underlying silicon bulk substrate to an active region. From the estimation based on the theoretical calculations, we believe that the effect of the absorption of direct x-rays is not crucial as much as the
measurements as demonstrated in Fig. 6.

The developed cascaded linear-systems analysis based on two independent signal and noise propagation in a detector is not sufficient to fully describe the measurements, but quite promising. If more detailed and significant processes describing how image information passes along the image chain within a detector are incorporated in the model, it provides a useful ability to predict and optimize a detector performance.

V. Conclusion

By inserting a blackout paper between a phosphor screen and a photodiode array, we have tried to measure the signal and noise characteristics induced by the direct x-ray photons unattenuated from a phosphor screen. And from the x-ray images obtained with and without the blackout paper, the signal and noise due to the optical photons emitted from a phosphor screen have been estimated. From the measurement results, although the magnitude of direct signal is small, its variance is significantly large, which gives rise to the reduction of total SNR of an image. On the contrary, from the theoretical estimation, the effect of direct x-ray absorption is negligible. From the Monte Carlo simulations, we conclude that the main cause of the degradation in noise characteristics and thus the reduction in SNR is not by the direct x-rays but by the other sources, such as the Compton scattered x-rays from a phosphor screen.
ACKNOWLEDGEMENT

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REFERENCES


Table and Figure Captions

TABLE I. The standard radiation qualities and characteristics used in this study.

Figure 1. Normalized noise power spectrum of an indirect-conversion CMOS imaging detector reproduced from the previous study [1]. The dashed and dotted lines describe least-squares fit curves to analyze the measured noise power spectrum.

Figure 2. A sketch describing an experimental layout for the measurement of direct x-ray signals by inserting a blackout paper between a phosphor screen and a photodiode array.

Figure 3. Schematic illustration of possible interactions to produce electronic signals in an indirect-conversion imaging detector, which is simply described as a Gd$_2$O$_2$S:Tb phosphor screen, a SiO$_2$ passivation layer, and a Si layer. PE and CS indicate the photoelectric absorption and Compton scattering events, respectively.

Figure 4. Mean-variance curve due to the direct x-ray photon absorption. The solid line is a first-order least-squares fit. It is noted that the data for the mammography imaging condition was excluded from the fitting analysis.

Figure 5. Mean-variance curves due to the optical photon absorption, which were estimated from the x-ray images obtained with and without the blackout paper. It is noted that the data for the mammography imaging condition was excluded from the fitting analysis.

Figure 6. Comparisons between the measured data and the theoretical calculations. (a) Signal, (b) variance, and (c) SNR.

Figure 7. Measured NPSs (normalized by the air kerma) with respect to two different imaging conditions.
Figure 8. Measured NPSs (normalized by the air kerma) with respect to two different phosphor screens.

Figure 9. Absorbed energy distributions within a $p$-$n$ junction layer obtained from the Monte Carlo simulations for three different Monte Carlo geometries; a phosphor screen plus a 2-$\mu$m-thick $p$-$n$ junction photodiode made on a 700-$\mu$m-thick Si bulk substrate, a 2-$\mu$m-thick $p$-$n$ junction photodiode made on a 700-$\mu$m-thick Si bulk substrate, and only a 2-$\mu$m-thick $p$-$n$ junction photodiode.
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<th>IEC filtration</th>
<th>IEC HVL</th>
<th>Measured HVL</th>
<th>IEC Tube voltage</th>
<th>Adjusted voltage</th>
<th>Calculated $q$</th>
<th>Measured $q$</th>
<th>Percent difference</th>
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<td>30,174</td>
<td>29,707</td>
<td>1.55 %</td>
</tr>
</tbody>
</table>
Figure 1

NNPS (mm²)

Spatial Frequency (mm⁻¹)

b

a x MTF²(f) + b

c x MTF²(f)

Measured
Figure 2
Figure 3
Figure 4

Variance (ADU²) vs. Mean Signal (ADU)

- Direct x-ray absorption
- Min-R2000™ screen
- Lanex Medium™ screen

Mammography
Radiography
Figure 5
Min-R2000™ phosphor screen
Radiography imaging condition

Figure 6
Figure 7
Figure 8

Radiography imaging condition

- Lanex Medium™ (Optical + Direct)
- Min-R2000™ (Optical + Direct)
- Min-R2000™ (Direct)
- Lanex Medium™ (Direct)
Figure 9

- PD coupled to a phosphor screen
- At the p-n junction of silicon substrate
- When considering only p-n junction layer