Aging of imaging properties of a CMOS flat-panel detector for dental cone-beam computed tomography

This content has been downloaded from IOPscience. Please scroll down to see the full text.
2017 JINST 12 P01005
(http://iopscience.iop.org/1748-0221/12/01/P01005)

View the table of contents for this issue, or go to the journal homepage for more

Download details:

IP Address: 164.125.73.25
This content was downloaded on 06/01/2017 at 00:51

Please note that terms and conditions apply.
Aging of imaging properties of a CMOS flat-panel detector for dental cone-beam computed tomography

D.W. Kim, J.C. Han, S. Yun and H.K. Kim

School of Mechanical Engineering, Pusan National University, Busan 46241, Republic of Korea
Center for Advanced Medical Engineering Research, Pusan National University, Busan 46241, Republic of Korea
E-mail: hokyung@pusan.ac.kr

ABSTRACT: We have experimentally investigated the long-term stability of imaging properties of a flat-panel detector in conditions used for dental x-ray imaging. The detector consists of a CsI:Tl layer and CMOS photodiode pixel arrays. Aging simulations were carried out using an 80-kVp x-ray beam at an air-kerma rate of approximately 5 mGy s$^{-1}$ at the entrance surface of the detector with a total air kerma of up to 0.6 kGy. Dark and flood-field images were periodically obtained during irradiation, and the mean signal and noise levels were evaluated for each image. We also evaluated the modulation-transfer function (MTF), noise-power spectrum (NPS), and detective quantum efficiency (DQE). The aging simulation showed a decrease in both the signal and noise of the gain-offset-corrected images, but there was negligible change in the signal-to-noise performance as a function of the accumulated dose. The gain-offset correction for analyzing images resulted in negligible changes in MTF, NPS, and DQE results over the total dose. Continuous x-ray exposure to a detector can cause degradation in the physical performance factors such the detector sensitivity, but linear analysis of the gain-offset-corrected images can assure integrity of the imaging properties of a detector during its lifetime.

KEYWORDS: Computerized Tomography (CT) and Computed Radiography (CR); X-ray radiography and digital radiography (DR)

They are now affiliated with the Samsung Electronics Co., Suwon, Gyeonggi-do 16677, Republic of Korea.
Corresponding author.
1 Introduction

Electronic flat-panel detectors (FPDs) are applied in x-ray radiography [1, 2] as well as dynamic and tomographic imaging [3, 4]. Consistent imaging properties are therefore important for their reliable use in clinical practice, even with “aging” due to continuous x-ray exposure. Amorphous silicon (a-Si)-based FPDs are known to be resistant to radiation damage and can be applied to megavoltage portal imaging [5, 6]. However, they have other types of instability over time such as ghosting or image lag due to the charge trapping/detrapping mechanism [7].

Complementary metal-oxide-semiconductor (CMOS)-based FPDs are based on the crystalline silicon process for realizing photodiode/transistor pixel arrays, which results in less charge-trapping density in pixel photodiodes and facilitates a finer design of the pixel architecture [8]. CMOS FPDs have many advantages over a-Si FPDs, such as smaller pixel pitch, higher fill factor, low electronic noise, high-speed readout, and less image lag [9, 10]. However, CMOS FPDs can also suffer from ghosting due to variations in the electrical characteristics of pixel transistors because of x-ray-induced charge-buildup effects within the oxide layers [11].

Similar to conventional indirect-conversion FPDs, CMOS FPDs are completed by overlaying the CMOS pixel array with a scintillation layer (typically CsI:Tl or Gd$_2$O$_2$S:Tb), which is used for converting incident x-ray quanta into optical quanta. The scintillator can reduce the radiation effect on the array. An additional use of a fiber-optic faceplate between the scintillator and the pixel array can further reduce the radiation effect by stopping the x-ray quanta unattenuated from the
Figure 1. Theoretical estimation of the fractional energy deposition within the CMOS photodiode layer compared to the CsI:Tl layer: (a) relative energy deposition within the photodiode layer as a function of energy for $t_{\text{CsI}} = 0.5$ mm and (b) relative energy deposition within the photodiode layer as a function of $t_{\text{CsI}}$ for an 80-kVp x-ray spectrum. We assume that $t_{\text{PD}} = 0.01$ mm. The $\bigcirc$ symbol in (a) indicates the calculation result for the mean energy of the spectrum, and the $\times$ symbol in (b) indicates the calculation result for $t_{\text{CsI}} = 0.5$ mm.

Scintillator [12, 13]. However, the scintillator itself can also be damaged by radiation exposure [14, 15], which could change the optical quantum yield.

Since wafer-scale large-area CMOS FPDs have emerged [16–18], their applications have been expanding. CMOS FPDs are particularly suitable for dental cone-beam computed tomography (CBCT) applications [19, 20]. CMOS FPDs are also anticipated for use in C-arm systems for interventional radiology [21, 22]. While these applications involve heavy x-ray irradiation, however, there has not been sufficient effort investigating performance degradation of CMOS FPDs due to radiation damage over their lifetime.

We investigated the long-term stability of imaging properties of a CMOS FPD for dental CBCT, including the dark and x-ray flood-field image signals and their standard deviations (i.e., noise). To examine changes in the detector gain or sensitivity, the characteristic function was analyzed as a function of the accumulated dose up to 0.6 kGy of air kerma at the entrance surface of the detector. We also assessed the modulation-transfer function (MTF), Wiener noise-power spectrum (NPS), and detective quantum efficiency (DQE).

2 Materials and methods

2.1 Direct-energy deposition in the CMOS pixel array

X-ray photons that are unattenuated by the scintillator can affect the CMOS pixel array. This depends on the finite thickness of the scintillator, which is chosen based on the trade-off between x-ray absorption and optical quantum scattering. However, the probability of this direct interaction of x-ray photons with the array is low because the electronic pixel components have small thickness ($\sim \mu m$).

We assume that only photoelectric interactions result in energy absorption events, and all the x-ray quantum energy is deposited. The energy deposition within the photodiode layer relative to
that within the CsI:Tl layer can then be estimated by
\[
\frac{\epsilon_{PD}}{\epsilon_{CsI}} = \frac{\int_0^{E_{max}} \Phi(E) [1 - \alpha_{CsI}(E)] \sigma_{PD}(E) \zeta_{PD}(E) dE}{\int_0^{E_{max}} \Phi(E) \alpha_{CsI}(E) \zeta_{CsI}(E) dE},
\]
(2.1)
where
\[
\alpha_j(E) = 1 - e^{-\mu_{tot}^j(E)t_j}
\]
(2.2)
and
\[
\zeta_j(E) = \frac{\mu_{pe}^j(E)}{\mu_{tot}^j(E)}.
\]
(2.3)
Equations (2.2) and (2.3) represent the quantum efficiency and the probability of photoelectric interaction over the total interactions in the \(j\)th layer (CsI and PD for the CsI:Tl and photodiode layers, respectively). \(\mu_{tot}\), \(\mu_{pe}\), and \(t\) denote the total attenuation coefficient, photoelectric coefficient, and thickness of the layer of interest. \(\Phi(E)\) denotes the incident x-ray photon spectrum.

Figure 1(a) shows the relative energy deposition within the photodiode layer as a function of the x-ray photon energy. We assume that \(t_{CsI} = 0.5\) mm (with a packing density of 80%) and \(t_{PD} = 0.01\) mm. Figure 1(b) shows the results for an 80-kVp tungsten spectrum as a function of \(t_{CsI}\). The energy deposition in the photodiode layer is less than \(10^{-4}\) times that in the 0.5-mm-thick CsI:Tl layer. These results suggest that the radiation effect on the photodiode layer due to direct x-ray photons would be negligible compared to that on the CsI:Tl layer.

### 2.2 Experimental setup

We obtained a CMOS FPD (Xmaru1215CF) developed for commercial dental CBCT systems from Rayence Co., Ltd. (Republic of Korea). The detector consists of a CsI:Tl scintillator and CMOS pixel arrays (photodiode and transistors). The thickness of the CsI:Tl is approximately 0.5 mm according to the manufacturer. The CMOS pixel array panel is tiled with four small unit panels,
each of which is three-side buttable and features 0.2-mm pixels arranged in a 300 × 360 format. These provide a field of view (FOV) of ∼ 120 × 150 mm². The maximum readout speed of the detector is 30 frames per second.

As shown in figure 2, the detector was mounted on an optical bench facing the beam exit of a tungsten-anode x-ray tube (EXG-6, The Rayence Co., Ltd., Republic of Korea). The distance between the x-ray focal spot and the detector (dSD) was variable. The field size of the irradiating x-ray beam was larger than the FOV of the detector, but the region for simulating radiation effects was limited to only one-quarter of the FOV, which corresponds to the size of a single unit panel. This was accomplished by blocking the remaining area with a 1-mm-thick lead plate. These two distinct regions are called the “open-field” and “shielded” regions, respectively. These regions were used to compare the dark signal and noise characteristics obtained from damaged and non-damaged regions of the detector.

The effects of radiation on a device are usually evaluated in terms of the total ionizing dose, which is the accumulated energy per unit mass of the device [23]. In this study, however, the air kerma (Gy) at the entrance surface of the detector was used instead to describe the total dose. The air-kerma rate (Gy s⁻¹) at the detector was measured with a calibrated ion chamber (Piranha R&F/M 605, RTI Electronics AB, Sweden), and the total dose was estimated by multiplying the air-kerma rate by the x-ray beam irradiation time.

2.3 Scenario for aging and characterization of the detector

The detector was heavily irradiated at dSD = 300 mm by the x-ray tube operating at 80 kVp and 9 mA, which yielded a kerma rate of 4.82 mGy s⁻¹. The irradiation continued until the total dose K reached 0.6 kGy. During the aging simulation, the dark images were obtained at every incremental step, ΔK₁ = 1 Gy up to a total dose of K = 0.2 kGy. Images were then obtained using ΔK₁₀ = 10 Gy for the K range of 0.2 kGy to 0.6 kGy. The flood-field and edge-phantom images were obtained at every ΔK₁₀ = 10 Gy up to K = 0.5 kGy and at ΔK₂₀ = 20 Gy for the remaining range. The edge-phantom images were used for the MTF analysis, and the flood-field images were used for the detector response and NPS analysis.

To mimic a typical setup for dental x-ray imaging, dSD was changed to 700 mm and the beam current was reduced to 7 mA when obtaining the flood-field and edge-phantom images. To consider the beam attenuation through the head and neck, aluminum plates with a total thickness of 21 mm were placed near the x-ray beam exit. With this setup, the air-kerma rate at the detector was measured as 45.69 µGy s⁻¹.

2.4 Analysis of imaging properties

2.4.1 Large-area signal and noise analysis

The dark image signal d at a given K was analyzed in 256×256 pixel regions for both 100 open-field and shielded region images. The corresponding dark image noise σd was determined from the standard deviation of 256 × 256 pixel values for ∼100 images, each of which was a subtraction image of two images divided by 2, thereby isolating only stochastic random noise in the subtracted image [24]. Similarly, the flood-field image signal s and noise σs at the air kerma x = 1.1 µGy were analyzed for ∼100 gain-offset-corrected images. The signal-to-noise ratio (SNR) for the
flood-field images is denoted by SNR. It is noted that the gain-offset correction of flood-field images is a standard procedure in digital imaging to avoid contamination in flood-field images from nonstochastic processes, and it is based on normalization with an average reference flood field (or flat field) [25–27]. The offset correction accounts for the nonuniform dark current effects in pixels, while the gain correction accounts for the nonuniform response of pixels.

The detector responses or characteristic functions at a given \(K\) were evaluated by measuring the average pixel values in \(256 \times 256\) pixel regions for an air kerma \((x)\) ranging from \(\sim 0.53 \mu\text{Gy}\) to \(1.98 \mu\text{Gy}\) (equivalent to \(\sim 0.06 \text{mR}\) to \(0.23 \text{mR}\)). Because the detector responses were not linear with respect to \(x\), as shown in figure 4(a), they were analyzed using a power-law model [28]:

\[
s(x; K) = g_1(K)x^{g_2(K)},
\]

where \(g_1\) and \(g_2\) are fit parameters. \(g_2\) represents the degree of linearity in the detector response, with unity indicating perfect linearity. \(g_1\) represents the detector gain in units of digital numbers per air kerma (DN \(\mu\text{Gy}^{-1}\)). The overall aging of the detector due to radiation damage can be parametrized by the gain change relative to the initial gain over the testing period [29]:

\[
R \equiv \frac{1}{g_1(0)} \frac{dg_1(K)}{dK},
\]

where \(g_1(0)\) is the detector gain measured at \(K = 0\) Gy.

The effect of the nonlinear detector response in flood-field images can be removed by converting the images into those at detector input using the inverse of eq. (2.4). Hence, the images are represented in units of air kerma (\(\mu\text{Gy}\)) or x-ray fluence \(\bar{q}\) (\(\text{mm}^{-2}\)) instead of DN. Then, the image noise and SNR for the linearized images are

\[
\sigma_q^2(K) = \left(\frac{\partial s(K)}{\partial \bar{q}}\right)^2 \sigma_q^2(K) = \left(\frac{x\bar{q}_0}{g_2(K)s(K)}\right)^2 \sigma_q^2(K),
\]

and

\[
\text{SNR}_q(K) = g_2(K)\text{SNR}_s(K),
\]

where \(\bar{q}_0\) is the incident x-ray fluence per unit air kerma (\(\text{mm}^{-2} \mu\text{Gy}^{-1}\)), which can be estimated readily using the energy-dependent x-ray fluence per unit air kerma using Boone’s model [30].

### 2.4.2 Fourier analysis

The MTF was analyzed using the edge-phantom images. To obtain the aliasing-free MTF, the over-sampled edge-spread functions (ESFs) were obtained from the slanted edge-phantom images. The ESFs were then differentiated to obtain line-spread functions (LSFs). The MTF was finally obtained by applying a fast Fourier transformation to the LSFs.

The NPS was obtained from the 2D Fourier analysis of the dark and flood-field images. The dark NPS was evaluated for the difference between two dark images, and the resultant dark NPS was corrected by dividing it by 2 [24].

These Fourier analyses of the imaging properties can only be applicable to a detector with a linear and shift-invariant (LSI) response and stationary noise. However, the investigated detector did not satisfy the LSI condition. Therefore, the MTF and NPS were analyzed using images that
were linearized with the inverse of eq. (2.4) [28]. The DQE is obtained from the linearized MTF and NPS:

\[ \text{DQE}(u) = \frac{x \tilde{q}_0 \text{MTF}^2(u)}{\text{NPS}(u)}. \]  

(2.8)

A single-valued metric instead of the spatial-frequency-dependent MTF is helpful for monitoring spatial resolution changes as a function of \( K \). We used the effective aperture [31] to describe the MTF change with respect to \( K \):

\[ a_{\text{eff}}(K) = \left[ 2\pi \int_{0}^{\infty} \text{MTF}^2(f; K) f df \right]^{-1}, \]  

(2.9)

where \( f = \sqrt{u^2 + v^2} \), and \( u \) and \( v \) are the Fourier conjugates of the \( x \) and \( y \) coordinates, respectively.

3 Results

3.1 Large-area results

Figure 3 summarizes the large-area signal and noise variations as functions of \( K \). The error bars indicate the standard deviation of 100 independent measurements. Figure 3(a) compares the dark image signals obtained from the respective open-field and shielded regions without any preprocessing. The dark signal in the shielded region was independent of \( K \), but that in the open-field region declined slightly as \( K \) increased (decreasing by \( \sim 8\% \) at \( K = 0.6 \) kGy compared to the initial level). Figure 3(b) shows the dark image noise as a function of \( K \). The fixed-pattern noise
Figure 4. (a) Detector response functions as functions of input dose with respect to various total input dose levels. (b) Changes in the gain and linearity of the detector response as functions of total input dose. (c) Relative rate of change of the detector gain as a function of total input dose.

Figure 5. Results calculated from linearized flood-field images as functions of total input dose: (a) image signal, (b) noise, and (c) SNR. (c) compares image SNR performances measured at the detector output and the detector input. The solid line in (c) indicates the theoretical estimation of the SNR for the linearized images using the average linearity [see figure 4(b)] and the average SNR (refer to the dotted line) measured at the detector output.

As shown in figure 3(c), the signal in the gain-offset-corrected flood-field images decreased rapidly in the early stage of the aging simulation, and then the decline became more gradual. The signal reached ~ 80% of its initial value at $K = 0.6$ kGy. The corresponding noise also decreased by ~ 20% over the total dose accumulation of 0.6 kGy, as shown in figure 3(d). Therefore, the SNR is constant for $K$ [see figure 5(c)].

Figure 4(a) shows the characteristic curves of the detector for various $K$ levels. The nonlinear detector responses were assessed by least-squares regression analysis with eq. (2.4), and the $R^2$ values were greater than 99.9% for all the detector response data measured over the entire range of $K$. As $K$ increased, the overall detector gain ($g_1$) decreased. This is shown more clearly in the plot of $g_1$ versus $K$ in figure 4(b), where $g_1$ decreases exponentially with increasing $K$. The aging parameter defined by eq. (2.5) is plotted in figure 4(c), where the dotted line is the analytical estimation from the eq. (2.5) with an exponential decay function describing the $g_1(K)$ data in figure 4(b). The aging parameter was negative, as expected from figure 3(c). The gain loss due to radiation damage was relatively large in the early period of the aging simulation ($K < 0.2$ kGy), and it was marginal in the rest of the period. As shown in figure 4(b), the nonlinearity in the detector response was not affected by the radiation damage ($\bar{g}_2 = 1.24$ over the entire range of $K$).
Figures 5(a) and 5(b) show the linearized image signal and noise variations as functions of $K$. The noise increased slightly in the aging simulation for the linearized images, whereas the linearized image signal was independent of $K$. These characteristics were well reflected in the SNR performance, as shown in figure 5(c). As expected from eq. (2.7), the SNR for the linearized images is larger by a factor equal to $g_2$ compared to that for the output images. Without correction for the nonlinear detector response, the DQE performance will be underestimated by a factor of $g_2$ squared.

3.2 Fourier results

The MTF curves measured at $K = 0$ Gy and 0.6 kGy are compared in figure 6(a), which shows a negligible difference between them. To investigate the MTF changes with $K$, the effective aperture was calculated using eq. (2.9) at every $\Delta K$. As shown in figure 6(b), the effective aperture was nearly independent of $K$, and the average value for the entire range of $K$ was $\sim 0.45 \text{ mm}^2$ (about 11 times larger than the pixel size). These observations show that the MTF performance is not affected by radiation damage.
The effects of heavy irradiation on the NPS are illustrated in Figure 7. As expected from figure 3(b), the dark image NPS was not affected by $K$, as shown in figure 7(a). Figure 7(b) shows that the noise-power spectral densities over the entire spatial-frequency region for the linearized flood-field images increased slightly in the early period of the aging simulation ($K < 0.2$ kGy). However, the degree of increase was insignificant.

The linear analysis results of DQE with respect to $K$ are presented in figure 8. The DQE($u$) performance was mainly governed by the NPS($u$) performance. DQE degraded slightly in the early period of the aging simulation ($K < 0.2$ kGy) over the whole range of spatial frequencies. Figure 8(b) shows the DQE value measured at a spatial frequency of 0.05 mm$^{-1}$ as a function of the total dose. The aging does not severely affect the DQE performance.

### 4 Discussion

Although we assumed that the CsI:Tl layer was thick enough to prevent the x-ray photons from absorbing into the photodiode pixel array, we observed a small decrease (~8% over the total dose) in the dark image signal with increasing $K$, as shown in figure 3(a). This may imply that there are interactions in the photodiode pixel arrays during aging simulations, and the interactions are probably due to secondary rather than primary x-ray photons. These secondary x-rays scatter from the CsI:Tl layer or other parts of the electronics, including the substrate and the detector housing. Using a Monte Carlo technique, Yun et al. [12] showed more than 10 times larger energy-absorption events in the photodiode layer when it was coupled to a phosphor screen compared to a bare photodiode layer.

The decreasing trend of the dark image signal with the irradiation contradicts the observations of a previous study [11], in which the dark leakage current increased with the accumulated dose. The reason for this contradiction is unclear. Antonuk et al. [32] compared the leakage current behaviors of two different photodiodes in conjunction with the same phosphor screen as functions of the delivered dose. They found no definite trend in the leakage current behavior and observed the same contradiction that we did.
The gain of the CMOS FPD decreased with irradiation, and we suspect that the degradation of the CsI:Tl is the main cause. The performance degradation of polycrystalline scintillators due to x-ray irradiation has not been investigated much. Van Eijk [14] indicates that radiation damage increases the optical quantum yield in CsI:Tl. On the contrary, Mori et al. [33] reported a 25% decrease in x-ray sensitivity of a 150-µm-thick CsI:Tl-coupled CMOS FPD using 200-kR x-ray irradiation from a tungsten target at 80 kVp. Another study similarly reported an 8% decrease using 16-kR x-ray irradiation at 30 kVp for a 165-µm-thick CsI:Tl [34]. For crystalline CsI:Tl scintillators, various studies reported a drop in optical quantum yield between 7% and 25% after ∼10 kGy [35]. This was attributed to the formation of color centers, which attenuate the number of optical quanta with certain wavelengths [36, 37]. High doses could also break down the scintillation mechanism, resulting in fewer optical quanta generated per interacting x-ray photon [36, 37].

The aging simulation showed decreases in both the image signal and noise, but the change in the SNR as a function of the accumulated dose was negligible. However, the decrease in signal could reduce dynamic range in images. This would be extremely significant for clinical imaging applications for the obvious reason of the loss of x-ray intensity range after attenuation by human anatomy.

Although continuous radiation exposure can gradually change the signal and noise level in images, gain-offset correction can provide stable image SNR performance in spite of radiation damage. Notably, nonlinearity of the detector response reduces the actual SNR performance. CMOS active-pixel designs have nonlinear response characteristics [38]. The reported offset values from linear assessments of CMOS FPD responses imply nonlinearities in small-signal responses. Han et al. [28] and Zhao et al. [39] discuss the causes in detail. In brief, the signal nonlinearity can originate from the varying conversion gain of the active pixels. This gain is inversely proportional to the capacitance of the photodiode, whereas the capacitance depends on the optical quanta-induced electronic signal generated in the photodiode, resulting in signal nonlinearity. If the detector response can be described by a power function, as in this study, the exponent represents the degree of linearity. This study shows that the actual SNR performance degrades in proportion to the degree of nonlinearity of the detector response. Hence, the DQE will be underestimated in proportion to the square of the nonlinearity factor (or $g^2$).

A total dose of 0.6 kGy corresponds to more than 60,000-patient examinations, assuming conservatively that the air kerma at the detector entrance surface is 10-µGy (corresponding to >1 mR) per frame for dental CBCT examinations and that the image reconstruction requires 1.000 projections. If 50 patients are scanned per day in a local hospital, the simulation would be equivalent to more than 3 years. The original goal was to simulate detector aging for up to 1 kGy, but an unexpected power failure occurred around $K = 0.62$ kGy. The detector still operated after the power failure and the results were consistent (however, the power failure caused the dark signal value to level down slightly). Nevertheless, we reported the results for $K$ less than 0.6 kGy.

The degrees of aging and recovery from radiation damage are dependent upon the beam quality used for irradiation such as energy and fluence rate. Compared to the air-kerma rate of ~45 µGy s$^{-1}$ for dental imaging in this work, the air-kerma rate used for aging simulations was 100 times larger. Furthermore, unlike this study, a detector system in the clinical environments can be allowed to have a recovery time from radiation damage because the workload for the detector system is much less than a single day. Therefore, this study may not describe correctly the aging of imaging properties.
of the CMOS FPD in clinical dental CBCT. Irradiation rate-dependent analysis accounting for variation in recovery times is remained as a further study. Most of all, an aging study with only one sample detector may not justify the observations in this work, and general conclusions cannot be made about the imaging properties of such detectors from different companies. To this end, more investigations should be performed with a large number of sample detectors from various vendors.

5 Conclusion

An aging simulation with an 80-kVp x-ray beam and exposure rate of 4.82 mGy s$^{-1}$ showed the decrease in gain or sensitivity of a CsI:Tl coupled CMOS detector. The rate of the gain decrease was high for the accumulated dose less than 0.2 kGy and then slowed down with further increases in the accumulated dose. The gain loss may cause a reduction of the dynamic range in clinical images. Although the detector showed a nonlinear response at low detector exposure level, the radiation damage did not affect the degree of nonlinearity. Linear analysis results of MTF, NPS, and DQE were independent of the accumulated dose. Radiation can change the physical performance of a detector, but proper gain-offset calibration and linear analysis of the images can yield consistent image signal-to-noise performance despite aging.

Acknowledgments

This work was supported by a 2-Year Research Grant of Pusan National University. The authors extend their gratitude to Rayence Co. for supplying the CMOS detector system.

References


