Scintillator high-gain avalanche rushing photoconductor active-matrix flat panel imager: Zero-spatial frequency x-ray imaging properties of the solid-state SHARP sensor structure

M. Wronski

Department of Radiation Oncology, Sunnybrook Health Sciences Center, Toronto, Ontario M4N 3M5, Canada

W. Zhao^{a)}

Department of Radiology, State University of New York at Stony Brook, Stony Brook, New York 11794-8460

K. Tanioka

Tokyo Denki University, Tokyo, 120-8551 Japan

G. DeCrescenzo and J. A. Rowlands

Thunder Bay Regional Research Institute, Thunder Bay, Ontario P7A 7T1, Canada

(Received 8 May 2012; revised 26 August 2012; accepted for publication 27 September 2012; published 5 November 2012)

Purpose: The authors are investigating the feasibility of a new type of solid-state x-ray imaging sensor with programmable avalanche gain: scintillator high-gain avalanche rushing photoconductor active matrix flat panel imager (SHARP-AMFPI). The purpose of the present work is to investigate the inherent x-ray detection properties of SHARP and demonstrate its wide dynamic range through programmable gain.

Methods: A distributed resistive layer (DRL) was developed to maintain stable avalanche gain operation in a solid-state HARP. The signal and noise properties of the HARP-DRL for optical photon detection were investigated as a function of avalanche gain both theoretically and experimentally, and the results were compared with HARP tube (with electron beam readout) used in previous investigations of zero spatial frequency performance of SHARP. For this new investigation, a solid-state SHARP x-ray image sensor was formed by direct optical coupling of the HARP-DRL with a structured cesium iodide (CsI) scintillator. The x-ray sensitivity of this sensor was measured as a function of avalanche gain and the results were compared with the sensitivity of HARP-DRL measured optically. The dynamic range of HARP-DRL with variable avalanche gain was investigated for the entire exposure range encountered in radiography/fluoroscopy (R/F) applications.

Results: The signal from HARP-DRL as a function of electric field showed stable avalanche gain, and the noise associated with the avalanche process agrees well with theory and previous measurements from a HARP tube. This result indicates that when coupled with CsI for x-ray detection, the additional noise associated with avalanche gain in HARP-DRL is negligible. The x-ray sensitivity measurements using the SHARP sensor produced identical avalanche gain dependence on electric field as the optical measurements with HARP-DRL. Adjusting the avalanche multiplication gain in HARP-DRL enabled a very wide dynamic range which encompassed all clinically relevant medical x-ray exposures.

Conclusions: This work demonstrates that the HARP-DRL sensor enables the practical implementation of a SHARP solid-state x-ray sensor capable of quantum noise limited operation throughout the entire range of clinically relevant x-ray exposures. This is an important step toward the realization of a SHARP-AMFPI x-ray flat-panel imager. © 2012 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4760989>]

Key words: amorphous selenium, avalanche multiplication gain, flat-panel imager

I. INTRODUCTION

In the last two decades, large area active matrix flat panel imagers (AMFPI) have been developed using active matrix arrays of amorphous silicon (α -Si) thin film transistors (TFT) and adopted widely in clinical x-ray imaging applications including digital radiography, fluoroscopy, and cone-beam computed tomography.¹ Their rapid advancement is fueled by the new clinical applications made possible by the rapid readout, high detective quantum efficiency (DQE), and compact size of AMFPI. However, there is one remaining challenge for

AMFPI: The DQE at very low exposures is degraded due to the readout electronic noise.^{2–5} This has been recognized as a major disadvantage compared to the more established imagers such as x-ray image intensifiers (XRII), which have internal signal gain.

Recently, many new AMFPI concepts have been proposed to improve the DQE of AMFPI at low exposures. The various strategies can be divided into two categories: one is to increase the x-ray-to-image charge conversion gain so that the x-ray quantum noise can overcome the electronic noise;^{6,7} and the other is to decrease the electronic noise by incorporating

amplification at each pixel of the TFT array,^{8,9} or using tiled CMOS amplified pixel sensors (APS).^{10,11} We are investigating the first strategy by the use of an AMFPI with avalanche gain, which we call scintillator high-gain avalanche rushing photoconductor AMFPI (SHARP-AMFPI).⁷ It is an indirect conversion AMFPI, where an x-ray scintillator such as structured cesium iodide (CsI) is used to convert x rays into optical photons. Instead of the *a*-Si photodiodes used in existing AMFPIs, a uniform layer of amorphous selenium (*a*-Se) photoconductor is incorporated to convert the optical photons into an electric signal. Additional signal gain is achieved by increasing the electric field across the *a*-Se allowing avalanche multiplication. As the degree of avalanche multiplication depends on the electric field strength E_{Se} , the signal gain is programmable. This programmable gain allows a wide dynamic range to be achieved while employing the same TFT array for signal readout as in existing AMFPIs.

In our previous Medical Physics Letter (MPL), the concept of a solid-state HARP photoreceptor structure was described, which included the addition of a distributed resistance layer (DRL), and its feasibility was demonstrated experimentally.¹² The HARP-DRL photoreceptor was shown to produce avalanche multiplication gains M of up to 10^4 without breakdown. The HARP-DRL solid-state structure will be incorporated into the development of a practical SHARP-AMFPI with a wide dynamic range by virtue of programmable gain in the HARP layer.

In the present work, the following three important x-ray detection characteristics of SHARP are investigated at zero spatial frequency: (1) the inherent signal and noise properties of the HARP-DRL avalanche photoreceptor for varying degrees of avalanche gain; (2) x-ray sensitivity and quantum noise limited performance at the lowest exposures encountered in clinical imaging, e.g., fluoroscopy; and (3) the dynamic range and linearity of HARP-DRL pertaining to the entire range of x-ray exposures encountered in clinical radiographic and fluoroscopic (R/F) applications.

II. MATERIALS AND METHODS

II.A. Optical imaging properties of the solid-state HARP structure

II.A.1. Construction of a solid-state HARP photoreceptor

The solid-state HARP-DRL sensor structure is shown schematically in Fig. 1(a).¹² Briefly, a 15 μm thick HARP multilayer structure with hole and electron blocking layers was deposited through thermal evaporation onto an ITO coated glass substrate. Then a 2 μm thick cellulose acetate (CA) layer was cast on top of the HARP. The CA layer, which constituted the DRL, was cured at room temperature for 24 h for the acetone solvent to fully evaporate. Then a semitransparent gold layer was deposited on top of the CA layer and constituted an electrode for a single detection element (del).

FIG. 1. (a) Schematic of the HARP-DRL solid-state sensor structure, which converts optical photons into electron-hole pairs in an *a*-Se layer. The high electric field strength in this layer causes avalanche multiplication of holes, producing signal amplification. The distributed resistance layer (DRL) prevents breakdown at the high field strengths required for avalanche multiplication. (b) The SHARP x-ray sensor is obtained by coupling a structured CsI scintillator to the HARP-DRL photoreceptor. This constitutes a solid-state indirect conversion x-ray detector with avalanche multiplication gain.

II.A.2. Measurement of signal and noise properties of HARP-DRL

The avalanche gain of HARP-DRL was first characterized by measuring the signal and dark current as a function of electric field strength using the method described previously,¹² and the apparatus shown in Fig. 2(a). The ITO electrode was biased positively using a high voltage (HV) power supply, and for dark current measurement the gold electrode was connected to the virtual ground input of an electrometer. Each measurement was taken 2 min after a change in HV bias to allow the dark current to stabilize. For optical measurement, the gold electrode was connected to an oscilloscope with an input load resistor of 10^6 ohm. A light emitting diode (LED) with peak wavelength of 465 nm was used as the optical excitation source and the width of each optical pulse was 2 ms. The peak potential of the signal associated with each optical excitation pulse was measured as a function of electric field E_{Se} , and the result was used to derive the relative signal gain. Since both the optical quantum efficiency, γ , and the avalanche gain M of *a*-Se increase as a function of electric field E_{Se} ,¹³ two empirical expressions were used to fit the HARP-DRL

FIG. 2. (a) Experimental apparatus used for measuring signal gain and dark current. An oscilloscope was used for measuring the peak output voltage produced by the HARP-DRL photoreceptor under pulsed optical excitation. An electrometer was used for measuring dark current. Only one instrument was used at a time, as indicated by the dotted lines. (b) Experimental apparatus used for measuring the noise properties of HARP-DRL by means of pulse height spectroscopy. (c) Experimental apparatus for measuring SHARP signal at very low x-ray exposures.

gain characteristics. For $E_{Se} < 75 \text{ V}/\mu\text{m}$, i.e., no avalanche, a three parameter model was used to determine $\gamma(E_{Se})$:

$$\gamma = \frac{E_{Se}}{aE_{Se} + b}, \quad (1)$$

where $1/a$ is the value of γ at infinite field and b is the electron-hole pair recombination term which reduces γ at low field. For $E_{Se} > 75 \text{ V}/\mu\text{m}$, where avalanche multiplication occurs, the McIntyre theory¹⁴ was used to model the avalanche gain characteristics. We assumed that only holes avalanche in a -Se, which is a reasonable assumption for a -Se biased below $100 \text{ V}/\mu\text{m}$, and the avalanche gain is given by

$$M = \exp[d_{Se}\beta_1 \exp(-\beta_2/E_{Se})], \quad (2)$$

where d_{Se} is the a -Se thickness and β_1 and β_2 are the impact ionization factors of a -Se.

The signal-to-noise ratio (SNR) of the solid-state HARP-DRL photoreceptor for the detection of optical photons was measured using the pulse-height spectroscopy (PHS) method, which is shown in Fig. 2(b). The ITO electrode was biased positively using a HV power supply, while the gold electrode was connected to the virtual ground of a low noise charge sen-

sitive amplifier (AMPTEK CoolFET A250CF) with a shaping time of $8 \mu\text{s}$. The electric field across the HARP-DRL, and hence avalanche gain M , was controlled by the HV bias. The width of the optical excitation pulse used for this experiment was $\sim 2 \mu\text{s}$. The shaped charge signal from HARP-DRL was connected to a multichannel analyzer (MCA). The extracted pulse height measurements for each optical excitation formed the PHS, which is a plot of the number of events as a function of the pulse height. The mean and standard deviation of the PHS were used to compute the signal and noise as a function of E_{Se} for a given optical exposure level. The instrumentation noise of the PHS apparatus was first measured with the HARP-DRL connected to the input of the charge amplifier, but without the HV bias or the optical excitation. The PHS method has been used previously by our group for medical imaging detector characterization to quantify the noise associated with amplification (gain) processes.^{15,16} Our objective is to establish the noise associated with the avalanche multiplication process in solid-state HARP-DRL.

II.A.3. Cascaded linear system model for signal and noise propagation in HARP-DRL

The potential x-ray imaging performance of SHARP-AMFPI has been investigated theoretically using a cascaded linear system model.^{7,17} Since the present work is focused on the effect of HARP-DRL on the zero-frequency performance, the signal and noise propagation pertaining to DQE(0) was used to investigate the effects of avalanche gain and its associated noise of a HARP-DRL. Shown in Fig. 3 is the zero-frequency propagation of signal and noise through a SHARP-AMFPI,¹⁷ and the stages associated with HARP are highlighted.

Previous study showed that following the cascaded linear system analysis, the DQE(0) of SHARP-AMFPI can be calculated using¹⁷

$$\text{DQE}(0) = \frac{\eta}{\left(1 + \frac{\sigma_g^2}{g^2} - \frac{1}{g}\right) + \frac{1}{\delta\gamma g} \left(1 + \frac{\sigma_M^2}{M^2}\right) + \frac{\sigma_a^2}{N\eta g^2 \delta^2 \gamma^2 M^2}}, \quad (3)$$

where as noted in Fig. 3, η and g represent the quantum absorption efficiency and the x-ray-to-photon conversion gain of the scintillator, respectively. The denominator of Eq. (3) highlights the three degradation factors of DQE(0), where the first term is due to the variation in gain in the scintillator, the second term is due to the optical quantum efficiency and avalanche gain in HARP, and the third term is due to the read-out electronic noise. While the main advantage of avalanche gain is to reduce the contribution of electronic noise by $1/M^2$, we have to ensure that the additional noise due to the stochastic nature of avalanche gain, σ_M^2 , is negligible. Since only holes avalanche in a -Se, it has been shown previously that the gain variance is given by^{14,18}

$$\sigma_M^2 = M^2 - M. \quad (4)$$

FIG. 3. Flow chart showing the signal and noise propagation through a zero spatial frequency cascaded linear system model for SHARP-AMFPI, with the stages associated with HARP-DRL shaded in grey. The gain and variance in gain for each stage is outlined.

Substituting Eq. (4) into Eq. (3), it can be shown that the noise due to secondary quanta, i.e., the second term in the denominator of Eq. (3), will be negligible when $\delta\gamma g \gg (2 - 1/M)$. This condition has been validated through experimental measurements to be true for the SHARP combination.¹⁷ The main purpose of the present investigation is to establish that the newly developed solid-state HARP sensor, HARP-DRL, follows the same avalanche gain noise characteristics as the HARP tube (with electron beam readout) used in our previous investigation. This investigation can be accomplished using SNR measurements with the apparatus shown in Fig. 2(b), and comparing the measurement results with theoretical prediction of the cascaded linear system model. The signal and noise propagation through the optical quantum efficiency (selection) and stochastic avalanche gain processes associated with HARP are highlighted in Fig. 3. The optical photons transmitted through the bias electrode of HARP, q_h , are converted to charge signal by *a*-Se with the optical quantum efficiency γ and then amplified by the avalanche multiplication gain M . The Poisson noise associated with q_h , $\sigma_{q_h}^2 = q_h$, is modified by the selection stage associated with γ , and then the avalanche amplification stage with gain variance of $\sigma_M^2 = M^2 - M$. Finally, the readout electronic noise σ_a^2 is added. The total noise variance resulting from q_h optical photons entering HARP-DRL is given by

$$\sigma_N^2 = (2M^2 - M)\gamma q_h + \sigma_a^2. \quad (5)$$

For large M , the gain variance $M^2 - M$ can be approximated as M^2 . The total noise variance then simplifies to

$$\sigma_N^2 = 2M^2\gamma q_h + \sigma_a^2. \quad (6)$$

By comparing the measured noise and SNR with that predicted by the model using Eq. (6), it can be determined whether there are any additional noise sources in HARP-DRL that are unaccounted for.

II.B. X-ray imaging properties of SHARP

II.B.1. Construction of a solid-state SHARP sensor structure

A SHARP x-ray sensor was made by coupling a 300 μm thick structured CsI:Tl scintillator (FOSHL, Hammamatsu Photonics) to the transparent glass substrate of the HARP-DRL sample. Optical grease was used to maximize the opti-

cal coupling efficiency from the CsI into the HARP-DRL. A diagram of the SHARP structure is shown in Fig. 1(b). It was adequate for zero-spatial frequency measurements.

II.B.2. X-ray detection with SHARP at very low exposure

The current from SHARP resulting from x-ray exposures was measured with the experimental apparatus shown in Fig. 2(c). The x-ray beam quality was 75 kVp from a tube with tungsten (W) target and 2 mm of aluminum filtration. A 30 cm thick block of lucite was placed at the x-ray tube output to mimic the attenuation produced by a patient. The distance between the x-ray source and the SHARP sensor was 1.5 m. A 15 ml ionization chamber (Fluke Biomedical 96035B) and a digital dosimeter (Fluke Biomedical model 35040) were used to measure the entrance exposure to SHARP. The signal current from the SHARP sensor under continuous x-ray exposure was measured directly using an oscilloscope (with 10^6 ohm input impedance to provide a voltage reading proportional to current), and the result was plotted as a function of the bias potential applied across the HARP-DRL photoreceptor.

II.B.3. Dynamic range and linearity

The experimental setup shown in Fig. 2(a) was used to determine the linear range of operation of HARP-DRL and infer the result to the SHARP sensor through the x-ray sensitivity measured in Sec. II.B.2. The HARP-DRL consisted of a 15 μm *a*-Se layer and a 2 μm CA layer. A 465 nm peak wavelength LED was used to excite the HARP-DRL with 2 ms optical pulses of variable photon fluence. The LED fluence was monitored using a photomultiplier tube (PMT), as shown in Fig. 2(a). One end of an optical fiber bundle (Newport model 77525) was placed near the HARP-DRL in a light-proof box and the other end was connected to the PMT. An optical attenuator was positioned between the fiber bundle and the PMT such that the PMT would not saturate at the higher photon fluences. The linearity of the PMT was confirmed over a very wide range of optical exposures. Using the setup shown in Fig. 2(c), the current produced by SHARP at $E_{\text{Se}} = 10 \text{ V}/\mu\text{m}$ was measured using an electrometer with an x-ray tube current of 20 mA, which corresponds to an x-ray exposure of 20 μR per x-ray pulse. Next, the same electrometer current

was reproduced using the setup in Fig. 2(a) by adjusting the LED optical fluence with the HARP-DRL also biased at $E_{Se} = 10 \text{ V}/\mu\text{m}$. This enabled the PMT output signal to be calibrated to the corresponding x-ray exposure of $20 \mu\text{R}$ per pulse. At this point, the PMT signal could be used to infer the SHARP *equivalent x-ray exposure* corresponding to any given LED fluence. Using the setup in Fig. 2(a) and starting with an equivalent x-ray exposure of 20 mR per pulse, the output signal from HARP-DRL at $E_{Se} = 10 \text{ V}/\mu\text{m}$ was plotted for a progressively decreasing LED fluence. Once the signal could no longer be measured, the HARP-DRL bias was increased to $73 \text{ V}/\mu\text{m}$, producing once again a measurable signal and allowing the LED fluence to be further reduced. This was repeated again for bias settings of 87 and $93 \text{ V}/\mu\text{m}$ enabling the detection of an equivalent x-ray exposure as low as $0.05 \mu\text{R}$ per pulse. This approach was used to investigate the wide dynamic range of SHARP through programmable gain.

III. RESULTS

III.A. Signal and noise properties of HARP-DRL

A square wave of known amplitude was applied to the test input of the preamplifier and from the width and mean value of the measured PHS we obtained the system electronic noise. During this measurement, the detector was connected to the preamplifier's input, since its capacitance affects the noise; however, no HV bias or external optical excitation was applied to the detector. The electronic noise inherent to the PHS setup was thus determined to be 230 electrons. Before acquiring measurement data, the source intensity for the pulsed optical excitation has been adjusted such that the signal and noise are represented by approximately the same number of electrons in the absence of avalanche gain ($M = 1$). Shown in Fig. 4(a) is the output signal of the HARP-DRL as well as the standard deviation (rms noise) plotted as a function of the avalanche multiplication gain M . The avalanche gain on the x axis of Fig. 4 was inferred directly from the measured sig-

nal (i.e., the relationship between measured signal and gain is linear by definition). Figure 4(b) shows the SNR for the same theoretical and experimental data. Although avalanche gain of $\sim 10^4$ has been demonstrated for HARP-DRL in our previous work,¹² the value of M was kept < 100 in the present investigation because it is sufficient to maintain x-ray quantum noise limited operation for SHARP-AMFPI.⁷

III.B. X-ray detection with SHARP at very low exposure

Shown in Fig. 5 is the measured signal and dark current obtained from the SHARP structure using the apparatus shown in Fig. 2(c). The x-ray signal data were acquired for two different settings. For HV biases below 1100 V , i.e., no avalanche, an x-ray tube current of 20 mA was used, corresponding to a measured exposure at the HARP-DRL of $20 \mu\text{R}$ per x-ray pulse. In the avalanche regime ($\text{HV} > 1100 \text{ V}$), the x-ray tube current was reduced to 1 mA (corresponding to a $1 \mu\text{R}$ x-ray exposure to SHARP). The photocurrent produced by HARP-DRL using optical excitation (reported in our previous work)¹² is also shown in Fig. 5 as a function of HV bias. The dark current was negligible compared to the signal current for all the data reported.

The empirical fitting parameters a and b used in Eq. (1) for optical quantum efficiency γ were 6.6 and 440 , respectively,¹⁹ and the impact ionization coefficients β_1 and β_2 were 1000 and 800 which are in close agreement to previously published values.^{19,20} The measured field dependent gain agrees very well with both the optical quantum efficiency (without avalanche for $\text{HV} < 1100 \text{ V}$) and avalanche ($\text{HV} > 1100 \text{ V}$) gain models. Furthermore, the HARP-DRL optical measurements agree closely with the SHARP x-ray measurements.

III.C. Dynamic range and linearity

Shown in Fig. 6 is the measured charge produced by HARP-DRL and plotted as a function of the equivalent

FIG. 4. (a) Graph showing the measured signal and noise produced by HARP-DRL for avalanche multiplication gains in the range 1-100 obtained using the apparatus shown in Fig. 2(b). Measured data are shown as squares and circles. Solid lines represent theoretical models shown in Fig. 3, with noise derived using Eq. (6). (b) Graph showing the corresponding signal-to-noise ratio (SNR) as a function of avalanche multiplication gain.

FIG. 5. Graph showing the dark current (triangles) and SHARP photocurrent (squares) measured using the apparatus depicted in Fig. 2(c) (x-ray source). Also shown is the HARP-DRL photocurrent (circles) measured using the apparatus shown in Fig. 2(a) (optical source). The solid and dashed lines represent the field dependent optical quantum efficiency and avalanche gain models, respectively, given by Eqs. (1) and (2).

FIG. 6. HARP-DRL output charge density (in electrons/mm²) as a function of the equivalent SHARP x-ray exposure per frame (R/frame) at different electric field strength. The corresponding total gain g (relative to HARP biased at 10 V/μm) is shown in the inset. Data for a field strength of 10 V/μm was acquired both at 30 Hz (circles) and at 1 Hz (rightward triangles). Data for all other field strengths were acquired at 30 Hz excitation pulse repetition. A reduction in gain with increased exposure is observed for 30 Hz readout rate, but not for 1 Hz readout rate (radiography). The mechanism is discussed in the text.

SHARP x-ray exposure, which covers clinically relevant fluoroscopy and radiography modes of operation. For comparison, the electronic noise associated with AMFPI is also shown. The measured charge is plotted for excitation pulses with a repetition rate of 30 Hz and pulse duration of 2 ms, which closely mimic the timing of the x-ray pulse sequence delivered by most modern pulsed fluoroscopy x-ray generators. The charge produced by HARP-DRL is shown for several gain settings. We use the sensitivity at $E_{Se} = 10 \text{ V}/\mu\text{m}$ as a reference gain of unity, and the relative gain varies up to 80 when E_{Se} is increased to $93 \text{ V}/\mu\text{m}$. For reference, one set of data was obtained at $E_{Se} = 10 \text{ V}/\mu\text{m}$ (gain of 1) with a relatively slow pulse repetition rate (1 pulse/second), mimicking DC.

IV. DISCUSSION

IV.A. Signal and noise properties of HARP-DRL and its implications for low dose x-ray imaging

It can be seen in Fig. 4 that the measured noise as a function of avalanche gain is in close agreement with the cascaded noise model given in Eq. (6). This indicates that the avalanche noise behavior of HARP-DRL has the same characteristics as other HARP structures which were used in previous studies and hence expected to have the ability to maintain $DQE(0)$ at the lowest exposures, i.e., $0.1 \mu\text{R}/\text{frame}$, encountered in medical imaging applications. Our experimental results in Fig. 5 demonstrate that SHARP can overcome the effects of electronic noise throughout the entire clinical fluoroscopic x-ray exposure range. However, it can also be seen in Fig. 4 that the measured noise starts to deviate from the theoretical value as avalanche gain approaches 100. This coincides with a noticeable increase in dark current across the HARP-DRL sample during the measurement. The increase in dark current was ascribed to the imperfection in sample preparation during this early stage of development, which led to some deterioration of performance with time under high voltage bias and repeated exposures. It is important to note that this slight deviation will not result in any degradation in x-ray imaging performance of SHARP because the condition of $\delta\gamma g \gg (2 - 1/M)$ is still satisfied.

IV.B. Dynamic range of SHARP for x-ray imaging

The wide dynamic range of SHARP for clinical x-ray imaging applications can be established through programmable gain by changing the operating electric field strength E_{Se} . This can be seen from Fig. 6 which shows the measured HARP-DRL output charge as a function of equivalent x-ray exposure encountered in R/F applications. For $E_{Se} = 10 \text{ V}/\mu\text{m}$ and 1 frame/second readout, which corresponds to radiography, the HARP-DRL response resembles that of existing AMFPI, which is linear as a function of exposure. However, at this E_{Se} the detector, like existing AMFPI, is not x-ray quantum noise limited when the exposure approaches the mean value used in fluoroscopy. At $E_{Se} = 73 \text{ V}/\mu\text{m}$, where the onset of avalanche multiplication occurs, the linear range

of operation at 30 frames/s encompassed most of the radiographic and pulsed high dose fluoroscopy regions. The moderate gain of 6, however, is not sufficient to overcome electronic noise below the mean detector exposure in fluoroscopy, which is $1 \mu\text{R}/\text{frame}$.

By further increasing E_{Se} to $87 \text{ V}/\mu\text{m}$ and $93 \text{ V}/\mu\text{m}$, corresponding to relative gains of 20 and 80, respectively, the signal charge produced by HARP-DRL at $0.1 \mu\text{R}/\text{frame}$ becomes sufficient to overcome electronic noise. Thus, by operating the detector at different avalanche multiplication gain settings, a wide piecewise dynamic range can be achieved for HARP-DRL without increasing the maximum charge received by the readout electronics, i.e., the TFT array in SHARP-AMFPI. A solid-state SHARP-AMFPI using the HARP-DRL photoreceptor could, for example, have three programmable gain settings. A gain of 50 could be used for low-dose fluoroscopy, a gain of 6 could be used for high-dose fluoroscopy, and a gain of 1 could be used for radiography.

One interesting phenomenon seen in Fig. 6 is that at 30 frames/s readout rate, the measured signal exhibits a non-linearity due to gradual reduction in sensitivity as exposure increases. This is ascribed to the buildup of space charge inside the DRL, which takes some time (e.g., 30 ms) to dissipate. The instantaneous charge buildup at high exposure rate decreases the electric field across HARP, and in turn, decreases the avalanche multiplication gain. This effect is more pronounced at larger field strengths since the avalanche gain characteristic is steeper at these fields [Eq. (2)], which means larger change in avalanche gain for a given change in E_{Se} . It is important to note that this reduction in gain does not lead to signal saturation because the HARP-DRL is still sensitive to radiation.

This reduction in gain is an important, self-limiting mechanism that restricts the total amount of charge produced in the avalanche photoconductor under very high exposure. It is, in fact, advantageous for a flat panel detector used in interventional radiology applications since part of the detector may be exposed to the direct x-ray beam. For a 30 cm patient, the direct x-ray beam will have 1000 times higher exposure to the detector than the mean exposure behind the patient. It can be seen from Fig. 6, that if a direct exposure occurs in the fluoroscopic mode, the rapid increase in photon-generated charge would gradually reduce the avalanche gain, and eventually no avalanche when $E_{\text{Se}} < 70 \text{ V}/\mu\text{m}$ (Fig. 5). This mechanism alleviates the problem of excessive image charge associated with a constantly high gain.

It should be noted that the linearity and signal saturation characteristics of HARP-DRL are largely dependent on the choice of DRL material and thickness. Although the DRL used in this work ($2 \mu\text{m}$ thick CA) is more than adequate for proper operation of HARP-DRL throughout the entire clinical range of x-ray exposures, further work is necessary to determine the optimal DRL characteristics not only from the perspective of charge transport but also from a practical standpoint pertaining to HARP-DRL integration with AMFPI technology. Challenges such as the manufacturability of large-area HARP-DRLs with uniform characteristics will need to

be overcome before this technology can supersede existing flat panel systems.

V. CONCLUSIONS

We have shown that the avalanche gain and noise characteristics of our newly developed solid-state HARP-DRL sensor structure follow the same behavior as the HARP tube (with a free surface and vacuum electron beam readout) used in previous investigations, which means that the avalanche gain noise of HARP-DRL does not cause any additional noise in SHARP-AMFPI. Experimental measurements have demonstrated that the charge produced by the solid-state SHARP x-ray sensor with avalanche gain is sufficient to overcome electronic noise at the lowest clinical exposures encountered in medical imaging applications. Furthermore, adjusting the avalanche multiplication gain in HARP-DRL enables a wide dynamic range which encompasses all clinically relevant medical x-ray exposures. The decrease in gain at the high exposure region designed for each operating point of HARP-DRL provides a self-limiting mechanism which prevents excessive generation of charge, thus protecting the underlying AMFPI electronics. The solid-state HARP-DRL sensor technology hence provides an important step toward the realization of a practical SHARP-AMFPI for a wide range of clinical x-ray imaging applications.

ACKNOWLEDGMENTS

The authors gratefully acknowledge financial support from the National Institutes of Health (R01 EB002655 and R01 CA148053). Research on DRL was carried out in part at the Center for Functional Nanomaterials, Brookhaven National Laboratory, which is supported by the U.S. Department of Energy, Office of Basic Energy Sciences, under Contract No. DE-AC02-98CH10886.

^{a)} Author to whom correspondence should be addressed. Electronic mail: Wei.zhao@sunysb.edu

¹ J. A. Rowlands and J. Yorkston, "Flat panel detectors for digital radiography," in *Medical Imaging: Volume 1 Physics and Psychophysics, Psychophysics*, edited by H. K. a. R. V. M. J. Beutel (SPIE, Bellingham, 2000), pp. 223–328.

² L. E. Antonuk, K. W. Jee, Y. El-Mohri, M. Maolinbay, S. Nassif, X. Rong, Q. Zhao, J. H. Siewerdsen, R. A. Street, and K. S. Shah, "Strategies to improve the signal and noise performance of active matrix, flat-panel imagers for diagnostic x-ray applications," *Med. Phys.* **27**, 289–306 (2000).

³ L. E. Antonuk, Q. Zhao, Y. El-Mohri, H. Du, Y. Wang, R. A. Street, J. Ho, R. Weisfield, and W. Yao, "An investigation of signal performance enhancements achieved through innovative pixel design across several generations of indirect detection, active matrix, flat-panel arrays," *Med. Phys.* **36**, 3322–3339 (2009).

⁴ J. Mort and P. Nielsen, "Transient accumulation of interfacial charge in photoconductor-dielectric systems," *Phys. Rev. B* **5**, 3336–3342 (1972).

⁵ A. Koch, J. M. Macherel, T. Wirth, P. De Groot, T. Ducourant, D. Couder, J. P. Moy, and E. Calais, "Detective quantum efficiency of an x-ray image intensifier chain as a benchmark for amorphous silicon flat panel detectors," *Proc. SPIE* **4320**, 115–120 (2001).

⁶ R. A. Street, S. E. Ready, K. Van Schuylenbergh, J. Ho, J. B. Boyce, P. Nylen, K. Shah, L. Melekhov, and H. Hermon, "Comparison of Pbl2 and Hgl2 for direct detection active matrix x-ray image sensors," *J. Appl. Phys.* **91**, 3345–3356 (2002).

- ⁷W. Zhao, D. Li, A. Reznik, B. J. Lui, D. C. Hunt, J. A. Rowlands, Y. Ohkawa, and K. Tanioka, "Indirect flat-panel detector with avalanche gain: Fundamental feasibility investigation for SHARP-AMFPI (scintillator HARP active matrix flat panel imager)," *Med. Phys.* **32**, 2954–2966 (2005).
- ⁸Y. El-Mohri, L. E. Antonuk, M. Konieczek, Q. Zhao, Y. Li, R. A. Street, and J.-P. Lu, "Active pixel imagers incorporating pixel-level amplifiers based on polycrystalline-silicon thin-film transistors," *Med. Phys.* **36**, 3340–3355 (2009).
- ⁹K. S. Karim and A. Nathan, "Active pixel sensor architectures in a-SiH for medical imaging," *J. Vac. Sci. Technol. A* **20**, 1095–1099 (2002).
- ¹⁰Y. S. Kim, K. H. Kim, and G. Cho, "Optimization of CMOS active pixels with high signal-to-noise ratio for high-resolution x-ray imaging," *Nucl. Instrum. Methods Phys. Res. A* **591**, 248–251 (2008).
- ¹¹S. Naday, E. F. Bullard, S. Gunn, J. E. Brodrick, E. O. O'Tuairisg, A. McArthur, H. Amin, M. B. Williams, P. G. Judy, and A. Konstantinidis, "Optimised breast tomosynthesis with a novel CMOS flat panel detector," in *10th International Workshop on Digital Mammography, IWDM 2010, June 16, 2010 – June 18, 2010* (Springer Verlag, Girona, Catalonia, Spain, 2010), Vol. 6136 LNCS, pp. 428–435.
- ¹²M. M. Wronski, W. Zhao, A. Reznik, K. Tanioka, G. DeCrescenzo, and J. A. Rowlands, "A solid-state amorphous selenium avalanche technology for low photon flux imaging applications," *Med. Phys.* **37**, 4982–4985 (2010).
- ¹³A. Reznik, B. J. M. Lui, Y. Ohkawa, T. Matsubara, K. Miyakawa, M. Kubota, K. Tanioka, T. Kawai, W. Zhao, and J. A. Rowlands, "The quantum efficiency of photo-charge generation in a-Se avalanche photodetectors," *Nucl. Instrum. Methods Phys. Res. A* **567**, 93–95 (2006).
- ¹⁴R. J. McIntyre, "Multiplication noise in uniform avalanche diodes," *IEEE Trans. Electron Devices* **13**, 164–168 (1966).
- ¹⁵J. A. Rowlands, G. DeCrescenzo, and N. Araj, "X-ray imaging using amorphous selenium: Determination of x-ray sensitivity by pulse height spectroscopy," *Med. Phys.* **19**, 1065–1069 (1992).
- ¹⁶I. M. Blevis, D. C. Hunt, and J. A. Rowlands, "X-ray imaging using amorphous selenium: Determination of Swank factor by pulse height spectroscopy," *Med. Phys.* **25**, 638–641 (1998).
- ¹⁷D. C. Hunt, K. Tanioka, and J. A. Rowlands, "X-ray imaging using avalanche multiplication in amorphous selenium: Investigation of intrinsic avalanche noise," *Med. Phys.* **34**, 4654–4663 (2007).
- ¹⁸T. Ohshima, K. Tsuji, K. Sameshima, T. Hirai, K. Shidara, and K. Take-toshi, "Excess noise in amorphous Selenium avalanche photodiodes," *Jpn. J. Appl. Phys., Part 1* **30**, 1071–1074 (1991).
- ¹⁹D. C. Hunt, S. S. Kirby, and J. A. Rowlands, "X-ray imaging with amorphous selenium: X-ray to charge conversion gain and avalanche multiplication gain," *Med. Phys.* **29**, 2464–2471 (2002).
- ²⁰K. Tsuji, Y. Takasaki, T. Hirai, J. Yamzaki, and K. Tanioka, "Avalanche phenomenon in amorphous selenium," *Optoelectron., Devices Technol.* **9**, 367–378 (1994).