Digital Radiology Using Active Matrix Readout of Amorphous Selenium

by

Wei Zhao

A thesis submitted in conformity with the requirements for the degree of Doctor of Philosophy
Graduate department of Medical Biophysics
University of Toronto

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To Mom, Dad and Hong
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Abstract

The concept of a direct, real-time, flat-panel detector is being investigated for digital radiology. The detector employs a layer of x-ray sensitive photoconductor (e.g. amorphous selenium) to directly convert x-rays to charge, and a large area integrated circuit called an active matrix which consists of a two dimensional array of thin film transistors (TFTs) for electronic image readout. The readout can be performed in real-time (30 frames per second) and thus can be applied in both radiography and fluoroscopy. The advantages of the direct, flat-panel detector include: instantaneous image readout, compact size in the form of a cassette, efficient use of x-rays to produce high quality x-ray images, and freedom from geometric image distortion. This work investigates practical solutions for making a high quality and reliable flat panel detector. The two factors affecting the reliability of the detector are: radiation hardness of TFTs and operation of the detector under the high voltage applied to the x-ray photoconductor. The image quality of the detector is measured in terms of detective quantum efficiency (DQE) at all spatial frequencies.

The radiation hardness of TFTs is measured at the life-time dose expected for a diagnostic x-ray imaging detector. The dominant effect of radiation is the negative shift of the TFT threshold voltage by less than 1 V, which is insufficient to affect the normal operation of the TFTs. Thus
the radiation hardness of TFTs is adequate for diagnostic x-ray imaging.

In order to assure a safe and reliable operation of TFTs under the high voltage bias to the x-ray photoconductor, a novel dual-gate TFT structure with self-protection against high voltage damage is proposed and investigated. The results show that an existing dual-gate TFT design can provide satisfactory high voltage protection for all radiographic applications.

The DQE of the detector as a function of spatial frequency was studied both theoretically using a cascaded linear systems model, and experimentally with a prototype detector with 160 $\mu$m pixel pitch. The results obtained using the theoretical model agree with those obtained from experimental measurements. Detector design parameters affecting the DQE are identified and strategies for maximizing DQE are proposed.
Acknowledgements

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- \( DQE(f) \): spatial frequency dependent DQE .................................................. 6
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- \( E_C \): conduction band edge .................................................................................. 30
- \( E_V \): valence band edge ....................................................................................... 30
- \( V_G \): gate voltage of a TFT .................................................................................. 30
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Chapter 1

Introduction

1.1 Current Radiological Imaging Methods

Projection radiological imaging encompasses a large variety of diagnostic tasks and imaging equipment and involves two major modalities: radiography and fluoroscopy. Radiography is the production of a frozen x-ray image using a single, brief x-ray exposure, and fluoroscopy is an imaging technique that produces a sequence of x-ray images in real-time.

Radiological imaging has played a very important role in diagnostics from the beginning of this century. Since the 1970’s, new digital imaging modalities have been developed which include computed tomography (CT), ultrasound (US), nuclear medicine and magnetic resonance imaging (MRI). However, basic x-ray imaging is still used and accounts for more than 50 % of diagnostic imaging examinations. [1] This is because: (1) The cost of diagnostic x-rays is much less than that for CT and MRI; (2) X-ray is the only imaging modality that is able to produce very high resolution images, e.g. higher than 10 line pairs (lp)/mm, making it an outstanding candidate for special imaging tasks such as mammography - i.e. imaging of the breast; [2] (3) X-ray is one of the major imaging modalities that can produce images in real-time (30 frames per second) and thus
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plays an important role in dynamic imaging studies of the heart and interventional procedures; (4)
An x-ray radiograph can be obtained using a short pulse of radiation which eliminates the problems
of image blurring due to patient motion. Now I will further examine the current imaging methods
used in both radiography and fluoroscopy.

1.1.1 Screen-film radiography

The first radiograph was produced by Roentgen in 1895 using a photographic plate. Shortly there-
after Pupin proposed using a phosphor screen coupled to a photographic film which greatly reduced
the exposure needed to produce a radiograph. [3] Screen-film technology has been continuously im-
proved: screen-films have been made such that the optical spectrum of the screens matches the
sensitivity of the films; many different screen-film combinations are available to suit different re-
quirements for x-ray sensitivity and image resolution. Today, most radiography is still performed
with screen-film systems.

Despite all these improvements, screen-film technology still has three important limitations:

(1) Since photographic film is used for image acquisition, display and storage, it is impossible
to optimize its performance for all these functions. Figure 1.1 shows the characteristic curve (the
optical density as a function of incident exposure) of two types of films. The display contrast of
a film is determined by the slope $\gamma$ or gradient of its characteristic curve. The exposure range
over which the film display gradient is significant (i.e. latitude) is very narrow. Any improvement
in contrast requires a decrease in latitude, as shown by the characteristics of film type A (higher
contrast) and B (larger latitude). In the regions where the exposure is low (toe) or high (shoulder)
there is almost no contrast. Thus in chest radiography, the images of the mediastinum (with high
x-ray attenuation) and lung (with low x-ray attenuation) are in the toe and shoulder regions of the
film, respectively, and result in very poor image contrast. In mammography, for women with dense
breasts, also referred to as DY-type breasts [4], the image of a tumour (with slightly higher x-ray attenuation than surrounding tissue) may not have enough contrast to be visible since both the tumour and its surrounding tissue are imaged in the toe region of the film. Thus the optical response of film is not suited for detecting wide variations of radiation exposure such as that required in chest radiography and mammography while giving sufficient contrast.

![Diagram of photographic characteristics of two hypothetical films](image)

**Figure 1.1:** The photographic characteristics of two hypothetical films: Film A has higher contrast and smaller latitude; Film B has lower contrast and larger latitude.

(2) Film granularity is an important noise source for screen-film radiography. An ideal x-ray imaging system should be *quantum noise limited* which means that x-ray quantum noise is the dominant source of the random fluctuation in the images. But for screen-films, above a spatial
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frequency of 5 lp/mm film granularity noise becomes dominant [5]. Thus screen-film is not an ideal system for imaging fine detail structures such as is required in mammography.

(3) The use of phosphor screens for x-ray detection degrades the image information at high spatial frequency. In a phosphor screen, each absorbed x-ray generates multiple light photons. These light photons perform a random walk before they can escape from the phosphor and expose the photographic film, as shown in Figure 1.2 (a). Consequently, the image appears blurred. The thicker the phosphor screen is made in order to absorb more x-rays, the more blurred the images are. This blurring causes a decrease in signal to noise ratio for high frequency image information which is fundamental and irreversible. One possible solution to this problem is to use a structured phosphor [cesium iodide (CsI)]. CsI can be grown in a structure reminiscent of a fiber optic, as shown in Figure 1.2 (b), which helps to guide the fluorescent light from deep within the phosphor to the surface where the image is formed. Although the image formed by structured CsI is less blurry than that formed by phosphor screens, since the separation between fibers is created by cracking, the channeling of light is not perfect. Although structured CsI has been used in some x-ray imaging systems (e.g. the x-ray image intensifier which will be discussed in the next section) to improve image resolution, due to its hydroscopic nature which requires the screen to be sealed in vacuum, it has not been applied to screen-film radiography.

As a result of these three limitations, the overall imaging performance of a screen-film, which can be described in terms of its detective quantum efficiency (DQE), is far from optimum for high spatial frequency information. DQE can be defined as the square of the signal to noise ratio (SNR) at the output of the detector to that at the input, and is usually given as a function of spatial frequency $f$: [6]

$$DQE(f) = \frac{SNR_{\text{out}}^2(f)}{SNR_{\text{in}}^2(f)}.$$ \hspace{1cm} (1.1)
Figure 1.2: Physical processes involved in production of an x-ray image: (a) random walk of light photons in phosphor; (b) light channeled by structured CsI screens; (c) charge drawn to the surfaces of the photoconductor in electrostatic.
SNR_{in}(f) is equal to the square root of the number of incident x-ray photons. SNR_{out}(f) is determined by both the x-ray absorption and added noise of the imaging detector. Ideally, DQE(f) should have a value of 100% over the spatial frequencies of interest. However for screen-films (e.g. Lanex regular/ortho G), as shown in Figure 1.3, DQE is 30% at zero spatial frequency, and as spatial frequency increases, DQE rapidly decreases to ~1% [7]. This means that screen-films are efficient x-ray detectors at low spatial frequency, but become non-ideal for high resolution imaging applications such as mammography.

Figure 1.3: The spatial frequency dependent detective quantum efficiency of a screen film (Lanex regular/ortho G) at 70 kVp (shown in solid line) (ref. by Bunch et al.); photostimulable phosphor systems at 70 kVp (dotted line) (ref. by Hillen et al. (87)); and an α-Se detector with electrometer readout method at 50 keV (dashed line) (ref. by Hillen et al. (88)).
1.1.2 X-ray image intensifier based fluoroscopy/fluorography

Earlier this century fluoroscopy was performed by directly viewing light emitted by a fluorescent screen which absorbed the x-rays transmitted through the patient [8]. Because the screen brightness was limited, the radiologist had to visualize the images following a lengthy period of dark adaptation. Detail perception was limited by the reduced visual acuity of the eye under dark adaptation, the poor light production efficiency of the fluorescent screen, and the inefficient collection by the eye of the fluorescent light quanta [9]. In the early 1950’s the x-ray image intensifier (XRII) was invented which overcomes the problems of fluorescent screens, and has since been universally applied in gastrointestinal examinations and in guidance of catheters in cardiovascular imaging. The XRII is a vacuum tube electro-optical device. Its large input phosphor screen (structured CsI) converts incident x-rays to light photons which are then converted to electrons by a photocathode. The electrons are accelerated and focussed through an electric field and converted back to light at a small output phosphor screen. A video camera can be coupled to the output of the XRII to produce x-ray images in real-time (30 frames per second) which are displayed on a video monitor. Because of the amplification of the electron signal provided by the XRII and the efficient coupling between the output screen of the XRII and the video camera, x-ray quantum noise limited images can be produced at relatively low x-ray exposure rates (e.g. 30 μR/second at the input phosphor of the XRII). Since the brightness of the monitor can be adjusted independently from the x-ray exposure, there is no problem of display dimness.

Fluorography is an imaging procedure made possible by the XRII. It permits multiple radiographs to be taken at a high rate. To accomplish this, a 100 mm film camera or 35 mm cine camera is added to the XRII imaging chain. Fluorography is very important for dynamic imaging studies such as swallowing, speech, walking, the joints of both upper and lower extremities, and
most importantly in the diagnosis and treatment of atherosclerosis.

The XRII/video system has become the dominant technology in real-time x-ray imaging. However, it has the following limitations: (1) Veiling glare, which is caused by x-ray, light and electrons inside the XRII, degrades the contrast of the images; (2) Geometric pincushion distortion, which is caused by the curved input screen; (3) Shading, i.e. the output images appear brighter at the centre than at the edge which could cause difficulties in quantitative analysis of the images; (4) Poor image resolution of the XRII/video chain, which is due to light scatter in the input and output phosphor screen, as well as the electron optics and the video camera readout, resulting in a limiting resolution of 2 lp/mm at the full field of view; (5) The need for a sophisticated mechanical supporting system, so as to permit interactive operation involving movement of the heavy XRII/video chain and x-ray tube together; (6) Sensitivity to magnetic field in the environment resulting in significant image distortion in the XRII, which changes as the system is moved. Thus a fluoroscopic system which maintains the advantage of high x-ray sensitivity of the XRII, while overcoming its disadvantages is desirable.

1.2 Development of Digital Radiology

1.2.1 Rationale

Although new digital imaging modalities such as CT, US and MRI have been highly developed, conventional radiology which supplies a large fraction of medical images has remained largely an analog technique. The major reasons are: (1) Technical difficulties, since radiological detectors have to be made with large area and high resolution which are a considerable challenge to present electronic technology; (2) Data handling, since the volume of digital data for each radiological image is much larger than that for other digital imaging modalities and thus they are more difficult
to handle. During recent years, with the rapid development of electronic and computer technology, digital radiological detectors have undergone considerable investigation and development. Advantages of digital radiology include: (1) The image detection function is separated from the display and storage functions, thus they can be optimized independently; (2) Any non-uniformity in the response of the detector can be corrected; (3) Digital image processing (e.g. contrast enhancement, feature segmentation) can improve the image visualization; [12, 13] (4) Computer aided diagnosis (CAD) can be implemented to assist radiologists in finding subtle abnormalities. [14, 15] For example, CAD has been investigated for the detection of microcalcifications [16–18] and masses [19] in mammography, as well as the detection of pneumothorax [20] and pulmonary nodules [21] in chest radiography.

1.2.2 Detectors for digital radiography

There are many different approaches to digital radiography, including scanning and broad x-ray beam geometries, as well as limited-area and large area detectors. Detectors using scanning x-ray beam geometries such as point scan and slit scan were the first methods brought to clinical evaluation because detector technologies for making a single element or a linear array of elements were less demanding. [22] Scanning x-ray geometries can effectively reduce scattered radiation reaching the detectors thus improving image quality, but have the drawback of inefficient x-ray tube utilization resulting in high tube loading and long patient exposure time. A scanning slot x-ray beam geometry has recently been proposed which can significantly reduce tube loading without seriously compromising the advantage of reduction of scattered radiation. [23] On the other hand, broad x-ray beam geometry using an area detector allows digital radiographs to be obtained with a single brief radiation exposure, as in conventional screen-film radiography. It would be attractive for digital radiography because of its high patient throughput and efficient use of the x-ray tube.
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However, it forgoes the advantage of scatter reduction.

From the point of view of detector development, there are two types of digital detectors: limited area and large area detectors. Limited-area detectors can only be used with a scanning geometry, whereas large area detectors can be employed with either a scanning or conventional broad beam geometry. [24] Here, the development of several area detectors for digital radiography will be reviewed.

1.2.2.1 Photostimulable phosphors

Since their commercial introduction over a decade ago, [25] photostimulable phosphor based detectors have become the most widely used digital radiographic detectors. These phosphors are almost all europium-activated barium fluorohalide compounds (e.g. BaFBr: Eu$^{2+}$). When x-ray photons interact with the photostimulable phosphor, the energy absorbed excites electrons in the phosphor. Rather than decaying immediately to give off light, some of the excited electrons are captured in nearby traps. This latent image can be subsequently read out pixel-by-pixel by scanning a finely focused red laser beam across the storage phosphor screen. The trapped electrons at each pixel are stimulated by the red light, freed from the trap and subsequently emit light photons at a shorter wavelength (blue). Thus the number of stimulated light photons is proportional to the x-ray energy absorbed at that pixel. The emitted blue light photons are collected by a light guide and then detected by a photomultiplier tube (PMT), amplified and digitized to form a digital image.

There have been two methods for practical implementation of the photostimulable phosphor technology. The first is in the form of cassettes which can easily fit into an existing x-ray room, and are carried between the x-ray units and a central reader, as shown in Figure 1.4. The second implementation is an integrated system which include both the phosphor plate and the reader, as shown in Figure 1.5. Although the integrated system is bulky, it allows images to be
obtained instantaneously after x-ray exposure without operator intervention. Both systems have been commercialized for general radiography (e.g. chest radiography).

Thus in contrast to screen-film radiography, the detection and display of x-ray images are completely separated in a photostimulable phosphor system. Once an x-ray exposure is detected by the photostimulable phosphor, a prescan of the image determines the proper sensitivity of the PMT. By adjusting the gain of the amplifier, the readout system can have an optimum response to the exposure level and image contrast stored on the photostimulable phosphor. This is more flexible in comparison with the fixed contrast and latitude of a screen-film combination. Therefore with
Figure 1.5: The concept of an integrated photostimulable phosphor system including the phosphor screen, the reader and the erasure lamp.

the use of a photostimulable phosphor system the need for repeated examinations due to improper exposure of the films will be reduced.

However as shown in Figure 1.3, the $DQE(f)$ of a photostimulable phosphor [26] retains the problem seen with screen-films, i.e. rapid decrease of DQE as spatial frequency increases. Although clinical studies have shown that the photostimulable phosphor system is suited to most aspects of general body radiography (including the chest) [27,28], its capability in the detection of subtle, fine detail pulmonary structures such as pneumothorax is inferior to the screen-films. [29] Thus because of the poor DQE of photostimulable phosphor systems at high spatial frequencies, they are not suitable to perform high resolution imaging tasks such as digital mammography [30].
1.2.2.2 Electrostatic systems using x-ray photoconductors

Amorphous selenium (a-Se) is the most highly developed x-ray photoconductor for radiology. When used in x-ray imaging, as shown in Figure 1.2 (c), an electric field is applied across the thickness of the a-Se layer in order for it to produce a measurable signal. When x-rays are absorbed in a-Se, the released electron-hole pairs are drawn to the surfaces of the a-Se layer with very little spread because the applied electric field is perpendicular to the surfaces. Hence the latent charge image on the surface of a-Se is not significantly blurred even if the plate is made very thick to absorb most incident x-ray photons. When a-Se was first used commercially in diagnostic imaging (i.e. in Xeroradiography) [31], it showed image resolution superior to screen-films but due to the deficiency of the toner readout method, it had poor x-ray sensitivity [32].

During recent years, several electronic digital readout methods have been developed for electrostatic charge images on a-Se. The very first approach was the scanning electrometer readout method, in which a linear array of electrometers is held close to the a-Se surface [33,34]. By capacitive coupling with the a-Se, the electrometers sense the electric field produced by the latent charge image. The electrometer array is scanned across the a-Se plate in order to read out the entire charge image. Since the field spreads in the gap between the electrometer and the latent charge image, the spatial resolution of this readout method is limited by the gap size. For example, the dotted line in Figure 1.3 shows the $DQE(f)$ of a prototype electrometer readout system with a 100 $\mu$m gap [34]. In contrast to the rapid drop of DQE with increasing spatial frequency for phosphor screens, the DQE of the electrometer readout method stays constant as spatial frequency increases until reaching the limitation of the readout method (3 lp/mm). Since it is not practical to have very small gaps, it is difficult to apply the electrometer readout method to digital mammography. However, the electrometer readout method has been commercialized for chest radiography.
(Thoravision) [35]. This system, as shown in Figure 1.6, has a 500 \( \mu \text{m} \) layer of a-Se deposited on the surface of a metal drum. The surface of the selenium is charged up to a high potential (\( \sim 3000 \) V) by rotating the drum under a corona charging system (or corotron) in order to sensitize a-Se to radiation. The drum is then stopped for the x-ray exposure. The charge image formed on the surface of a-Se is read out by rotating the drum under an array of electrometers. The geometric image distortion which arises from the projection on the cylindrical drum surface is digitally corrected, and the limiting resolution of the imager is 2.5 lp/mm [35].

![Diagram of the system](image)

Figure 1.6: Chest imager (Thoravision) using electrometer readout of corotron charged a-Se drum.

Various approaches to a photo-induced discharge (PID) laser readout method have also been developed which are better than the electrometer readout method in demonstrating the high image resolution of a-Se based electrostatic systems [36–39]. Shown in Figure 1.7 is a schematic diagram of the air gap PID method [39]. In this method, a long transparent electrode is held close to the a-Se surface, and a laser beam is scanned across the a-Se surface pixel-by-pixel. When the laser beam interacts with the surface of a-Se, it discharges the latent image. The counter-charge induced on
the transparent electrode proportional to the latent image of that pixel flows to a charge sensitive amplifier and is then digitized. Since the discharge of latent image is confined to the region of the laser beam, the resolution of the PID method is defined by the spot size of the scanning laser beam (e.g. 50 µm), and is independent of the gap. This method has been shown to be particularly suited to digital mammography. Variations on the PID readout method that are suitable for general radiography have also been proposed [40].

![Diagram showing the concept of air-gap photoinduced discharge (PID) readout method for a-Se.](image)

Another laser based readout method combines digital readout with conventional toner development [41]. In this method, phosphor coated liquid toner was used to develop the latent charge image formed by a-Se. Then a laser beam is scanned across the surface and the light emitted from the luminescent toner is detected and digitized.
These α-Se digital systems based on electrometer and laser readout methods have been successfully used to produce high quality radiographs, better than those produced by photostimulable phosphor systems. However, they have the same practical disadvantages as the photostimulable phosphors. The systems can either be implemented as cassette based detectors which require operator intervention, or as an integrated system (e.g. Thoravision for digital chest radiography) which is bulky. Therefore in order to fully utilize the imaging potential of α-Se, an image readout method with instantaneous readout and a compact size in the form of a cassette is desirable.

1.2.3 Digital fluoroscopy/fluorography

Digital fluoroscopy/fluorography systems have been developed based on digitization of the output signal from a XRII/video chain [42–44]. The method used was to suspend the electron beam scanning of the video camera during a short pulse of x-rays so that the charge image can be integrated on the target of the video camera. Then the electron beam scanning is resumed, reading out the image which is then digitized. Digital fluorography is playing an important role in dynamic imaging studies, especially for cardiac and vascular diseases. Digital subtraction angiography (DSA) is an important example of this approach which is used in imaging of blood vessels. In DSA, contrast agent (iodine) is injected to the blood vessels through a catheter, and digital images are acquired both before (mask) and after (contrast-containing) the injection of contrast. Then the mask image is logarithmically subtracted from the contrast-containing images to obtain an image of the blood vessels alone.

The images obtained using digital fluoroscopic / fluorographic systems have also been used for volume image reconstruction [45, 46] and quantitative analysis for diagnosis of cardiovascular diseases [47]. In these applications the geometric distortion in the images needs to be corrected [48–50].
Although the concept of adding a digitization stage to a XRII/video system is simple and has had many clinical applications, the intrinsic limitations of the XRII still exist: reduced contrast due to veiling glare, geometric distortion, shading, poor resolution, bulky and sensitivity to magnetic field. They need to be improved or replaced by better real-time imaging methods.

1.2.4 Flat Panel Detectors for Both Radiography and Fluoroscopy

In the previous section, digital x-ray imaging detectors developed for either radiography or fluoroscopy were discussed. Recently, the rapid development of large area integrated circuits makes possible the concept of flat-panel, solid-state image detectors for this purpose.

In the past decade, there has been intensive effort in the development of large-area integrated circuits. This is in contrast to the very large scale integrated (VLSI) technology using crystalline silicon where the size of each element becomes smaller in order to build more sophisticated circuits on the same silicon chip. The applications of large-area integrated circuits include large screen displays and large area image sensors, where the challenge is to make the integrated circuit operate over a very large area. The semiconductors that have been used in this technology are amorphous, such as hydrogenated amorphous silicon (a-Si:H), or polycrystalline materials, such as polycrystalline silicon (poly-Si) or cadmium selenide (CdSe).

A major type of large area integrated circuit is an active matrix which consists of a two dimensional array of thin film transistors (TFTs) and pixel electrodes. The motivation for this development is to make flat-panel active matrix liquid crystal displays (AMLCD) for portable computers [51]. Another important type of large area integrated circuit is a large area image sensor made by integrating photodiodes at each pixel of the array of TFTs, and it has been applied to image scanning and image detection of charged particles, gamma rays and light [52,53]. Currently, the largest active matrix TFT array reported is 27.6 cm x 18.4 cm in size with 90 \( \mu \)m square
TFT-photodiode arrays have been made in size of 26 cm x 26 cm with 508 μm square pixels, or 20 cm x 25 cm in size with 127 μm square pixels. Further increases in area, up to 60 cm x 60 cm and decreases in pixel size are anticipated later this decade. This technology of large area integrated circuits appear promising for x-ray imaging, and two different approaches have been proposed: (1) the indirect method, which uses a phosphor screen to convert x-rays to light and then obtain the image with an optically sensitive TFT-photodiode array; (2) the direct method, which uses a photoconductor to directly convert x-rays to a charge image and then readout with an active matrix TFT array.

1.2.4.1 The Indirect Method

The indirect method for making flat-panel detectors, as shown in Figure 1.8, has been studied by several investigators for digital diagnostic and portal imaging. It employs an x-ray scintillator to convert incident x-rays to an optical image, which is then detected with a large area flat-panel image sensor. Figure 1.8 (a) is the cross-section of a pixel, which consists of an individual a-Si:H photodiode on the optical image sensor to detect light generated from the continuous x-ray scintillator, and the photodiode is connected to a TFT for image readout. Shown in Figure 1.8 (b) is a schematic diagram of image readout. The TFTs act as electronic switches. The on and off state of the TFT is controlled by the potential on its gate. The pixels in the same row share a common gate control line; the pixels on the same column share a common data line which is connected to an external charge amplifier. During image readout, an external scanning control circuit generates a signal to turn on the TFTs one row at a time and transfer the charge signal from the pixels to the external charge amplifiers. The signal is then multiplexed and digitized. The total readout time for the detector is proportional to the number of rows. The readout can be in real-time (30 frames per second) for detector sizes suitable for fluoroscopy. Therefore the flat-panel detectors based on
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TFT readout can potentially be used for both digital radiography and fluoroscopy.

As a flat panel detector, the indirect method has many advantages over the other approaches for digital radiology. It is compact and provides instantaneous readout. It also has many advantages over the digital fluoroscopic/fluorographic systems based on XRII/video chain: (1) no problem of veiling glare, since the readout method is electronic thus no electron-optics are involved; (2) freedom from pin-cushion image distortion, since the detector is flat; (3) unaffected by magnetic field; and (4) light in weight. However, the use of phosphor screens or CsI for x-ray detection retains the same problem of rapid decrease of DQE when spatial frequency increases which results in a sub-optimal imaging system, especially for tasks requiring high resolution.

1.2.4.2 The Direct Method

In order to overcome the inherent limitation of $DQE(f)$ of phosphor screens, we have proposed an electrostatic or direct x-ray conversion flat-panel detector. In the direct method, as shown in Figure 1.9, a uniform layer of x-ray sensitive photoconductor (e.g. a-Se) serves as the radiation detector and directly converts the incident x-rays to a charge image. Under a positive high voltage applied to the top surface of a-Se, the holes generated by radiation in a-Se are driven by the electric field to the bottom surface of the a-Se layer, where they are collected by an active matrix. Each pixel of the active matrix consists of a pixel electrode for charge collection; a storage capacitor for holding the image charge; and a TFT for image readout. During image readout, similar to the indirect method, the external scanning control circuit generates a signal to turn on the TFTs one row at a time and transfer the charge image from the pixel electrodes to the external charge amplifiers, and the readout rate can be in real-time. The direct method not only has the same practical advantages of flat panel detectors, but also has two advantages over the indirect method: (1) better $DQE(f)$ due to the electrostatic image formation; and (2) simpler structure for the large area integrated
Figure 1.8: Concept of the indirect method for making flat-panel detectors: (a) Cross-sectional view of one pixel; (b) schematic diagram showing the image readout.
circuit since the need for a photodiode at each pixel is eliminated.

Figure 1.9: Schematic diagram showing the structure of a flat-panel radiological detector using the direct method. The detector incorporates a layer of amorphous selenium for directly converting x-rays to charge and an active matrix for charge readout.

Recently, a variation of the direct method was proposed in which a thick dielectric layer (comparable to the thickness of a-Se) was added between the a-Se layer and the top bias electrode [60]. During x-ray exposure, the holes generated by radiation are driven to the bottom surface of the a-Se and collected by the active matrix, similar to our direct method. However the electrons generated by radiation, which are driven to the top surface of a-Se, are trapped at the interface between the a-Se and the dielectric layer (instead of going to the top bias electrode as in our direct method). Due to the procedure necessary to remove trapped electrons at the interface between
the a-Se and the dielectric layer, this variation cannot be applied to real-time x-ray imaging and is therefore restricted to radiography.

1.3 Potential Impact of the Proposed Research

It is the objective of this work is to understand how to make a direct, real-time flat-panel x-ray imaging detector practical and reliable, and with optimum detective quantum efficiency that is superior to other digital x-ray imaging detectors. Once such a detector (which is also referred to as the self-scanned a-Se detector) is available, it will have a significant impact on how digital radiology is performed.

1.3.1 Radiography

A direct, real-time flat-panel detector is in the form of a cassette which can easily fit into an existing radiography room. Because the detector is integrated with the electronic readout system, images can be obtained instantaneously after the x-ray exposure without operator intervention. These practical advantages combined with the high DQE at all spatial frequencies provided by a-Se will make our approach ideal for digital radiography. The instantaneous availability of high quality digital radiographs will be of great advantage in many clinical procedures, and here imaging guided breast needle biopsy will be used as an example.

In breast needle biopsy, the patient’s breast is immobilized by compression and two mammograms are taken at different angles to provide a stereotaxic view for guiding the placement of the needle. During the entire biopsy procedure, at least 3 sets of mammograms need to be obtained: before insertion of needle; after needle insertion and before biopsy; and after biopsy (for verification) [61]. With conventional screen-film mammography, the patient has to remain immobilized
during each of the approximately 5 minute delays while the film is processed and evaluated. With the aid of the instant digital mammograms produced by the direct, flat-panel detectors, the time needed for the whole procedure can be greatly reduced, thus reducing the discomfort of the patient.

1.3.2 Fluoroscopy

The conventional XRII/video systems used in fluoroscopy are heavy and bulky, thus they require sophisticated mechanical supporting system in order to move the XRII/video chain and the x-ray tube together during interactive operations. Replacing the XRII/video system with the compact and light flat-panel detector will make both the detector and the entire mechanical system simpler and more compact. This is especially important for interventional procedures, where the movement of the radiologist is often restricted by the bulky XRII/video system.

Since the flat-panel detectors uses electronic image readout, it eliminates the fundamental problems of veiling glare of an XRII caused by light and electron scatter inside the XRII [10]. Thus the images will have much better contrast than those produced by an XRII which will facilitate the detection of low contrast objects.

The flat-panel detectors do not have geometric image distortion (e.g. pincushion distortion) as in a curved detector such as an XRII. Unlike the XRII, the flat-panel detectors are not affected by magnetic field and therefore they are ideal for three dimensional (3D) volume imaging in interventional procedures. XRII based volume imaging systems have been investigated for reconstruction of 3D vascular images during interventional procedures by acquiring multiple images while rotating the XRII around the patient. However before reconstruction can be performed, the images have to be corrected for geometric distortion due to the curved XRII input screen and the effect of the earth's magnetic field [45]. The use of a flat-panel detector will not only allow a much lighter system that is easier to rotate, but also provide distortion free images that require essentially no
In some clinical procedures, such as image guided therapy, the combination of x-ray fluoroscopic imaging and MRI is advantageous since MRI provides high sensitivity to soft tissue lesions and has the capability of monitoring temperature changes, whereas x-ray imaging provides high image update rate and better signal to noise ratio, as well as better visualization of metallic objects (i.e. implants) and bone than MRI. However because the conventional XRII/video system is very sensitive to magnetic field, it can not be placed close to an MRI system. The use of a real-time flat-panel detector will therefore allow the combination of x-ray imaging and MRI for better guidance of interventional procedures.

1.4 Outline of the Thesis and Summary

During my previous work [62], I performed a theoretical study of the feasibility of our direct, flat-panel detector. The potential signal and noise performance of the detector at zero spatial frequency was analyzed in order to find whether the detector can be x-ray quantum noise limited for all x-ray imaging applications (radiography and fluoroscopy).

The goal of the present work is to investigate the practical difficulties in making the detector feasible and reliable, and also to study the imaging performance of the detector in the form of DQE for all spatial frequencies in order to optimize the detector design for each x-ray imaging application.

The first factor that can affect the practical feasibility and reliability of the detector is the radiation hardness of the TFT, since in the region of the detector that is exposed to radiation, the TFTs are the only components susceptible to radiation damage. Therefore in Chapter 2, an experimental study of the radiation hardness of the CdSe TFTs used in our investigation is described. Based on our experimental results and well established theory for the mechanism of radiation dam-
age to crystalline silicon metal-oxide-semiconductor field effect transistors (MOSFETs) which have similar structure and principle of operation as TFTs, the factors affecting the radiation hardness of TFTs and their dependence on the design parameters of the TFTs have been established. This will provide a scientific basis for selection of TFT design in order to ensure reliable operation of the detector for its life-time. A paper based on this chapter has been submitted to Medical Physics [63].

The second factor that can affect the feasibility and reliability of the detector is the high voltage applied to the top surface of the a-Se layer in order to sensitize it to radiation, and how irreversible damage to the TFTs (which are at the bottom surface of the a-Se layer) is to be avoided. Therefore in Chapter 3, the conditions under which the high voltage applied to a-Se can damage the TFTs are discussed. A novel concept for making TFTs with self-protection against high voltage damage is described. It incorporates a dual-gate design for the TFTs, where the bottom gate is the ordinary scanning control gate and the top gate is the high voltage protection gate. Compared to the approach of adding a second TFT at each pixel for high voltage protection previously proposed by other investigators, the advantage of our dual-gate design is its reduced complexity. Experimental measurements on dual-gate TFT samples demonstrate the feasibility of high voltage protection. A theoretical model for dual-gate TFTs, which was verified by experimental measurements on TFT samples, has been developed in order to predict high voltage protection characteristics for different dual-gate TFT design parameters. The pixel x-ray response for direct, flat-panel detectors with high voltage protection has also been predicted for different x-ray imaging applications. This chapter has been submitted as a paper to Medical Physics [64].

In Chapter 4, a theoretical analysis is performed for the spatial frequency dependent DQE of the direct, flat-panel detectors. This analysis allows us to understand all the major factors affecting the $DQE(f)$ and thereby maximizing $DQE(f)$ during detector design. The results of DQE analysis
CHAPTER 1. INTRODUCTION

will also provide a basis for comparing the imaging performance of the direct flat-panel detectors with other types of digital x-ray imaging detectors. This chapter has been submitted as a paper to Medical Physics [65].

In Chapter 5, the construction and evaluation of a prototype flat-panel detector are presented. The experimental measurements of x-ray sensitivity, resolution [in the form of modulation transfer function (MTF)] and spatial frequency-dependent DQE all agree well with theoretical predictions. The experimental results not only demonstrated the practical feasibility of making an operational flat-panel detector, but also verified our understanding of all the factors contributing to $DQE(f)$ of our detector, therefore allowing optimization of $DQE(f)$ during future detector design for clinical x-ray imaging applications. This chapter has been submitted as a paper to Medical Physics [66].

The future work, which is described Chapter 6, will involve investigating practical solutions for maximizing $DQE(f)$ for various x-ray imaging applications. Several factors concerning the design for fluoroscopy and mammography are described.

In summary, a direct, real-time flat-panel detector using an x-ray photoconductor (e.g. a-Se) and active matrix has the potential to perform all radiological imaging tasks with high $DQE(f)$. With the present work, I have established all the factors affecting the practical feasibility of the detector, and strategies for ensuring the detector reliability and maximizing $DQE(f)$ for specific x-ray imaging applications. It is expected that the cost of large area integrated circuits will decrease with their growing applications in AMLCDs and image sensors. Once a digital imaging system of high image quality and low cost is feasible, digital radiology and its advantages will become more widespread.
Chapter 2

Radiation Hardness of Cadmium Selenide Thin Film Transistors

2.1 INTRODUCTION

Recently, there have been intensive investigations into the feasibility of making large area, flat-panel x-ray imaging detectors for digital radiology. The flat-panel detectors utilize the technology of large area integrated circuits made with amorphous [e.g. hydrogenated amorphous silicon (a-Si:H)] or polycrystalline semiconductor materials [e.g. polycrystalline cadmium selenide (CdSe)]. The large area integrated circuit consists of a two-dimensional array of thin film transistors (TFTs), and is often referred to as an active matrix. An x-ray imaging sensor has been previously proposed which employs a layer of x-ray photoconductor to convert x-rays to charge and an active matrix for self-scanned readout [67,68]. In Figure 1.9, such a sensor is made by depositing a photoconductive layer of amorphous selenium (a-Se) on the surface of an active matrix. A constant electric field is

\footnote{This chapter has been submitted as a paper to Med. Phys., entitled “Digital radiology using active matrix readout of amorphous selenium: Radiation hardness of cadmium selenide thin film transistors.”}
applied across the a-Se layer through a bias electrode deposited on its top surface. When x-rays are absorbed in the a-Se layer, electrons and holes are created in pairs and drawn to the appropriate surfaces by the applied electric field. With positive potential on the bias electrode, holes are driven to the bottom surface and collected by each pixel electrode and stored on the pixel capacitors of the active matrix. The charge image can then be read out by the active matrix by turning on the TFTs one row at a time using their common gate lines. The image charge on all the pixels of the selected row is transferred to the corresponding vertical (data) lines and integrated by the charge amplifiers. The multiplexer then converts these signals to a serial output for subsequent digitization. The image acquisition of the detector can be in real-time (i.e. 30 frames per second), therefore it can potentially be applied in both fluoroscopy and radiography. Self-scanned active matrix detectors based on indirect conversion of x-rays using scintillator (e.g. CsI) and photodiode arrays have been proposed by other investigators [56-59]. The potential advantages of the direct conversion of x-rays to charge versus the indirect approach are the higher image resolution provided by electrostatic image formation, and easier adaptation of active matrix technology developed for liquid crystal displays (LCDs), because there is no need for photodiodes on the active matrix.

The active matrices used in our investigation are made with CdSe semiconductor. It is important to ensure that the normal operation of the TFTs will not be damaged by the level of radiation received during the life-time of the detector. In this chapter, possible mechanisms of radiation damage to TFTs are described based on known effects of radiation on n-channel crystalline silicon (c-Si) metal-oxide-semiconductor field effect transistors (MOSFETs) and a-Si:H TFTs, which have similar structure and principle of operation as CdSe TFTs. Experimental results of radiation hardness studies using individual CdSe TFT samples will be presented. The radiation dosage used in the experiments started at 50 Gy which is an estimate of the life-time dose expected
for a diagnostic x-ray imaging detector (0.05 cGy / image x 100 images / day x 200 days / year x 5 years), and increased up to 1000 Gy.

2.2 BACKGROUND

The structure of a CdSe TFT is shown with its cross-section in Figure 2.1. The entire structure is built on a glass substrate. The metal gate (usually made of chromium) is on the bottom, followed by the silicon dioxide (SiO$_2$) gate insulator layer formed using plasma enhanced chemical vapor deposition (PECVD). The CdSe thin film is deposited using thermal evaporation in amorphous form and recrystallized at moderate temperature (<400°C), and then covered with the top oxide (a passivation layer of SiO$_2$). The drain and source electrodes are formed above these layers and connected to the CdSe through vias, i.e. holes etched in the top oxide. The thickness of the gate insulator used in CdSe TFTs is typically 400 nm, which is almost an order of magnitude higher than that used in c-Si MOSFETs and twice as much as that of the silicon nitride (Si$_3$N$_4$) gate insulator used in a-Si:H TFTs. These thick SiO$_2$ gate insulation layers are a main factor contributing to the high voltage capabilities for CdSe TFTs, which can withstand drain voltages of hundreds of volts [69, 70]. The CdSe films are n-type, and the TFTs have similar principle of operation as n-type c-Si MOSFETs. The significant difference is that the semiconductor layer in CdSe TFTs is typically 50 nm thick, which is much thinner than the semiconductor wafers (a few hundred microns) used in c-Si MOSFETs.

The mechanisms of damage by ionizing radiation in c-Si MOSFETs have been well studied, and four parameters (discussed further below) were found to change as a result of radiation: threshold voltage, subthreshold swing, leakage current and field effect mobility [71]. Studies reveal that radiation also affects all these characteristic parameters in a-Si:H TFTs with the exception
of field effect mobility [72-74]. In this section, possible mechanisms of the effect of radiation on
the characteristic parameters of CdSe TFTs are examined, based, where relevant, on the previous
work on MOSFETs and a-Si TFTs.

2.2.1 Threshold voltage

The principle of operation of TFTs is based on the field effect, as in MOSFETs. As shown in
the energy band diagram in Figure 2.2 (a), the Fermi energy level $E_F$ of CdSe is closer to the
conduction band edge $E_C$ than to the valence band edge $E_V$, similar to the situation in an n-type
MOSFET. When a positive voltage is applied to the gate ($V_G > 0$), as shown in Figure 2.2 (b),
the conduction band $E_C$ bends towards $E_F$. Under this condition, free electrons are generated in
the CdSe layer and form a channel. With a source-drain voltage applied, current will flow along
the CdSe channel, i.e. the TFT is turned on. The threshold voltage of a TFT, $V_T$, is the smallest
gate voltage required to turn on the TFT. It is usually determined from the transfer characteristic
curve, i.e. the channel current $I_D$ as a function of gate voltage $V_G$. When $V_G$ is smaller than $V_T$,
the electrons created in CdSe are filling the localized states in the semiconductor band gap. When
$V_G$ is greater than $V_T$, some electrons are in the conduction band and hence free to move in the channel and the CdSe layer is in the conductive state.

When radiation interacts with the SiO$_2$ gate insulator of CdSe TFTs, it is expected that positive space charge will be created in the gate insulator, as found in n-type MOSFETs using SiO$_2$ gate insulators [75]. When radiation is absorbed by the SiO$_2$ layer, electrons and holes are generated and driven towards opposite surfaces of the SiO$_2$ layer under the influence of the electric field $E_{ox}$. Since electrons have much higher mobility ($20 \text{ cm}^2/\text{Vs}$) in SiO$_2$ than holes ($10^{-11} \text{ cm}^2/\text{Vs}$), and the trap density of electrons ($10^{14} \text{ cm}^{-3}$) is much lower than that of holes ($10^{18} \text{ cm}^{-3}$), most electrons are swept out of the gate insulator (within $10^{-12}$s) and holes are left behind. A fraction of these holes, which strongly depends on the material properties of SiO$_2$, will be trapped in deep trapping sites in the gate insulator. As shown in Figure 2.2 (c), holes trapped in the SiO$_2$ layer cause the conduction band in CdSe to bend toward $E_F$, in the same way as a positive $V_G$. Therefore trapped holes cause a negative shift of the threshold voltage $V_T$.

Under positive gate bias, as shown in Figure 2.3 (a), holes are driven towards the semiconductor-insulator interface (subsequently referred to as semiconductor interface) and are therefore preferentially trapped there. Under negative bias [Figure 2.3 (b)], holes are trapped close to the gate-insulator interface (subsequently referred to as the gate interface) [76]. The relationship between $V_T$ shift, $\Delta V_T$, and the distribution of charge trapped in the gate insulator is given by [71]:

$$\Delta V_T = -\int_0^{d_{ox}} \frac{n_{ht}(x)x dx}{\epsilon_{ox} \epsilon_0}, \quad (2.1)$$

where $\epsilon_{ox}$ and $d_{ox}$ are the dielectric constant and the thickness of the gate insulator, $\epsilon_0$ is the permittivity of free space, $n_{ht}(x)$ is the density of holes trapped at distance $x$ from the gate interface. From this expression it is evident that holes trapped near the gate interface have much less effect on the threshold voltage shift than the holes trapped near the semiconductor interface.
Figure 2.2: Energy band diagram of a CdSe TFT: (a) flat-band condition under zero gate bias; (b) band bending under positive gate bias and creation of electrons in the CdSe layer; (c) with positive space charge in the gate insulator and zero gate bias, electrons are created in the CdSe layer.
The mean location of trapped charge in the gate insulator can be represented by the charge centroid $x_{tr}$ defined as:

$$x_{tr} = \frac{\int_0^{d_{ox}} n_{ht}(x) x \, dx}{\int_0^{d_{ox}} n_{ht}(x) \, dx}. \quad (2.2)$$

In MOSFETs, it was found for SiO$_2$ layers with ordinary thicknesses ($<50$ nm), the density of trapped holes peaks near the interfaces and falls off rapidly with distance from the interface, as shown in Figure 2.3. In thicker SiO$_2$ layers ($>100$ nm), there is substantial hole trapping in the bulk of the oxide [71]. A special case for bulk trapping is when a uniform trap density is assumed throughout the thickness of the gate insulator and the trapping is relatively weak. The distribution of the trapped holes will be triangular, skewed towards the interface with semiconductor or the gate depending on the direction of $E_{ox}$ [77], as shown in Figure 2.4. The charge centroid $x_{tr}$ can then be calculated, using Eq. 2.2, as $\frac{2}{3}d_{ox}$ and $\frac{1}{3}d_{ox}$ for positive and negative gate bias, respectively.

Based on the radiation response of SiO$_2$ in n-type MOSFETs, the following model of the dependence of threshold voltage shift on the gate insulator material properties was developed for CdSe TFTs. When ionizing radiation is absorbed by the gate insulator, the charge generated per unit area, $\Delta N_r$, is given by:

$$\Delta N_r = \frac{e D \rho_{ox} d_{ox}}{W_{\pm}}, \quad (2.3)$$

where $e$ is the electronic charge, $D$ is the radiation dose, $\rho_{ox}$ is the density of SiO$_2$ and $W_{\pm}$ is the energy required to create an electron-hole pair in SiO$_2$. $W_{\pm}$ is electric field dependent and was found to be approximately 17 eV for $E_{ox} > 4$ MV/cm, and increases as $E_{ox}$ decreases [78]. Then substituting the values for $\Delta N_r$ (Eq. 2.3) and $x_{tr}$ (Eq. 2.2) into Eq. 2.1, $\Delta V_T$ is given by:

$$\Delta V_T = -\frac{e D \rho_{ox} d_{ox}}{\varepsilon_{ox} \varepsilon_0 W_{\pm}} f_T x_{tr}, \quad (2.4)$$

where $f_T$ is the fraction of hole trapping in the gate insulator, determined by the product of hole trapping cross section $\sigma_{ht}$ and the hole trap density $N_{ht}$. (In MOSFETs, it was found experimentally
Figure 2.3: Schematic diagram illustrating the mechanism for TFT threshold voltage shift by irradiation: (a) hole trapped near the interface between the gate insulator and the semiconductor under a positive gate bias voltage; (b) hole trapped near the interface between the gate and the gate insulator under a negative gate bias voltage. $E_{ox}$ is the electric field across the gate insulator.
Figure 2.4: Triangular distribution of trapped holes in the gate insulator under the assumption of uniform hole trap density throughout the thickness of the gate insulator and relatively weak trapping: (a) Under positive gate bias, holes are created uniformly in the gate insulator by the radiation, and drawn towards the semiconductor interface; (b) Under negative gate bias, holes are uniformly created in the gate insulator by the radiation, and drawn towards the gate interface.
that $f_T$ could range from 0.01 to 1 depending on the material properties of the SiO$_2$ [71]). Eq. 2.4 shows that the largest value of $|\Delta V_T|$ results when a positive gate bias is applied during irradiation and all the holes trapped are at the semiconductor interface, i.e. $x_{tr} = d_{ox}$. Under these conditions Eq. 2.4 shows that $\Delta V_T$ is proportional to $d_{ox}^2$. Figure 2.5 shows the values for $|\Delta V_T|$ calculated as a function of $d_{ox}$ for $D = 50$ Gy (lifetime dose) and for $f_T = 0.2$ and 1. If a relatively thin layer of SiO$_2$ ($d_{ox} = 100$ nm) and a low $f_T$ ($f_T = 0.2$) are used, $|\Delta V_T|$ is 0.4 V which is insignificant. However with a thicker SiO$_2$ layer ($d_{ox} = 500$ nm) and $f_T = 1$, $|\Delta V_T|$ could be as high as 40 V, which makes the normal operation of a transistor impossible. Therefore the material properties and the thickness of the gate insulator are very important for maintaining adequate radiation hardness for a practical x-ray imager.

Figure 2.5: Calculated threshold voltage shift, $-\Delta V_T$, due to radiation dose of 50 Gy as a function of gate insulator (SiO$_2$) thickness. Holes were assumed to be trapped at the gate insulator - semiconductor interface with a trapping fraction of $f_T$. 
2.2.2 Subthreshold swing

The subthreshold region of the transfer characteristic curves of a CdSe TFT refers to the exponential increase of $I_D$ as a function of $V_G$ before the semiconductor channel starts to fully conduct. In this region the majority of the electrons created in the CdSe layer are utilized to fill the localized states in the CdSe bandgap. The subthreshold swing, $S$, is defined as the change in gate voltage required to increase $I_D$ by an order of magnitude [79]:

$$S = \frac{dV_G}{d(\log I_D)}.$$

The subthreshold swing of the TFT is closely related to the density of localized states $D_{it}$ in the bandgap. In fact $S$ is proportional to $D_{it}$ as [79]:

$$S = \frac{kT \ln(10) D_{it}}{C_{ox}},$$

where $k$ is Boltzmann's constant, $T$ is the absolute temperature, and $C_{ox}$ is the capacitance of the gate insulator per unit area. Thus the higher the density of electron localized states, the shallower the subthreshold region and the larger the subthreshold swing. It has been shown in both MOSFETs and $\alpha$-Si:H TFTs that radiation generates localized states in the semiconductor band gap and thus causes an increase in the subthreshold swing [74,80,81]. Therefore a similar effect of radiation can be expected in CdSe TFTs.

2.2.3 Field effect mobility

The field effect mobility $\mu_{FE}$ is a measure of the drift velocity of the electrons in the semiconductor channel traveling under a source-drain bias voltage. Compared with the electron mobility $\mu_0$ in a perfect crystal, $\mu_{FE}$ is smaller due to: (1) the trapping and releasing of electrons by trap states just below the conduction band, e.g. the conduction band tail states in $\alpha$-Si:H TFTs; (2) the interface
traps in the gate insulator located within one or two atomic bond distances (\(\sim 0.5 \text{ nm}\)) from the semiconductor layer, which causes electrons in the semiconductor conduction band to make quantum mechanical transitions into and out of these traps. In c-Si MOSFETs, a considerable decrease in \(\mu_{FE}\) was found after irradiation which was attributed to increase of interface traps in SiO\(_2\) [82,83]. In a-Si:H TFTs, radiation was found to have no effect on \(\mu_{FE}\) [74]. Either observation could be expected for CdSe TFTs.

2.2.4 Leakage current

Leakage current \(I_{DL}\) is the channel current when the TFT is in the off state under a negative gate voltage, e.g. \(V_G = -5 \text{ V}\) in x-ray imaging operations [84]. In CdSe TFTs, \(I_{DL}\) is usually between \(10^{-14}\) and \(10^{-13} \text{ A}\), which is comparable to that in a-Si:H TFTs and much smaller than that in c-Si MOSFETs.

Radiation causes an increase in \(I_{DL}\) in both MOSFETs and a-Si:H TFTs. In MOSFETs it was found that irradiation could increase the leakage current by as much as an order of magnitude, which was attributed to an increase of interface state density by irradiation which increased the generation of carriers in the semiconductor film [85,86]. In a-Si:H TFTs, the increase of \(I_{DL}\) is due to a combination of the negative \(V_T\) shift and the shallower subthreshold slope, which brought \(V_G = -5 \text{ V}\) into the subthreshold region of the characteristic curves. Increase in \(I_{DL}\), similar to that in a-Si TFTs, is expected for CdSe TFTs since similar changes in threshold voltage and subthreshold swing are expected.
2.3 MATERIALS AND METHODS

2.3.1 TFT structure and geometry

The structure of the TFTs used in the radiation damage study is shown in Figure 2.1, and described in section 2.2. The layer thicknesses are summarized in Table 2.1 along with the details of the geometry of the CdSe channel (i.e. channel width \( W \) and length \( L \)).

<table>
<thead>
<tr>
<th>gate insulator thickness (nm)</th>
<th>400</th>
</tr>
</thead>
<tbody>
<tr>
<td>top oxide thickness (nm)</td>
<td>330</td>
</tr>
<tr>
<td>CdSe thickness (nm)</td>
<td>50</td>
</tr>
<tr>
<td>gate insulator capacitance (nF cm(^{-2}))</td>
<td>8.6</td>
</tr>
<tr>
<td>channel width ( W ) (( \mu )m)</td>
<td>36</td>
</tr>
<tr>
<td>channel length ( L ) (( \mu )m)</td>
<td>25,30,35,40,50,60</td>
</tr>
<tr>
<td>radiation dose (Gy)</td>
<td>50,100,200,1000</td>
</tr>
<tr>
<td>gate bias ( V_{GB} ) (V)</td>
<td>-20,-10,-5,0,10,20</td>
</tr>
</tbody>
</table>

Table 2.1: TFT geometry, gate bias, and irradiation parameters used in the radiation damage experiments.

2.3.2 Irradiation parameters

The irradiations were performed using a \(^{60}\)Co source (1.25 MeV photon energy). The TFT samples were placed 80 cm from the source in a \(^{60}\)Co radiation therapy unit. A 2 mm thick aluminum (Al) build-up layer was placed on top of the TFT samples. The dose rate (to water) was measured as 1.92 Gy / minute. This was then multiplied by 0.85 to obtain the dose to the SiO\(_2\) gate insulator (1.63
Gy / minute) based on the differences in stopping power ratio between water, Al and SiO₂ [87].

The $I_D - V_G$ characteristic curves of the TFTs were measured before irradiation and within 3 hours after the accumulated dose of 50, 100, 200 and 1000 Gy. During irradiation, the source and drain of each TFT were connected together and grounded, and a gate bias voltage $V_{GB}$ was applied. In order to cover the entire range of possible gate bias for TFT operation, the values of $V_{GB}$ chosen were: 0, -5, ±10 and ±20 V. Each value of $V_{GB}$ was applied to a group of six TFTs with the same structure and channel width of 35 μm, but with different channel length ranging from 25 to 60 μm. The details of the accumulated radiation dose and $V_{GB}$ applied to each type of TFTs are summarized in Table 2.1. $V_{GB}$ was disconnected within one minute after irradiation and measurement of characteristic curves started. After completion of irradiation experiments, the room temperature recovery of the TFT characteristics was monitored for a time period of 30 days with $V_{GB} = 0$ established by connecting all three terminals of a TFT together.

### 2.3.3 Methods for measuring characteristic parameters

The $I_D - V_G$ curves of the TFTs were measured using an HP4156A semiconductor parameter analyzer. Four parameters were extracted from each $I_D - V_G$ curve: threshold voltage, subthreshold swing, field effect mobility, and leakage current. Figure 2.6 shows an example $I_D - V_G$ curve. The methods for extracting the four parameters are as follows:

#### 2.3.3.1 Threshold voltage

The threshold voltage $V_T$ was defined as the value of $V_G$ required to generate a given channel current $I_{DC}$. The value of $I_{DC}$ was selected to be 1 nA, as illustrated in Figure 2.6. This is to be compared with the TFT on current of the order of several μA. A typical $V_T$ for the CdSe TFTs used in the experiments is 1 V.
Figure 2.6: A typical TFT transfer characteristic curve, i.e. channel current $I_D$ as a function of gate voltage $V_G$. The curve is plotted with $I_D$ in both logarithmic and linear scale: the left ordinate shows $I_D$ in logarithmic scale and the right ordinate is $I_D$ in linear scale. This set of curves is used to illustrate the definition of threshold voltage $V_T$, subthreshold swing $S$, and slope $g_m$ (used to determine the field effect mobility $\mu_{FE}$ at gate voltage $V_{G0}$).

2.3.3.2 Subthreshold swing

The subthreshold swing, $S$, is extracted as the inverse of the slope of the $I_D - V_G$ curve (with $I_D$ in logarithmic scale) in the subthreshold region: $S = dV_G / d(\log I_D)$. $S$ was measured at $I_D = 1$ nA in the experiment, as illustrated in Figure 2.6.

2.3.3.3 Field effect mobility

In CdSe TFTs, there is a factor that can potentially affect the measurement of $\mu_{FE}$ due to the polycrystalline structure of the CdSe layer, as shown in Figure 2.7 (a). At the grain boundaries
in polycrystalline CdSe films, there are a large number of electron traps. These traps, when filled, form a potential barrier $E_B$ between individual CdSe grains, as shown in Figure 2.7 (b). The potential barrier $E_B$ reduces the number of free electrons, and hence the channel current $I_D$, by $\exp(-E_B/kT)$. This effect is regarded as a reduction in the effective $\mu_{FE}$ by $\exp(-E_B/kT)$ [88].

When the positive gate voltage $V_G$ increases, more free electrons are generated in the CdSe grains, which reduce the height of $E_B$. Hence the effective $\mu_{FE}$ in CdSe TFTs shows a gate voltage dependence. Usually when $V_G$ is high, e.g. $> 30$ V, the height of $E_B$ is negligible and $\mu_{FE}$ is independent of $V_G$. This factor needs to be taken into consideration during the measurement of $\mu_{FE}$.

The value of $\mu_{FE}$ was measured at $V_G = 35$ V where $\mu_{FE}$ is no longer a function of $V_G$. When $\mu_{FE}$ is independent of $V_G$, it can be determined from the transconductance $g_m$ ($g_m = dI_D/dV_G$) using [79]:

$$\mu_{FE} = g_m \frac{L}{C_{ox} V_{DS} W},$$

where $V_{DS}$ is the voltage across the drain and source of the TFT. As shown in Figure 2.6, $g_m$ was measured as the tangent to the $I_D - V_G$ curve in linear scale. After the irradiation, $\mu_{FE}$ was measured at $V_G = 35$ V + $\Delta V_T$ to take into account the threshold voltage shift.

### 2.3.3.4 Leakage current

As shown in Figure 2.6, the leakage current $I_{DL}$ was determined as the $I_D$ at $V_G = -5$ V, which is the gate voltage used in x-ray imaging applications to turn off the TFTs [89].
Figure 2.7: The grain boundary barrier in a CdSe layer: (a) diagram showing the polycrystalline structure of a CdSe TFT and trapped electrons at the grain boundaries; (b) the energy band diagram of the CdSe showing the grain boundary barrier height $E_B$ caused by electron trapping at the grain boundaries.

\[ E_F \]
CHAPTER 2. RADIATION HARDNESS OF TFTS

2.4 RESULTS AND DISCUSSION

2.4.1 Threshold voltage

Radiation caused negative threshold voltage shift for all the TFTs used in the experiments. In this section, the $\Delta V_T$ will be examined as a function of: (1) TFT channel geometry, i.e. $W/L$ ratio; (2) gate bias voltage during irradiation $V_{GB}$; (3) radiation dose $D$; and (4) recovery of $V_T$, i.e. $\Delta V_T$ as a function of time, at room temperature.

2.4.1.1 TFT geometry dependence

Figure 2.8 shows $\Delta V_T$ induced by $D = 50$ Gy as a function of TFT channel length $L$. The experimental errors were estimated from the repeatability of $V_T$ measurement. It is known that the characteristic curves of CdSe TFTs shift by a small amount between measurements due to temporary charge trapping in the gate insulator from the gate bias voltage applied during the measurement. Shown in Figure 2.8, the repeatability errors in $\Delta V_T$ were small compared to the variation of $\Delta V_T$ between TFTs. The inter-TFT variability is probably attributable to the non-uniformity in gate insulator thickness and trapping properties across an active matrix panel on which the TFTs were made. From Figure 2.8, it is concluded that $\Delta V_T$ is independent of $L$ (over a range of $\sim 3$ fold). This is consistent with the theory shown in Eq. 2.4 that $\Delta V_T$ is independent of the TFT channel geometry.

In order to minimize the effect of gate insulator thickness variation on the measurement of $\Delta V_T$, the values of $\Delta V_T$ obtained from TFTs with different $L$ under the same irradiation condition were averaged to produce the subsequent results of $\Delta V_T$ dependence on gate bias voltage and radiation dose.
Figure 2.8: The threshold voltage shift $\Delta V_T$ as a function of channel length $L$ at radiation dose of 50 Gy. The solid lines are the mean $\Delta V_T$ for TFTs with different $L$. The gate bias voltage during irradiation, $V_{GB}$ were $\pm 20$ V.
2.4.1.2 Bias voltage dependence

The $\Delta V_T$ is plotted as a function of $V_{GB}$ in Figure 2.9. The minimum $|\Delta V_T|$ is in the region of $V_{GB}$ between -5 and 0 V, which corresponds to a zero electric field $E_{ox}$ in the gate insulator. The small deviation of minimum $E_{ox}$ from $V_{GB} = 0$ V is ascribed to charge trapping in the gate insulator during manufacturing of the TFTs, which results in a non-zero flat-band voltage. $|\Delta V_T|$ increases with $|V_{GB}|$ (i.e. $E_{ox}$) regardless of the polarity of $V_{GB}$, which is to be expected from the increase in the number of holes released at higher $E_{ox}$ due to decrease in $W_\pm$. $\Delta V_T$ is greater for positive $V_{GB}$ (when the holes were mostly trapped near the CdSe interface) than that for negative $V_{GB}$ (when the trapped holes were near the gate interface). The fact that $\Delta V_T$ is non-zero for $V_{GB} < 0$ indicates bulk trapping in the gate insulator (since gate interface trapping only will result in $x_{tr} = 0$ and hence $\Delta V_T = 0$).

According to Eq. 2.4, the amplitude of $\Delta V_T$ at a given radiation dose level strongly depends on the following three factors: (1) $W_\pm$ of SiO$_2$; (2) The centroid of the trapped holes $x_{tr}$; and (3) the fraction of hole trapping $f_T$. The measurement of $\Delta V_T$ alone is insufficient for unique determination of all three parameters. While values for $W_\pm$ and $x_{tr}$ could be determined independently through photocurrent measurement and threshold voltage shift measurement with a fraction of the gate insulator etched away, respectively [71], these experimental studies are beyond the scope of this thesis. However if reasonable assumptions were made for $W_\pm$ and $x_{tr}$, an estimate of $f_T$ can be obtained. For example, at $V_{GB} = \pm 20$ V, where $E_{ox} = 0.5$ MV/cm, the value for $\Delta V_T$ is $-4.0 \pm 0.1$ V at $V_{GB} = 20$ V, and $-1.5 \pm 0.1$ V at $V_{GB} = -20$ V. The ratio of $\Delta V_T$ between positive and negative $V_{GB}$, which is $\sim 2.7$, is not far from that expected from uniform hole trap density in the gate insulator (i.e. a ratio of 2 due to $x_{tr} = \frac{1}{3}d_{ox}$ and $\frac{2}{3}d_{ox}$), indicating that hole trapping in the bulk is dominant. The values for $x_{tr}$ can be determined as $\frac{1}{3}d_{ox}$ and $\frac{2}{3}d_{ox}$ for $V_{GB} = -20$ V and 20
Figure 2.9: $\Delta V_T$ as a function of $V_{GB}$ at a radiation dose of 50 Gy.

$V$, respectively, indicating that there are also interface traps. If $W_{\pm} = 23$ eV is assumed for $E_{ox} = 0.5$ MV/cm, as found in c-Si MOSFET gate insulators, the value for $f_T$ can be calculated as 0.25 using Eq. 2.4.

2.4.1.3 Dose dependence

Figure 2.10 shows the dose dependence of $\Delta V_T$ at $V_{GB} = \pm 10$ V measured up to 1000 Gy. The change in $\Delta V_T$ is approximately linear with radiation dose for $D \leq 200$ Gy, then the slope of the curves start to decrease. The mechanism for the sub-linear increase of $\Delta V_T$ at high dose can be attributed to decrease of $E_{ox}$ in the bulk of the gate insulator as a result of hole trapping. For example, if there is a triangular distribution for the holes trapped in the gate insulator, as shown
in Figure 2.4, the magnitude of $E_{ox}$ will follow the second power of the distance $x$. Therefore $E_{ox}$ is much stronger near the interface than in the bulk, which resulted in a lower charge generation in the bulk and also an increase in the recombination of radiation generated electrons with the trapped holes at the interface. The saturation of $\Delta V_T$ at high radiation doses was also observed for c-Si MOSFETs [90] and a-Si TFTs [74].

![Graph showing $\Delta V_T$ as a function of radiation dose at $V_{GB} = \pm 10$ V.](image)

Figure 2.10: $\Delta V_T$ as a function radiation dose at $V_{GB} = \pm 10$ V.

2.4.1.4 Recovery of $V_T$

The time course of the change of $\Delta V_T$ at room temperature after irradiation was monitored over a month under a gate bias voltage of zero, and the results are shown in Figure 2.11. For TFTs with the largest $|\Delta V_T|$ ($V_{GB} = 20$ V), $|\Delta V_T|$ reduced by $\sim 20\%$ after 31 days. This could be due
to electron tunneling from the CdSe to the insulator which neutralized the holes trapped in the SiO₂ very near to the CdSe interface [91,92]. The fact that the recovery is only partial suggests that the majority of the trapped holes are located too far from the semiconductor interface to be neutralized by electron tunneling from the semiconductor. This is consistent with the finding that hole trapping in the bulk of the gate insulator is significant.

Figure 2.11: The room temperature recovery of $V_T$ after irradiation with $D = 200$ Gy and $V_{GB} = \pm 20$ V and 0 V. $V_{GB} = 0$ during recovery.

For $V_{GB} = 0$ V and -20 V, there was no noticeable recovery. This is consistent with the model because the majority of the trapped holes are located in the bulk or close to the gate interface, and therefore tunneling from the CdSe will not be relevant. Tunneling from the gate, although possible, will have no effect on $\Delta V_T$ since the holes trapped very near to the gate (i.e. $x = 0$) do not contribute to $\Delta V_T$, as shown in Eq. 2.1.
CHAPTER 2. RADIATION HARDNESS OF TFTS

Although fast and complete recovery of $V_T$ at elevated temperature (e.g. $> 100 \, ^\circ\text{C}$) has been observed in MOSFETs [92], this approach cannot be applied to CdSe TFTs when used with a-Se in x-ray imaging applications. This is because an a-Se layer has to be kept below 50 $^\circ\text{C}$ in order to avoid recrystallization [93]. However, other recovery methods such as exposure to UV light (e.g. 5 eV light which is below the bandgap of SiO$_2$) through a very thin gate electrode could possibly be applied to CdSe TFTs. The UV light could cause electron injection from either the gate electrode or the semiconductor (depending on $E_{ox}$ and the locations of the trapped holes) and neutralize the trapped holes [94]. However at present the $\Delta V_T$ resulting from irradiation, especially under $V_{GB} \leq 0$, is regarded as essentially permanent.

2.4.2 Subthreshold swing

An increase in subthreshold swing $S$, i.e. $\Delta S > 0 \, \text{V}$, was observed after irradiation. The results were analyzed as a function of radiation dose and bias condition. Room temperature annealing of the subthreshold swing was also monitored.

2.4.2.1 Change in subthreshold swing

Figure 2.12 shows the experimental $\Delta S$ values as a function of radiation dose $D$ at $V_{GB} = -10 \, \text{V}$. $\Delta S$ increases linearly at low dose and starts to saturate at 200 Gy. At highest $D$ (1000 Gy), $\Delta S$ is $\sim 0.25 \, \text{V/decade}$. Figure 2.13 shows $\Delta S$ as a function of $D$ for different $V_{GB}$. The curves for $V_{GB} \geq 0 \, \text{V}$ are shown in Figure 2.13 (a), and those for $V_{GB} \leq 0 \, \text{V}$ are shown in Figure 2.13 (b). In general, the following $V_{GB}$ dependence was observed: (i) $V_{GB} = 0 \, \text{V}$ resulted in the smallest $\Delta S$; (ii) For each $V_{GB}$ polarity, higher $E_{ox}$ generates higher $\Delta S$. 
Figure 2.12: The change in subthreshold swing, $\Delta S$, as a function of radiation dose. $V_{GE}$ was -10 V during irradiation.
Figure 2.13: The change in subthreshold swing, $\Delta S$, as a function of radiation dose with: (a) $V_{GB} \geq 0$ during irradiation; (b) $V_{GB} \leq 0$ during irradiation.
2.4.2.2 Subthreshold swing recovery

The subthreshold swing of the TFTs were monitored for a month after the irradiation. Figure 2.14 shows the results for two TFTs with different $V_{GB}$ ($L = 40 \text{ \mu m}$). For $V_{GB} = -20 \text{ V}$, $S$ recovers quickly to its original value ($\Delta S = 0 \text{ V}$) within the uncertainty possibly due to TFT instability. Similar recovery was observed for $V_{GB} = 0 \text{ V}$. However for $V_{GB} = 20 \text{ V}$, no significant change in $S$ with time is seen (i.e. there is no recovery of $S$). Therefore for $V_{GB} \leq 0 \text{ V}$, there is no permanent effect of radiation on $S$. But for $V_{GB} > 0 \text{ V}$, $\Delta S$ is permanent.

![Figure 2.14: The subthreshold swing plotted as a function of time after irradiation. $V_{GB}$ was $\pm 20 \text{ V}$ and recovery was measured at room temperature.](image)

$\Delta S > 0$ indicates that radiation creates localized states in the band gap of CdSe (i.e. dangling bonds). This was also observed in c-Si MOSFETs and a-Si TFTs. The mechanism of this effect has been studied for c-Si MOSFETs, and it was believed to be related to radiation generated
holes in the gate insulator which later formed localized states in the semiconductor at the gate insulator - semiconductor interface [95]. A well-supported theory proposes that radiation induced holes react with hydrogen in the gate insulator and create hydrogen ions (H⁺). These H⁺ ions are driven to the semiconductor interface by $E_{ox}$ (for positive $V_{GB}$), where they pick up electrons from the semiconductor and become highly reactive H⁰, which then react with the Si-H bonds in the semiconductor (trap precursors) to produce silicon dangling bonds (the interface trap) and H₂, and the effect is permanent [77, 96, 97]. The relationship between the creation of interface traps and the radiation-generated holes in the gate insulator is responsible for the $E_{ox}$ dependence of $\Delta S$. Saturation of $\Delta S$ at high radiation levels was also observed in c-Si MOSFETs which was attributed to a fixed density of trap precursors at the gate insulator - semiconductor interface [98].

In the CdSe TFTs used in the experiment, the SiO₂ gate insulator is formed by PECVD using the chemical reaction between silane (SiH₄) and nitrous oxide (N₂O):

$$\text{SiH}_4 + 4 \text{N}_2\text{O} \rightarrow \text{SiO}_2 + 4\text{N}_2 + 2\text{H}_2\text{O} \quad (2.8)$$

and this process is known to produce SiO₂ films that contains hydrogen [99]. The hydrogen is bonded to silicon as Si - H and to oxygen as Si - OH and H₂O. Also, hydrogen bonds in CdSe (possibly formed by hydrogen diffusion from the gate insulators to the CdSe layer during TFT manufacture) are believed to reduce the density of localized states in the semiconductor thin films. Therefore the H⁺ ion based theory proposed for MOSFETs may be applicable to CdSe TFTs. This theory can explain the experimental results of increase of $\Delta S$ with $E_{ox}$ under positive $V_{GB}$, and the saturation of $\Delta S$ at high radiation levels. Under negative $V_{GB}$, there is no increase in $S$ since the hydrogen ions cannot be driven to the semiconductor interface to react with the trap precursors. This is consistent with the observation of no long term effect of $S$ under negative $V_{GB}$.
2.4.3 Field effect mobility

Shown in Figure 2.15 are the $I_D - V_G$ curves for a typical TFT before and after 200 Gy of irradiation. The slope $g_m$ changed from $(2.45 \pm 0.03) \times 10^{-7}$ to $(2.50 \pm 0.03) \times 10^{-7}$ A / V after irradiation. The difference is within the experimental error, and therefore there is no change of $\mu_{FE}$ as a result of radiation. This indicates that irradiation does not cause significant increase in the conduction band tail states, nor any increase in electron trap density in the gate insulator very near the CdSe layer.

![Figure 2.15: The transfer characteristic curve (i.e. channel current $I_D$ as a function of gate voltage $V_G$) plotted on linear scale for a TFT before and after irradiation with 200 Gy. $V_{GB}$ was -5 V. The measured slope $g_m$ for calculation of the field effect mobility is indicated beside each curve.](image)
2.4.4 Leakage current

Figure 2.16 shows the change of TFT leakage current $I_{DL}$ (measured at $V_G = -5$ V) as a function of radiation dose for TFTs irradiated with $V_{GB} = \pm 10$ V and -5 V. It shows that with $V_{GB} = -5$ V and -10 V, there is no significant change of $I_{DL}$ with irradiation. However with $V_{GB} = 10$ V, $I_{DL}$ starts to increase significantly after a radiation dose of 100 Gy. This result is consistent with the finding of a larger threshold voltage shift $\Delta V_T$ when $V_{GB} > 0$. The increase of $I_{DL}$ at $V_{GB} = 10$ V is because the negative shift of the $I_D-V_G$ curve brings $V_G = -5$ V (when the TFT is usually turned off) into the subthreshold region. But under $V_{GB} < 0$, when $\Delta V_T$ is small, $I_{DL}$ is unaffected by irradiation.

![Graph showing the leakage current of TFTs before and after irradiation with 50 Gy. The values for $V_{GB}$ during irradiation were $\pm 10$ V and -5 V.](image)

Figure 2.16: The leakage current of the TFTs before and after irradiation with 50 Gy. The values for $V_{GB}$ during irradiation were $\pm 10$ V and -5 V.
2.5 IMPLICATIONS FOR X-RAY IMAGING

The primary focus of this section is to examine the implications of the radiation effects for diagnostic x-ray imaging. Implications for portal imaging will also be briefly discussed. In radiographic imaging applications, the CdSe TFTs are turned off ($V_{GB} = -5 \text{ V}$) during radiation in order for charge created in the $a$-Se layer to integrate on the pixel electrodes. In fluoroscopy, the active matrix is operated in real-time (30 frames / second) and each TFT is turned on for less than 10 $\mu$s once in every $1/30$ second [84]. Therefore the TFTs remain in the off state during most of the imaging procedures. Thus the effects of radiation for $V_{GB} < 0 \text{ V}$ are the most important.

The experiments with CdSe TFTs have shown that under $V_{GB} < 0 \text{ V}$, radiation causes a negative shift of the threshold voltage $V_T$ and an increase of its subthreshold swing. Since the TFTs are used as switches in x-ray imaging applications, the effects of radiation are evaluated according to whether the TFTs can function normally with gate voltages of $V_{G(\text{off})} = -5 \text{ V}$ and $V_{G(\text{on})} = 10 \text{ V}$ after irradiation.

2.5.1 Threshold voltage shift and leakage current

The experimental results showed that radiation causes negative $V_T$ shift, thus the on characteristics of the TFTs will not be degraded. However it is important to ensure that the leakage current $I_{DL}$ measured at $V_{G(\text{off})} = -5 \text{ V}$ will not increase as a result of the negative $V_T$ shift, i.e. $V_{G(\text{off})} = -5 \text{ V}$ will not move into the subthreshold region of the characteristic curves. In order to satisfy this condition, as shown with the characteristic curve of a typical CdSe TFT in Figure 2.6, $|\Delta V_T|$ cannot be greater than 5 V. The measurement of leakage current $I_{DL}$ as a function as radiation dose, as shown in Figure 2.16, shows that this will not happen under a gate bias voltage of $V_{GB} = -5 \text{ V}$ for dose levels up to 200 Gy. Therefore the TFTs can retain normal operation at operational
voltages of $V_{G(\text{eff})} = -5 \text{ V}$ and $V_{G(\text{on})} = 10 \text{ V}$ for the expected detector life-time in diagnostic imaging applications.

If the present CdSe TFTs are applied to portal imaging applications where the expected life-time dose is $10^5 \text{ Gy}$ [74], the requirement of $|\Delta V_T| \leq 5 \text{ V}$ will not be satisfied. Shown in Figure 2.10 when $V_{GB} = -10 \text{ V}$, $\Delta V_T$ is -9.18 V at $D = 1000 \text{ Gy}$. It can be expected that the value for $|\Delta V_T|$ under the radiation dose used in portal imaging applications will cause a significant increase of $I_{DL}$. Therefore for this application, $|\Delta V_T|$ of CdSe TFTs would have to be decreased. Since $\Delta V_T$ depends only on the properties of the gate insulator, this could be accomplished using combinations of the following methods: (1) reducing the gate insulator thickness; (2) decreasing $f_T$ by radiation hardening techniques; (3) choosing a different gate insulator material, such as Si$_3$N$_4$, which has higher dielectric constant and much larger $W_\pm$ than SiO$_2$ [100,101]. A recent study on radiation hardness of a-Si TFTs showed that devices produced with a 200 nm thick Si$_3$N$_4$ gate insulator were adequate for portal imaging [74]. This approach could possibly be applied to CdSe TFTs. The disadvantages, however, of such a thin gate insulator layer are a reduction in high voltage capability and an increase in data line capacitance which increases the electronic noise of the detector. This analysis shows that radiation hardness is a system parameter which cannot be considered in isolation from other important imaging parameters.

2.5.2 Subthreshold swing

The increase of subthreshold swing by irradiation, i.e. the shallower subthreshold region, will result in a decrease in $I_D$ when the TFT is turned on with $V_G = 10 \text{ V}$. However, since the negative $V_T$ shift results in an increase in $I_D$ for the on state, the effect of $\Delta S$ is compensated. Shown in Figure 2.13 (b), $\Delta S$ is $\sim 0.05 \text{ V}$ at $D = 50 \text{ Gy}$ and $V_{GB} = -5 \text{ V}$. Taking the change of $I_D$ in the subthreshold region to be 4 orders of magnitude, the effect of $\Delta S$ is like a shift of the on region of
the $I_D - V_G$ curve to the right by 0.20 V (0.05 V x 4). This is to be compared with the $\Delta V_T$ of -0.3 V. Therefore the *on* current of the TFT should not be degraded by radiation. Furthermore, since a room temperature recovery of $S$ has been observed for $V_{GB} \leq 0$ V, there is no permanent effect of radiation on $S$ in x-ray imaging applications.

2.6 CONCLUSIONS

The radiation hardness of CdSe TFTs was studied for dose level of up to 1000 Gy (20 times the estimated lifetime dose). The effects of radiation noted were a shift of threshold voltage and an increase in subthreshold swing. For the radiation levels and TFT gate bias voltages used in diagnostic x-ray imaging applications, the dominant effect is the shift of threshold voltage of up to $\sim 1$ V. However, this amplitude is small enough to ensure correct operation of the TFTs under normal operating voltages. Therefore, the radiation hardness of CdSe TFTs is adequate for the application in diagnostic x-ray imaging. Methods for improving the radiation hardness of CdSe TFTs for applications requiring higher radiation dose includes using thinner gate insulator and improving the quality of SiO$_2$ in order to decrease the fraction of holes trapped in the gate insulator, as well as using other gate insulator materials (such as Si$_3$N$_4$) which generate fewer charge carriers upon absorption of radiation.
Chapter 3

Detectors with High Voltage Protection

3.1 INTRODUCTION

The concept of making a direct, flat-panel detector using the technology of a large area active matrix, which consists of a two dimensional array of thin-film transistors (TFTs) is being investigated [67,89]. A schematic diagram of the structure of our flat-panel detector is shown in Figure 1.9. The detector employs a layer of x-ray sensitive photoconductor such as amorphous selenium (a-Se) to directly convert the incident x-rays to charge. A positive high voltage (several thousand volts) is applied to the top surface of a-Se in order to establish an electric field across the a-Se layer. Holes created by x-rays are driven by the electric field to the bottom surface of a-Se, where they are collected and electronically read out by the active matrix. Hence the detector is also referred to as the self-scanned a-Se detector. Each pixel of the active matrix consists of a pixel

\footnote{This chapter has been submitted as a paper to Med. Phys., entitled “Digital radiology using active matrix readout of amorphous selenium: Detectors with high voltage protection”.

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electrode for charge collection, a capacitor $C_P$ for charge storage, and a TFT for image readout. During readout, the TFTs are turned on one row at a time to transfer the image charge from the pixels to the external charge amplifiers. The readout rate of the self-scanned $\alpha$-Se detector can be in real-time, i.e. 30 frames per second, and therefore it can be applied in fluoroscopy as well as radiography.

Under normal detector operation, the detector is scanned continuously except during radiographic x-ray exposure. The potential on the pixel electrode $V_P$ (which equals the drain-source voltage of the TFT $V_{DS}$) is usually less than 10 V for the x-ray exposure used in diagnostic imaging applications [84], which is safe for the TFTs. However a combination of the following two events may cause high voltage to build up on a pixel: (1) a suspended detector scan (during radiographic exposure or due to malfunctioning of the scanning electronics); and (2) an accidental over-exposure (up to the maximum output allowed by the x-ray generator). Under this fault condition, the pixel potential $V_P$ (i.e. $V_{DS}$) can keep rising as a function of x-ray exposure and approaches the potential of the high voltage bias on the top surface of $\alpha$-Se, thus undoubtedly damaging the detector beyond repair. Although this fault condition is unlikely, it is perhaps to be expected occasionally during the lifetime of a detector and thus a high voltage protection mechanism is necessary.

Previously, two methods have been proposed for protecting the active matrix from high voltage damage. The first is to use an alternative detector configuration [60], where a layer of insulator (thickness comparable to $\alpha$-Se) is added between the $\alpha$-Se layer and the top bias electrode. Charge generated by x-rays is trapped at the interface between the insulator and the $\alpha$-Se layer, which reduces the electric field in the $\alpha$-Se, thus providing a saturation of pixel potential at high x-ray exposure and ensuring that high voltage damage cannot occur [102]. However, this method requires a procedure to remove the trapped charge between two subsequent x-ray exposures, which
means it can not be applied to real-time x-ray imaging and is restricted to radiography. The second method is applicable to real-time detectors by adding a high voltage protection TFT at each pixel of the active matrix [103], as shown in the pixel circuit diagram in Figure 3.1 (a). The gate of the protection TFT is connected to the pixel electrode and its threshold voltage, i.e. the gate voltage required to turn on the TFT, is selected to be a safe potential for the TFTs. When \( V_P \) reaches the threshold voltage, the protection TFT turns on and drain excess charge from the pixel electrode to ground. However adding another TFT at each pixel will inevitably result in a decrease in production yield and possibly a significant decrease in pixel fill factor (the fraction of the pixel area used for image charge collection). It may also require extra processing steps in manufacturing. Therefore it is desirable to minimize the complexity of the active matrix design.

In this chapter, a novel approach to high voltage protection applicable to a real-time, self-scanned a-Se detector, which requires only one TFT at each pixel is described. As shown in Figure 3.1 (b), the TFT employs a dual-gate structure, one is the scanning control gate (subsequently referred to as the control gate) and the other is the high voltage protection gate (subsequently referred to as the protection gate) which is connected to the pixel electrode. Under normal x-ray exposure conditions, the control gate performs its function as in an ordinary single gate TFT, and the protection gate has little effect. However under fault x-ray conditions, the protection gate overrides the control gate and automatically turns on the TFT before \( V_P \) (i.e. \( V_{DS} \)) reaches a damaging value. The excess charge then drains through the TFT to the external charge amplifiers so limiting the pixel potential rise, thus avoiding high voltage damage. The saturation of \( V_P \) for a detector with high voltage protection will undoubtedly limit its dynamic range. However with appropriate design of the dual-gate TFT geometry, the detector can provide sufficient dynamic range for all diagnostic x-ray imaging applications.
Figure 3.1: Circuit diagram of a single pixel of a self-scanned a-Se detector with high voltage protection: (a) In addition to the scanning control TFT, there is a high voltage protection TFT which drains excess charge to ground; (b) The approach of using a single TFT with dual-gate structure: the scanning control gate and the high voltage protection gate.
In Section 3.2 of this chapter, the mechanisms for high voltage damage and the high voltage performance of TFTs are discussed, and the detector configuration for high voltage protection using a dual-gate TFT structure is described. In Section 3.3, the characteristics of dual-gate TFTs with different design parameters are predicted using a model, and the results are experimentally verified with cadmium selenide (CdSe) dual-gate TFT samples. In Section 3.4, the pixel x-ray response of an imaging detector with high voltage protection is predicted for both fault and normal x-ray exposure conditions, and the relationship between the dynamic range and the high voltage protection characteristics of the detector are investigated. Finally in Section 3.5, strategies for choosing dual-gate TFT parameters for protection in different x-ray imaging applications, and methods for practical implementation of dual-gate TFT structures are discussed.

3.2 BACKGROUND

3.2.1 Principles of TFT operation

As shown in Figure 3.2, a TFT has three electrical connections: the gate is for the control of the state of the TFT, the drain and source are symmetrical and connected to the ends of the semiconductor channel. The principle of operation of TFTs is similar to that of n type metal-oxide-semiconductor field effect transistors (MOSFETs). When a negative potential (e.g. -5 V) is applied to the gate, the semiconductor layer behaves as an insulator which results in a source-drain leakage current of \(<10^{-13}\)A and the TFT is in the off state. However when a positive potential is applied to the gate (e.g. 10 V), free electrons are generated in the semiconductor layer and form a channel. With a source-drain voltage applied, current of the order of \(10^{-6}\)A can flow through the channel and the TFT is in the on state.
3.2.2 Mechanisms of high voltage damage in TFTs

When a high voltage is applied to the drain of a TFT, it could damage the TFT by one of two mechanisms: (1) dielectric breakdown of the gate insulator layer; and (2) collapse, i.e. decrease of the TFT on current. Dielectric breakdown depends on both the thickness and the dielectric strength of the gate insulator. For example, in some a-Si:H TFTs where the gate insulator is a 200 nm thick layer of Si₃N₄, damage can occur at drain-source voltage $V_{DS} > 20$ V. For the CdSe TFTs used in our investigation, the gate insulator is usually a 400 nm thick layer of SiO₂ which has a breakdown field of $\sim 4$ MV/cm [70]. Therefore breakdown of the gate insulator will not occur provided $V_{DS}$ is less than 150 V. Collapse of TFT on current is usually the damage mechanism for TFTs with high dielectric breakdown voltage. The collapse phenomenon of CdSe TFTs upon the application of a high drain voltage has been observed by many investigators [69, 104, 105]. Shown in Figure 3.3 (a) are the normal output characteristic ($I_D - V_{DS}$) curves of a CdSe TFT when it is turned on with positive gate voltage $V_G$. However after the TFT is subjected to $V_{DS} > 100$ V, the low voltage region of the output characteristics is distorted, or collapsed, as shown in
Figure 3.3 (b). This suppression of TFT channel current at low $V_{DS}$ has been attributed to the trapping of hot electrons in the gate insulator near the drain [106]. These trapped electrons give rise to a depletion region near the drain which produces a potential barrier in the conduction band. Although collapsed TFTs show partial recovery over the time course of several days [107], since in x-ray imaging applications the time between x-ray exposures cannot be longer than a few seconds, the collapse of TFT channel current caused by high voltage can be regarded as irreversible for all practical purposes.

3.2.3 High voltage characteristics of TFTs

CdSe TFTs are known for their capabilities of withstanding high voltages (e.g. $V_{DS} = 100$ V) [69, 104]. Figure 3.4 is the $I_D - V_{DS}$ curve of an ordinary CdSe TFT (channel width and length of 36 and 50 $\mu$m, respectively) showing normal characteristics with $V_{DS}$ up to 100 V. This inherent high voltage capability provides flexibility for the design of TFTs with sufficient high voltage protection for practical x-ray imaging detectors. Although ordinary hydrogenated amorphous silicon ($a$-Si:H) TFTs usually operate with $V_{DS} \leq 20$ V [108], they can probably be made to stand much higher voltages (e.g. $V_{DS} > 50$ V) with improvements to the dielectric breakdown properties of the gate insulators.

High voltage TFTs have also been made with a special drain electrode offset structure. This approach has been applied to both poly-Si [109] and $a$-Si:H [110] TFT technology. In these high voltage TFTs, as shown in Figure 3.5 (a), the gate covers only a small fraction of the channel length, and the drain electrode is offset from the gate and no longer overlaps it. The conductivity of the gated portion of the channel is controlled by the gate as in the ordinary TFTs, and the offset region is governed by the resistance of the intrinsic semiconductor thin films. The purpose of this offset region is to prevent excess field from breaking down the gate insulator ($Si_3N_4$). The channel
Figure 3.3: High voltage damage to a CdSe TFT: (a) The normal output characteristic curves (channel current $I_D$ as a function of drain-source voltage $V_{DS}$) of a CdSe TFT measured with gate voltage $V_G$ ranging from 10 to 20 V; (b) The collapsed output characteristic ($I_D - V_{DS}$) curves of the same CdSe TFT after a high $V_{DS} > 100$ V was applied.
current at low drain voltage is very small because of the high resistivity of intrinsic semiconductor films. At high drain voltages, the channel current is established by the strong electric field at the edge of the gate electrode (near the offset region) pulling the accumulated charge from the gated region into the offset region, then the charge travels to the drain electrode in a space-charge-limited fashion. The output characteristics of such high voltage TFTs are illustrated in Figure 3.5 (b). Because of the high channel resistance at low drain voltage, which is the operational voltage range for diagnostic x-ray imaging, these high voltage TFTs with offset structure are not suitable for x-ray imaging applications.
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Figure 3.5: (a) The cross section of a high voltage TFT with offset structure (note that the thicknesses of layers are not to scale); (b) The output characteristic ($I_D - V_{DS}$) curve of a high voltage TFT with offset structure.
3.2.4 Dual-gate TFTs for high voltage protection

The most common TFT structure used in active matrix technology has a single gate on the bottom. A dual-gate TFT structure, as shown in Figure 3.6 (a), has an additional top gate. Dual-gate CdSe TFTs with the top and bottom gate connected together have been previously used in active matrix displays [111]. Dual-gate a-Si:H TFTs have also been used in fundamental studies of device parameters such as: semiconductor-gate insulator interface properties and the effects of the drain and source electrode contacts [112]. The top and bottom gates affect the TFT channel conductivity in a similar fashion, thus the dual-gate TFTs have higher on current than the single gate TFTs. A dual-gate TFT structure is also better shielded from the effects of external electric field, and it has been applied to our prototype self-scanned a-Se detector in order to shield the otherwise exposed TFT channel from the electric field in a-Se [89]. Shown in Figure 3.7 is the transfer characteristic curve of a TFT on this imaging detector (with the top and bottom gates connected together). Its on current is twice as high as its single gate counterpart [113].

Shown in Figure 3.6 (b) is the novel concept of high voltage protection using dual-gate TFT structure, in which the top gate is connected to the pixel electrode for high voltage protection, and the bottom gate is for scanning control. The relationship between dual-gate TFT design and its ability to protect against high voltage damage will be investigated theoretically in the following section.

3.3 CHARACTERISTICS OF DUAL-GATE TFTS

The effect of gate voltage on channel conductivity is proportional to the gate insulator capacitance, therefore the characteristics of dual-gate TFTs for high voltage protection is dependent on the design parameters such as: top and bottom gate insulator material and thicknesses. In this section,
Figure 3.6: TFTs with high voltage protection: (a) The cross-section of a TFT with dual-gate structure; (b) The cross section of a single pixel of a real-time, self-scanned a-Se detector with dual-gate TFT structure for high voltage protection. (note that the thicknesses of layers are not to scale)
Figure 3.7: The transfer characteristic ($I_D - V_G$) curve of a dual-gate TFT on our x-ray imaging detector with the top and bottom gates connected together. The drain-source voltage $V_{DS} = 1$ V.
a model for dual-gate TFTs will be developed which can predict the characteristics of TFTs with different design parameters. This is achieved through the following steps: (1) development of the dual-gate TFT model; (2) determination of semiconductor characteristic constants required by the model; (3) modeling characteristic curves of dual-gate TFTs with different design parameters; and (4) experimental verification of the model using TFT samples with a given set of design parameters.

3.3.1 Dual-gate TFT model

The gates of a TFT affect the semiconductor channel conductivity by the field effect. Figure 3.8 shows the energy band diagram of a dual-gate, n-type field effect transistor with a thick semiconductor layer. Under zero gate bias (\(V_G = 0\)) and no fixed charge in the gate insulators, as shown in Figure 3.8, the semiconductor is in the flat-band condition, and the Fermi energy \(E_F\) for the n-type semiconductor is closer to the conduction band edge \(E_C\) than the valence band edge \(E_V\).

The density of free electrons \(n_0\) under flat-band condition for non-degenerate semiconductor can be approximated by Boltzmann's distribution [79]:

\[
n_0 = N_C \exp\left(-\frac{E_C - E_F}{kT}\right), \tag{3.1}
\]

where \(N_C\) is the density of states in the conduction band, \(k\) is Boltzmann's constant, and \(T\) is the absolute temperature. When positive gate voltages are applied (\(V_G > 0\)), as shown in Figure 3.8 (b), the conduction band of the semiconductor near the gate insulator interface bends from its bulk value \(E_C\) towards the Fermi energy \(E_F\) by the amount of energy \(U\), and more free electrons are generated in the semiconductor which eventually form a conductive channel [114]. The thickness of the conductive region at each gate insulator-semiconductor interface is comparable to the characteristic Debye length of the semiconductor which is determined by the carrier concentration \(n_0\) [79]. The Debye length for CdSe with \(n_0 = 10^{14} \, \text{cm}^{-3}\) is \(\sim 0.5 \, \mu\text{m}\) [115]. However when \(V_G < 0\), as shown in
Figure 3.8 (c), the conduction band bends away from $E_F$ and the number of free electrons decreases thus permitting the TFT to be turned off [114].

The characteristics of dual-gate CdSe TFTs have been modelled by Chen et al. [115, 116], who obtained the energy band bending $U$ in the semiconductor at given top and bottom gate voltages by solving Poisson's equation. $U$ was determined both across the thickness of the CdSe layer and along the channel. For our dual-gate TFTs, the CdSe thickness is only 50 nm, which is approximately an order of magnitude smaller than the Debye length. In this case the dual-gate TFT model can be simplified to a constant $U$ across the thickness of the semiconductor layer (i.e. "band shifting"). The TFT channel was thus divided only along the channel length into $M$ segments, as shown in Figure 3.9. The free charge carrier (electron) density $n_m$ at segment $m$ was determined by $U_m$ at that segment.

When bias voltages are applied to the gates of the TFT, excess charge carriers $Q$ are capacitively generated in the semiconductor channel. For each small segment $m$ in Figure 3.9, the potential of the semiconductor channel can be represented by a constant $V_m$. Thus the excess number of carriers in each segment, $Q_m$, can be described by:

$$Q_m = \frac{W L}{M} [C_{TG}(V_{TG} - V_{TF} - V_m) + C_{BG}(V_{BG} - V_{BF} - V_m)].$$  \hspace{1cm} (3.2)$$

where $W$ and $L$ are the channel width and length of the TFT, respectively; $C_{TG}$ and $C_{BG}$ are the gate insulator capacitance per unit area for the top and bottom gates, respectively; $V_{TG}$ and $V_{BG}$ are the bias voltages on the top and bottom gates, respectively; $V_{TF}$ and $V_{BF}$ are the flat-band voltages, i.e. the gate voltage required to have no band bending (or shifting), for the top and bottom gates, respectively. A non-zero flat-band voltage is due to charge trapped in the gate insulator during the manufacture of the TFTs.

The convention used for $U_m$ is that when free electrons are generated ($Q_m > 0$) upon the
Figure 3.8: The band diagram of a semiconductor under the field effect: (a) flat-band condition under zero gate bias ($V_G = 0$) and no fixed charge in the gate insulators; (b) band bending under positive gate bias ($V_G > 0$); (b) band bending under negative gate bias ($V_G < 0$).
application of gate voltages, the band shift $U_m$ is denoted positive. From Boltzmann’s distribution in Eq. 3.1, the free electron density $n_m$ after a band shifting of $U_m$ can be determined using the free electron density $n_0$ at the flat-band condition from [79]:

$$n_m = n_0 \exp\left(\frac{U_m}{kT}\right).$$  \hspace{1cm} (3.3)

In order to determine the value of $U_m$ for a given set of gate voltages, the localized states in the semiconductor band gap have to be considered. When $Q_m$ are induced in the semiconductor by the gate voltages, a fraction of $Q_m$ will fill the traps in the semiconductor band gap (localized states) and are not mobile. Therefore $Q_m$ is given by the sum of excess free electrons $(n_m - n_0)$ and the filled localized states [116]:

$$\frac{M Q_m}{W L d_s} = e n_0 [\exp(\frac{U_m}{kT}) - 1] + e N_L U_m,$$  \hspace{1cm} (3.4)

where $d_s$ is the thickness of the semiconductor layer, $e$ is the electron charge, $N_L$ is the density of localized states (traps) in the bandgap of the semiconductor which is assumed independent of energy. Using Eq. 3.4, the band shift $U_m$ and in turn the density of free carriers $n_m$ can be solved.
for a given set of top and bottom gate voltages. Then the channel conductivity $\sigma_m$ of segment $m$ of the semiconductor layer can be calculated as the product of $n_m$ and the electron field effect mobility, $\mu_{FE}$, of the semiconductor [79]:

$$\sigma_m = e \mu_{FE} n_m. \tag{3.5}$$

In polycrystalline semiconductor layers, there are a large number of electron traps located at the grain boundaries separating crystallites. When these traps are filled, they can form a grain boundary barrier, $E_B$, which reduces the number of free carriers that contribute to conduction, and hence the conductivity $\sigma_m$. This can be treated as a decrease in the effective $\mu_{FE}$ and has been investigated by Levinson et al. [88] for polycrystalline CdSe as a function of the grain boundary surface electron trap density $N_t$ and the applied gate voltage. It was shown that $\mu_{FE}$ at a given $V_G$ can be determined from the free carrier concentration $n_m$ and the grain boundary electron trap surface density $N_t$ using:

$$\mu_{FE} = \mu_{FE0} \exp\left(-\frac{e^2 N_t^2}{8\epsilon kT n_m}\right), \tag{3.6}$$

where $\epsilon$ is the permittivity of the semiconductor and $\mu_{FE0}$ is the field effect mobility without the grain boundary barrier. Eq. 3.6 shows that the higher the grain boundary trap density, the lower the $\mu_{FE}$. As gate voltages increases, $n_m$ increases and the exponential term in Eq. 3.6 approaches unity. Therefore at high gate voltages, $\mu_{FE}$ saturates at $\mu_{FE0}$. The $\mu_{FE0}$ for the CdSe TFTs used here was determined from the slope of transfer characteristic curves (channel current $I_D$ as a function of gate voltage $V_G$) in the region of $V_G > 30$ V, and $\mu_{FE0} = 160$ cm$^2$/Vs [113]. Knowing the channel conductivity $\sigma_m$ from Eq. 3.5 and the TFT geometry, the channel current $I_{D_m}$ can be calculated using [79]:

$$I_{D_m} = \sigma_m W d_s \frac{V_{DS_m}}{L/M}. \tag{3.7}$$
Because of the current continuity in the semiconductor channel, $I_{D_m}$ for all the segments are equal.

For a given $I_D$, $V_{DS_m}$ for each segment can be determined using Eq. 3.7. These $V_{DS_m}$ can then be integrated to obtain the total potential drop across the channel, $V_{DS}$.

In order to derive the overall output characteristic curve of a dual-gate CdSe TFT, i.e. $I_D$ as a function of $V_{DS}$, the following steps were followed during the modelling:

1) Specifying the top and bottom gate potentials $V_{TG}$ and $V_{BG}$;

2) Specifying a channel current $I_D$, ensuring that $I_D < I_{max}$ which is the maximum channel current at a given gate bias condition;

3) Specifying the number of segments $M$ for the TFT channel ($M = 500$ was chosen in our simulation);

4) Starting from the source, calculating the band shifting of each segment $U_m$ using Eqs. 3.2 and 3.4, and then determine the conductivity of each segment $\sigma_m$ using Eq. 3.5;

5) Calculating the source-drain voltage of each segment $V_{DS_m}$ using Eq. 3.7, and integrate $V_{DS_m}$ to obtain the overall source-drain voltage $V_{DS}$;

6) If the dual-gate TFT is operating in high voltage protection mode, compare $V_{DS}$ to $V_{TG}$. Repeat steps 2) to 5) until $V_{DS} = V_{TG}$.

### 3.3.2 Experimental determination of $n_0$, $N_L$ and $N_i$

Before predicting the TFT characteristics, the semiconductor parameters $n_0$, $N_L$ and $N_i$ in the TFT model have to be established. These parameters were obtained by fitting the model to experimentally measured characteristic curves of our dual-gate CdSe TFTs, the structure of which is shown in cross section in Figure 3.10. In our present dual-gate TFT process, both the top and bottom gate insulators are SiO$_2$. The source and drain electrodes make contact with the CdSe layer through etched vias in the top insulator, and thus they overlap the CdSe channel by $\sim 5$
\( \mu m \) with the current design rules. Because of this overlap and the gaps (5 \( \mu m \) each) between the top gate and source and drain electrodes, the top gate does not completely cover the TFT channel. The channel \( W \) and \( L \) are 20 and 50 \( \mu m \), respectively. The thicknesses of the top and bottom gate insulators are 330 and 360 nm, which result in a \( C_{TG} \) and \( C_{BG} \) of 10.5 and 9.6 nF cm\(^{-2} \), respectively. The dual-gate TFT samples were fabricated on glass substrates and cut and wire-bonded in DIP packages for measurement. The characteristic curves were measured with a HP4156A semiconductor parameter analyzer.

![Diagram of dual-gate TFT](image)

Figure 3.10: Cross section of the dual-gate TFT used in experimental measurement. The source and drain overlap with the channel \( P_S \) and \( P_D \) are 5 \( \mu m \). The gap between the top gate and the source and drain, \( G_S \) and \( G_D \) are also 5 \( \mu m \). The channel width \( W \) and length \( L \) are 20 and 50 \( \mu m \), respectively.

In order to obtain an experimental characteristic curve of the dual-gate TFTs for data fitting using the theoretical model developed for an idealized dual-gate TFT structure, the effect of the incomplete coverage of the top gate was minimized by measuring the characteristic curve with \( V_{TG} = 0 \) V. Shown in Figure 3.11 is the \( I_D - V_{BG} \) curve of the dual-gate TFT measured with \( V_{TG} = 0 \) V and \( V_{DS} = 0.1 \) V. Because of the very small \( V_{DS} \), the gate generated excess charge \( Q_m \).
and hence $U_m$ and $\sigma_m$ are uniform along the channel, therefore Eq. 3.4 can be simplified to:

$$\frac{C_{BG}(V_{BG} - V_{BF})}{d_s} = e n_0 \left[ \exp\left(\frac{U}{kT}\right) - 1 \right] + e N_L U$$

(3.8)

Figure 3.11: The experimentally measured bottom gate transfer characteristic curve ($I_D - V_{BG}$) of a dual-gate CdSe TFT with top gate voltage $V_{TG} = 0$V and drain-source voltage $V_{DS} = 0.1$ V (open circles), compared to the fitted curve using a dual-gate TFT model (solid line).

During data fitting, the values for $N_L$ and $n_0$ determined by Chen et al. [116], as well as the value for $N_t$ determined by Levinson et al. [88], were used as guidelines for choosing initial values for these three parameters. Then for each set of values for $N_L$, $n_0$ and $N_t$, the associated $I_D - V_G$ curve was calculated using the first five modelling steps summarized in Section 3.3.1. After a number of iterations, the fitted curve in Figure 3.11 shows a good agreement with the experimental measurement. It predicted a bottom gate flat-band voltage value of $V_{BF} = 2.2$ V indicating that electrons were trapped in the gate insulator during TFT manufacture. The predicted semiconductor parameters were: $n_0 = 4.6 \times 10^{14}$ cm$^{-3}$, $N_L = 9.9 \times 10^{17}$ cm$^{-3}$eV$^{-1}$, and $N_t = 6.5 \times 10^{10}$ cm$^{-2}$. 
3.3.3 Prediction of TFT characteristics for high voltage protection

Using the above determined values for $N_L$, $n_0$ and $N_t$, the six modelling steps summarized in Section 3.3.1 were followed to predict the characteristics of a dual-gate CdSe TFT operated in high voltage protection mode (i.e. $V_{TG} = V_{DS}$). Since the values for flat-band voltages depend critically on TFT manufacturing and may vary between different processes, for the purpose of the calculation, $V_{TF} = V_{BF} = 0$ was used.

Shown in Figure 3.12 are the control gate transfer characteristics, i.e. the channel current $I_D$ as a function of the bottom gate voltage $V_{BG}$ at different $V_{DS} = V_{TG}$. The minimum $I_D$ used in the calculation was $10^{-13}$ A in order to reflect the realistic off current of the CdSe TFTs as shown in Figure 3.11. The thicknesses for the top and bottom gate insulators, $d_{ti}$ and $d_{bi}$, are 2 μm and 360 nm, respectively. As shown in Figure 3.12, when $V_{DS} = 10$ V, which corresponds to the maximum pixel potential under normal diagnostic x-ray exposure, the TFT is maintained in the off state with $V_{BG} = -5$ V. However when $V_{DS} = 50$ V, the characteristic curve moves towards the left and $I_D$ increases to $10^{-7}$ A at $V_{BG} = -5$ V and the TFT is in the on state.

Figure 3.13 shows the calculated output characteristic curves, i.e. $I_D$ as a function of $V_{DS} = V_{TG}$, with $V_{BG} = -5$ V for TFTs with different $d_{ti}$. It shows that with $d_{ti} = 330$ nm, $I_D$ increases to $10^{-7}$ A at $V_{DS} = 10$ V. With thicker $d_{ti}$ of 1 and 2 μm, $I_D$ reaches $10^{-7}$ A at $V_{DS} = 28$ and 56 V, respectively.

3.3.4 Experimental verification

In order to verify experimentally the validity of the dual-gate TFT model, the characteristics of the CdSe TFT samples described in Section 3.3.2 were measured, and the results were compared with the calculation using the model. As shown in Figure 3.10, since the bottom gate has complete
Figure 3.12: Calculated bottom gate transfer characteristic curves \((I_D - V_{BG})\) of a dual-gate TFT operated in the high voltage protection mode. The top gate and drain-source voltages used were \(V_{TG} = V_{DS} = 10\) and \(50\) V. The top and bottom gate insulator thicknesses are: \(d_{ti} = 2\) µm and \(d_{bi} = 360\) nm.

coverage of the entire channel, it was used as the protection gate and was connected to the drain of the TFT. The top gate was used as the control gate and was biased with \(V_{TG} = -5\) V.

Figure 3.14 (solid line) shows the measured output characteristic curve of a CdSe TFT operated in the high voltage protection mode. It shows that the TFT can remain in the off state for \(V_{DS} < 7\) V. Further increase of \(V_{DS}\) causes \(I_D\) to increase significantly and \(I_D\) is higher than \(10^{-7}\) A at \(V_{DS} = 12.5\) V. Also shown in Figure 3.14 (dashed line) is the theoretical prediction of the output characteristic curve for the dual-gate TFT with high voltage protection obtained using the TFT model, and it shows good overall agreement with the experimental measurement, except for the leakage current. The higher leakage current in the measurement can probably
Figure 3.13: The output characteristic curves of a dual-gate TFT operated in the high voltage protection mode \((V_{TG} = V_{DS})\). The bottom gate voltage \(V_{BG}\) is -5 V, and the bottom gate insulator thickness \(d_{bi}\) is 360 nm. The output characteristics are for top insulator thickness \(d_{ti}\) of 330 nm, 1 and 2 \(\mu m\).

be attributed to top gate leakage current. Since a large fraction of the top gate insulator layer was deposited using thermal evaporation, it may result in higher leakage than the bottom gate insulator which was deposited using plasma enhanced chemical vapor deposition (PECVD). The good agreement between the modelled and measured characteristic curves in Figure 3.14 verifies the validity of our dual-gate TFT model, and demonstrates the feasibility of high voltage protection.

The capability of a dual-gate TFT with high voltage protection to turn on the TFT at high \(V_{DS}\) is to be compared with the output characteristic curve of an ordinary single gate TFT (without high voltage protection), shown in Figure 3.14 with dotted line, where the TFT remains in the \textit{off} state as \(V_{DS}\) increases.
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3.4 X-RAY RESPONSE OF DETECTORS WITH HIGH VOLTAGE PROTECTION

In this section, the calculated TFT characteristic curves, as shown in Figure 3.13, are used to predict the pixel x-ray response for detectors with high voltage protection under normal and abnormal operational conditions. For a detector with high voltage protection, it is desirable for the x-ray response of a pixel to satisfy the following two requirements: 1) The $I_D$ of the dual-gate TFT increases at high x-ray exposure level such that the pixel potential $V_P$ reaches a saturation value $V_{sat}$ smaller than the maximum safe voltage $V_{max} = 50$ V; 2) Below saturation, the detector should
have linear response over the x-ray exposure range required for the particular diagnostic imaging application.

Shown in Figure 3.15 is an equivalent circuit diagram of a pixel of the real-time, self-scanned a-Se detector. The pixel potential $V_P$ increases as a function of the difference between the photo-conductive current generated in a-Se $I_{Se}$ and the channel current $I_D$ of the TFT, i.e. $(I_{Se} - I_D)$. The value of $I_{Se}$ can be calculated using [84]:

$$I_{Se} = \frac{1}{W} \int_{0}^{E_{max}} \Phi(E) A_P \eta(E) \frac{\mu_{ab}(E)}{\mu(E)} E dE,$$

(3.9)

where $W$ is the energy required to create an electron-hole pair in a-Se, $E_{max}$ is the maximum x-ray energy for the spectrum, $E$ is the incident x-ray photon energy, $\Phi(E)$ is the x-ray fluence rate, $A_P$ is the pixel area, $\mu(E)$ and $\mu_{ab}(E)$ are the x-ray attenuation and absorption coefficients of a-Se, and $\eta(E)$ is the x-ray quantum efficiency determined from:

$$\eta(E) = 1 - e^{-\mu(E) d_{Se}},$$

(3.10)

where $d_{Se}$ is the thickness of the a-Se layer. At the beginning of x-ray exposure, the TFT leakage current $I_D$ is negligible and thus $V_P$ increases linearly with x-ray exposure. As $V_P$ increases beyond a certain point, the protection gate turns on the TFT which causes an increase in $I_D$ and decrease in pixel x-ray sensitivity. When $I_D = I_{Se}$, there is no further image charge accumulation on the pixel and $V_P$ reaches its saturation value $V_{sat}$. Therefore the dual-gate TFT characteristics (shown in Figure 3.13) can be replotted to show the dependence of $V_{sat}$ on the pixel signal current $I_{Se}$, as shown in Figure 3.16. It shows that $V_{sat}$ increases as a function of $I_{Se}$, thus for the same TFT design, $V_{sat}$ will vary in different x-ray imaging applications. In this section, the pixel x-ray response are predicted using the calculated dual-gate TFT characteristic curves and $I_{Se}$ under fault and normal x-ray exposure conditions. Unless otherwise stated, the standard TFT geometry used in the calculation is $d_{bi} = 360 \text{ nm}$, $W = 20 \mu m$ and $L = 50 \mu m$. 

Figure 3.15: Diagram showing the equivalent circuit of each pixel for the calculation of pixel x-ray response. $I_{Se}$ is the photoconductive current generated in $a$-Se, $I_D$ is the channel current of the TFT, $V_P$ is the pixel voltage, $C_P$ is the total pixel capacitance, and $V_{Se}$ is the high voltage applied to the top surface of the $a$-Se.

3.4.1 Under fault x-ray exposure condition

An example of fault x-ray exposure condition, as summarized in Table 3.1, has maximum x-ray parameters, i.e. tube potential of 150 kVp and current of 800 mA. The source-detector distance (SDD) is 20 cm, i.e. the detector is placed against the x-ray collimator. Under such condition, the x-ray exposure rate at the input of the detector is $\sim 400$ R/s. For a 500 $\mu$m $a$-Se layer with an electric field $E_{Se}$ of 10 V/$\mu$m, using Eq. 3.9, a photoconductive current of $I_{Se} = 1.7 \times 10^{-7}$A can be generated for a 200 $\mu$m square pixel (the largest pixel size suitable for radiographic applications [84]).

As shown in Figure 3.16, this value of $I_{Se}$ corresponds to a $V_{sat}$ value of 10 V, 30 V or 60 V for top gate insulator thicknesses of $d_{ti} = 330$ nm, 1 or 2 $\mu$m, respectively. These values show that $V_{sat}$ for a given $I_{Se}$ is proportional to $d_{ti}$, which suggest that high voltage protection characteristics can be optimized by adjustment of $d_{ti}$. 
Table 3.1: The x-ray conditions and detector parameters used in the calculation of pixel x-ray response.
Figure 3.16: The saturation value $V_{sat}$ for the pixel potential as a function of pixel signal current $I_{Se}$ generated by the a-Se layer for different $d_{hi}$. The value for $d_{hi}$ is 360 nm.

### 3.4.2 Under normal diagnostic exposure

In order to assure that detectors with high voltage protection can satisfy the dynamic range required in radiographic applications, the pixel x-ray response under normal diagnostic x-ray exposure was calculated. In diagnostic imaging procedures, all the pixels on the detector are exposed to x-rays for the same length of time $t_x$, however they encounter a wide range of $I_{Se}$ due to variations in x-ray attenuation by different human anatomy. For each $I_{Se}$ value, the pixel x-ray response was calculated using a time interval of $\delta t = 1 \mu s$ over the entire exposure time $t_x$. The increase of $V_P$, $\Delta V_P$, within each $\delta t$ was calculated using:

$$\Delta V_P = \frac{(I_{Se} - I_D) \delta t}{C_P},$$  \hspace{1cm} (3.11)
where $C_P$ is the pixel storage capacitance. At the end of each $\delta t$, the TFT channel current $I_D$ was updated according to the new $V_P$ value and the characteristic curve of the dual-gate TFTs shown in Figure 3.13. $V_P$ at the end of x-ray exposure was calculated by integrating $\Delta V_P$ over $t_x$. In order to take into account any image signal loss due to TFT leakage current before the entire image is read out, a charge decay time of 1 s was included in the calculation of the final $V_P$. During the decay time, $V_P$ was updated at a decay time interval $\delta t_d$ of 1 ms using:

$$V_P = V_P e^{-\frac{\delta t_d}{RC_P}} ,$$

(3.12)

where $R$ is the resistance of the TFT determined from the slope of the characteristic ($I_D - V_{DS}$) curve for a given $V_P$ (i.e. $V_{DS}$) value.

### 3.4.2.1 Mammography

The mammographic x-ray parameters used in our calculation are summarized in Table 3.1, which results in a maximum exposure of 0.9 R to the detector. Since a pixel pitch of 50 $\mu$m is required for mammography, the standard TFT channel length of $L = 50$ $\mu$m is no longer adequate. The TFT geometry chosen for a mammographic detector has channel width and length of 10 and 25 $\mu$m, respectively, which results in the same $W/L$ ratio as the standard TFT geometry and therefore identical TFT characteristics. The maximum $I_{Se}$ corresponds to the region of the detector that is outside the region of the breast, i.e. receiving the unattenuated (raw) beam. Based on an exposure time of $t_x = 0.75$ s, the raw exposure of 0.9 R corresponds to a maximum $I_{Se}$ of $3.4 \times 10^{-12}$ A per pixel. The standard value of $C_P$ used in the calculation is 1 pF, and in order to illustrate the effect of $C_P$ on pixel x-ray response, the calculation was also performed for $C_P = 0.5$ pF. Shown in Figure 3.17 is the predicted pixel x-ray response of a mammographic detector. It shows that with $C_P = 1$ pF, even with a very thin top gate insulator layer of $d_t = 330$ nm and the standard
bottom gate insulator of $d_{bi} = 360$ nm, the detector has a linear response up to 1.5 R, which not only satisfies the dynamic range of 0.6 - 240 mR required for mammography [84], but also includes the raw exposure (0.9 R) to a mammographic detector. When $C_P$ is reduced to 0.5 pF, the slope of the x-ray response increases and the dynamic range is reduced by a factor of two. Therefore for a given dual-gate TFT geometry, the choice of $C_P$ is also important in order to obtain an adequate detector dynamic range.

Figure 3.17: The predicted detector pixel response curves for a mammographic detector with high voltage protection. The pixel capacitance used was $C_P = 1$ and 0.5 pF. The TFT parameters used in the calculations are: top gate insulator thickness $d_{ti} = 330$ nm, and bottom gate insulator thickness of $d_{bi} = 360$ nm.

3.4.2.2 Chest radiography

Chest radiography is the most frequently used radiographic procedure. The normal x-ray exposure conditions used for chest radiography are summarized in Table 3.1, which results in a maximum
exposure of 10 mR to the detector. With a pixel pitch of 200 μm, the maximum $I_{sc}$ generated for an x-ray exposure time of $t_x = 5$ ms is $\sim 8.0 \times 10^{-10}$ A. Shown in Figure 3.18 is the pixel x-ray response for chest radiography. It shows that with $d_{ti} = 330$ nm and $C_P = 1$ pF, the detector has a linear response up to 9 mR, which encompasses the range of 30 μR - 3 mR required for chest radiography [84]. With $d_{ti}$ increased to 1 μm, as also shown in Figure 3.18, the range of linear x-ray response extends to 27 mR to include the raw radiation exposure of 10 mR.

![Figure 3.18](image-url)

Figure 3.18: The predicted detector pixel response curves for a chest radiographic detector with high voltage protection. The standard pixel capacitance of $C_P = 1$ pF was used. The TFT parameters used in the calculations are: bottom gate insulator thickness of $d_{bi} = 360$ nm, and top gate insulator thicknesses of $d_{ti} = 330$ nm and 1 μm.
CHAPTER 3. DETECTORS WITH HIGH VOLTAGE PROTECTION

3.5 DISCUSSION

3.5.1 Strategies for design optimization

As shown in Section 3.4, the pixel potential saturation value $V_{sat}$ depends on the following two factors: (1) pixel signal current $I_{Se}$; and (2) the thicknesses of the two gate insulators, $d_{ti}$ and $d_{bi}$. The dynamic range of an x-ray imaging detector depends on $V_{sat}$ as well as the pixel capacitance $C_P$. Therefore the strategy recommended for designing an x-ray imaging detector with high voltage protection is: (1) choose the values for $d_{ti}$ and $d_{bi}$ based on $V_{sat} < V_{max}$ and the maximum $I_{Se}$ under fault x-ray conditions; (2) adjust the value of $C_P$ in order to satisfy the required dynamic range for the specific x-ray imaging application. The results in Section 3.4 showed that with the standard $d_{bi} = 360$ nm and $C_P = 1$ pF, a practical $d_{ti}$ of less than 1 $\mu$m can ensure a safe pixel saturation potential $V_{sat}$ and satisfy the required dynamic range for both mammography and chest radiography.

3.5.2 Practical implementation of high voltage protection

The most common dual-gate TFT structure is the top contact structure, which has been used in both CdSe TFTs and a-Si:H TFTs. As shown in Figure 3.10, the source and drain electrodes make contact with the semiconductor layer through vias in the top gate insulator. There are two methods for forming the protection (top) gate with top contact dual-gate TFTs: 1) By extending the pixel electrode (drain) to cover most of the channel, as shown in Figure 3.19 (a). This is the simplest approach as it does not require extra processing steps. Since there must be a gap between the drain and source electrodes, this section of the channel cannot be controlled by the protection gate. However this section of the channel is permanently turned on by the top bias electrode of the a-Se layer and holes that get trapped at the interface between a-Se and the top SiO$_2$ layer,
therefore it will not prevent the top gate from turning on the TFT for high voltage protection; 2) By adding two more levels of process to build a extended (mushroom) pixel electrode structure, as shown in Figure 3.19 (b), so that the section of the channel under the gap can also be controlled by the top gate. However with both methods, since the source electrode overhangs the TFT channel, this shielded section cannot be turned on by the protection gate. Therefore as shown in Figure 3.19, the shielded section has to be heavily doped (n type) so as to make it conducting permanently for example by controlled lateral diffusion of the chromium source electrode in the channel [105].

3.6 CONCLUSIONS

A novel concept for high voltage protection for a real-time, self-scanned a-Se detector has been investigated theoretically and experimentally. It employs a dual-gate TFT at each pixel, the bottom gate is for scanning control, and the top gate is connected to the pixel electrode for high voltage protection. Before the pixel potential reaches a damaging value, the protection gate turns on the TFT which limits the pixel potential rise to \( V_{\text{sat}} \) thus avoiding high voltage damage. The TFT characteristic curves predicted using a dual-gate TFT model showed that \( V_{\text{sat}} \) is proportional to the ratio between the thicknesses of the top and bottom gate insulators. The validity of the model was experimentally verified with measurements from one type of dual-gate CdSe TFT samples. The pixel x-ray response of detectors with high voltage protection was predicted for both normal and abnormal x-ray exposure conditions. It was shown that the dynamic range of the detector also depends on the pixel capacitance. The model showed that with the standard pixel capacitance of 1 pF and practical top gate insulator thicknesses of less than 1 \( \mu \text{m} \), the detector can provide a safe \( V_{\text{sat}} \) while simultaneously satisfying the dynamic range requirements for diagnostic x-ray imaging (e.g. mammography and chest radiography).
Figure 3.19: Cross section showing practical implementation of high voltage protection using dual-gate TFT structure: (a) extending the drain electrode to form the top gate; (b) adding another insulation layer and a metal layer to form the top gate and the pixel electrode (i.e. the mushroom structure). (note that the thicknesses of layers are not to scale)
Chapter 4

Theoretical Analysis of Detective Quantum Efficiency

4.1 INTRODUCTION

There has been rapid development of large area, flat-panel digital x-ray imaging detectors for diagnostic radiology using active matrix technologies. An active matrix consists of a two dimensional array of thin film transistors (TFT) or switching diodes made with amorphous or polycrystalline semiconductor materials. There are two general approaches to making flat-panel x-ray imagers, the first is the indirect method. It uses phosphor screens or structured cesium iodide (CsI) scintillators to first convert x-rays to visible light, which is then read out with an active matrix array with an additional photodiode at each pixel [56–59]. The second approach is the direct method, which is also referred to as a self-scanned a-Se detector. It uses an x-ray photoconductor such as amorphous selenium (a-Se) to convert x-rays directly to a charge image, which is then electronically read out.

\^This chapter has been submitted as a paper to Med. Phys., entitled “Digital radiology using active matrix readout of amorphous selenium: Theoretical analysis of detective quantum efficiency”.

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by an active matrix [60,67]. In the direct method, our approach is to make a real-time detector [67], as opposed to a radiographic detector [60]. The structure of the real-time self-scanned a-Se detector is shown schematically in Figure 1.9, where a bias electrode is deposited directly on the top surface of the a-Se layer in order to establish an electric field, and the bottom surface of a-Se is in contact with the active matrix. Each pixel of the active matrix consists of a TFT for image readout, a pixel electrode to collect image charge generated by a-Se, and a storage capacitor for holding the charge before readout. The horizontal gate lines turn on the TFTs one row at a time and transfer image charge from pixel electrodes to vertical data lines which are connected to external charge amplifiers. The multiplexer converts the parallel amplified signal to a serial output for digitization.

A spatial frequency independent theoretical study of the signal and noise performance of the self-scanned a-Se detector (henceforth referring to the real-time approach, unless otherwise specified) has previously been performed based on the x-ray response and electronic noise of a single pixel [84]. The analysis was applied to detector parameters proposed for three examples of x-ray imaging applications: mammography, chest radiography and fluoroscopy. The results showed that the detector can potentially be a x-ray quantum noise limited system for all these applications. However, this study did not provide a easy means to compare the imaging performance of the self-scanned a-Se detector to that of other detector technologies. Therefore in this chapter a more complete study of the signal and properties of the self-scanned a-Se detector is presented in the form of spatial frequency dependent detective quantum efficiency (DQE), which is evaluated as a function of detector design parameters such as pixel fill factor and electronic noise.

Detective quantum efficiency of an imaging detector may be defined as the ratio of the square of the output image signal to noise ratio (SNR) to that of the input. It can be studied as a function of spatial frequency, thereby provides information on the overall signal and noise performance of
CHAPTER 4. THEORETICAL ANALYSIS OF DQE

an imaging detector [6]. The spatial frequency dependent DQE, \(DQE(f)\), is a powerful means to intercompare the imaging performance of different detectors. Several investigations of \(DQE(f)\) for the indirect flat-panel approach have been reported, which include: (1) Experimental measurement of \(DQE(f)\) of a fluoroscopic flat-panel detector using CsI scintillators and a-Si TFT/photodiode arrays [57]; (2) Experimental measurement of the \(DQE(f)\) of a full-size fluoroscopic flat-panel detector under clinical evaluation, which employs a CsI scintillating screen, and an active matrix with a switching diode and a photodiode at each pixel [58]; (3) Theoretical analysis of \(DQE(f)\) using a cascaded imaging system model for an indirect flat-panel detector [117]. However, although these were digital detectors, no systematic study of the effect of aliasing has been reported. Also, there has been no study reported for the spatial frequency dependent DQE of the direct flat-panel detectors.

Shown in Figure 4.1 is a cross-sectional view of the self-scanned a-Se detector, where the width of each pixel is \(a\) and the pixel pitch is \(d\). For all practical purposes, \(a \leq d\) and image charges in the gap between pixel electrodes are lost and do not contribute to the image. Since a-Se has nearly perfect resolution, the only presampling filtration is by the aperture function of the pixel electrodes. Since the pixel aperture function is a sinc function \(\text{sinc}(af)\) with the first zero at the spatial frequency of \(1/a\), and the sampling Nyquist frequency \(f_{NY}\) with pixel pitch \(d\) is \(f_{NY} = 1/2d\), such detectors are inherently undersampled and hence aliasing is inevitable and cannot be ignored. Thus in this chapter, the \(DQE(f)\) for direct flat-panel detectors using self-scanned readout of a-Se is analyzed theoretically and with full account of the effects of aliasing. An interpretation of \(DQE(f)\) for such undersampled a-Se detectors is proposed. The propagations of x-ray signal and noise through different detector stages are described, and an expression for \(DQE(f)\) of the complete detector is derived. The spatial frequency dependent DQE is predicted for detector parameters
proposed for mammography, chest radiography and fluoroscopy. The dependence of $DQE(f)$ on x-ray exposure, added noise and active matrix design (such as the pixel fill-factor) is analyzed. Strategies for maximizing $DQE(f)$ are discussed.

![Cross-sectional view of a-Se detector](image)

Figure 4.1: The cross-sectional view of the self-scanned a-Se detector.

4.2 BACKGROUND

4.2.1 Cascaded linear system analysis

The propagation of signal (in the form of mean number of quanta per unit area $\bar{S}$) and noise (in the form of noise power spectrum (NPS)) through cascaded linear systems has been studied by Rabbani et al. [118,119] Cunningham et al. extended the theory of signal and noise propagation and obtained a generalized expression for $DQE(f)$ of a cascaded linear system [120]. According to
the cascaded linear systems theory, each stage of an imaging detector may consist of only one of these three types of processes: (1) an amplification process; (2) a stochastic blurring process and (3) a deterministic blurring process.

For stage \(i\) with amplification only, such as the conversion from x-rays to light photons by a phosphor screen, the output mean number of quanta per unit area \((\Phi_i)\) and NPS \([S_i(f)]\) are, respectively [118,119]:

\[
\Phi_i = g_i \Phi_{i-1}
\]

\[
S_i(f) = g_i^2 S_{i-1}(f) + \sigma_{g_i}^2 \Phi_{i-1},
\]

(4.1)

where \(\Phi_{i-1}\) and \(S_{i-1}(f)\) are the mean number of quanta and the noise power spectrum incident on stage \(i\), respectively, and \(g_i\) and \(\sigma_{g_i}^2\) are the mean and variance of the gain of the \(i\)th stage. A special case of amplification stage is the binary selection process, such as x-ray detection. For x-ray detection, \(g_i\) and \(\sigma_{g_i}^2\) in Eq. 4.1 equal \(\eta\) and \(\eta(1 - \eta)\), respectively [120], where \(\eta\) is the quantum efficiency of the detector.

Stochastic blurring is a process in which quanta are dispersed randomly into a spatial distribution with a probability given by the point spread function (PSF). An example of such a process is the scattering of light photons in a phosphor screen. For stage \(i\) with stochastic spreading only, and a modulation transfer function (MTF) of \(T_i(f)\), the spatial distribution of the output image quanta \(\Phi_i(f)\) and NPS \(S_i(f)\) are, respectively [118]:

\[
\Phi_i(f) = \Phi_{i-1}(f) T_i(f),
\]

\[
S_i(f) = T_i^2(f) S_{i-1}(f) + [1 - T_i^2(f)] \Phi_{i-1}.
\]

(4.2)

In some detectors, there are also blurring stages that follow the conventional convolution process, i.e. deterministic blurring, such as the integration of image information onto individual
pixels in a digital detector [121]. For stage $i$ with deterministic blurring and MTF of $T_i(f)$, the output signal spectrum $\Phi_i(f)$ and noise power spectrum $S_i(f)$ are, respectively:

\[
\Phi_i(f) = \Phi_{i-1}(f)T_i(f),
\]
\[
S_i(f) = T_i^2(f)S_{i-1}(f).
\]  

(4.3)

The method of DQE analysis using a cascaded linear systems model has been applied, by various investigators, to different x-ray imaging systems and has proven to be a powerful tool. Systems examined include: an electronic portal imaging system [122]; an x-ray image intensifier / TV camera chain [123]; a hybrid direct conversion mammographic detector using thick silicon photodiodes and CCD [124]; and an indirect flat-panel detector [117]. The intention of the present work is to apply these well established tools to the self-scanned $a$-Se detectors, and address the problems that are specific to $a$-Se, i.e. aliasing due to undersampling. Therefore a fourth type of process is added into the cascaded systems model: aliasing. However, the addition of aliasing makes the interpretation of $DQE(f)$ differ from its original definition.

4.2.2 Digital detectors with aliasing

The concepts of MTF, NPS and DQE have proven to provide useful descriptions of the resolution, noise and signal to noise ratio of analog linear systems, such as screen-film [6, 125]. These linear system parameters also apply to digital detectors with no undersampling. However the interpretation of these parameters becomes more difficult in an undersampled digital detector, where aliasing occurs when spatial frequency components higher than the sampling Nyquist frequency ($f_{NY}$) are folded back to the low frequency region. Since the system response for an aliased detector depends not only on the properties of the detector, but also on the relative phase of the input signal to the detector pixel positions, the requirement of shift invariance for a linear system is no longer satisfied.
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The effects of aliasing on MTF and NPS on undersampled digital detectors have been previously studied [126,127]. In a review article on the interpretation of MTF, NPS and DQE in an undersampled digital detector [128], Dobbins provided a detailed explanation of the main difficulties involved in applying the classical definition of MTF and NPS to an undersampled detector. For MTF, the use of presampling MTF is suggested if one is interested in the detector response to a single frequency sinusoidal wave. For NPS of an undersampled detector, the NPS always contains the aliased components from spatial frequencies higher than $f_{NY}$. This is because noise power arising from frequencies above $f_{NY}$ and aliased to frequencies below $f_{NY}$ cannot be distinguished from noise arising from below $f_{NY}$. Therefore the NPS available to us for measurement is always the aliased NPS.

Thus in the present DQE analysis, the presampling MTF and the aliased NPS were used. Therefore for stage $i$ with aliasing only, the signal spectrum $\Phi_i(f)$ and the noise power spectrum $S_i(f)$ are, respectively:

$$
\Phi_i(f) = \Phi_{i-1}(f),
$$

$$
S_i(f) = \sum_{n=-\infty}^{\infty} S_{i-1}(f - \frac{n}{d}),
$$

(4.4)

where $d$ is the sampling distance, i.e. the pixel pitch of the active matrix. As described by Dobbins [128], this method for dealing with aliasing correctly represents the detector response to a single frequency sinusoidal input. It will provide an unambiguous description of the detector properties up to the Nyquist frequency.
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4.3 THEORY AND METHODS

4.3.1 Signal and noise propagation

The propagation of signal and noise through the detector was analyzed using the cascaded linear systems model. As shown with the cross-sectional view in Figure 4.1, the self-scanned a-Se detector can be divided into the following stages: (1) x-ray attenuation by a-Se; (2) stochastic blurring due to a-Se; (3) gain stage associated with conversion of x-rays to electron-hole pairs in a-Se; (4) addition of electronic noise due to dark current shot noise in a-Se; (5) integration of image charge on pixel electrodes; (6) aliasing due to undersampling; (7) addition of electronic noise due to the TFTs and the charge amplifiers.

Although the study of $DQE(f)$ is often limited to monoenergetic x-rays [129], here the effect of a broad x-ray spectrum is included in the analysis of $DQE(f)$, which has previously been considered in the study of $DQE(f)$ of phosphor screens [130,131] and represents the realistic detector performance in a diagnostic imaging application. In this section, the signal and noise propagation for a broad x-ray spectrum is described, from which the results for monoenergetic x-rays can easily be derived.

The flow chart shown in Figure 4.2 indicates the mechanisms (gain, blurring, aliasing, or addition of noise) for signal and noise propagation at each of the seven detector stages. It will be used in the following derivation for the propagation of signal and noise through the self-scanned a-Se detector. Before x-rays enter the detector, i.e. at stage 0, the x-rays have an energy spectrum $\Phi_0(E)$ and a maximum photon energy $E_{\text{max}}$. The mean input x-ray quanta per unit area $\Phi_0$ can then be calculated by integrating over the entire x-ray energy spectrum, and obtain a spatially
white input signal spectrum $\mathcal{F}_0(f)$:

$$\mathcal{F}_0(f) = \mathcal{F}_0 = \int_0^{E_{\text{max}}} \Phi_0(E) dE.$$  (4.5)

Since the x-ray quantum noise also has a spatially white spectrum and follows the Poisson distribution, the noise power spectrum $S_0(f)$ at the input to the detector is given by:

$$S_0(f) = \mathcal{F}_0.$$  (4.6)

Then following the flow chart in Figure 4.2, the mean number of quanta per unit area $\Phi_i$, the signal spectrum $\Phi_i(f)$, and the noise power spectrum $S_i(f)$ at each of the seven stages of the detector can be determined as follows.

4.3.1.1 X-ray Attenuation

The interaction of incident x-ray quanta with the $a$-Se layer is a binary selection process which is a special case of an amplification stage. The energy dependent x-ray quantum efficiency, $\eta(E)$, of $a$-Se can be calculated using:

$$\eta(E) = 1 - e^{-\mu(E) d_{a-Se}},$$  (4.7)

where $\mu(E)$ and $d_{a-Se}$ are the linear attenuation coefficient and the thickness of the $a$-Se layer, respectively. Substituting $\eta(E)$ into Eq. 4.1 for the signal and noise propagation through a binary selection process:

$$\Phi_1(f) = \Phi_1 = \int_0^{E_{\text{max}}} \eta(E) \Phi_0(E) dE,$$

$$S_1(f) = \Phi_1 = \int_0^{E_{\text{max}}} \eta(E) \Phi_0(E) dE.$$  (4.8)

It shows that the x-ray quantum noise still follows the Poisson distribution after attenuation by the $a$-Se layer, and both the signal and noise power spectra are spatially white.
1. x-ray attenuation  
\[ \eta(E) \]

2. a-Se blurring  
\[ T_s(f) \]

3. amplification  
\[ g(E), A_s(E) \]

4. shot noise  
\[ S_d(f) \]

5. aperture function  
\[ T_a(f) \]

6. aliasing  
\[ S_6(f) = \sum S_5(f-n/d) \]

7. electronic noise  
\[ S_e(f) \]

Figure 4.2: The flow chart showing the propagation of signal and noise power spectra through the seven detection stages of the self-scanned a-Se detector.
4.3.1.2 Blurring by a-Se

The inherent resolution of a-Se layers has been studied theoretically by Que and Rowlands [132]. Seven sources of blurring were considered: range of primary photoelectrons; reabsorption of k fluorescence; reabsorption of Compton scattered photons; charge diffusion; oblique incidence of x-rays; the electrostatic and the space charge effects. It was found that for diagnostic energy x-rays the blurring due to oblique incidence is the main effect. For lower energy x-ray photons, such as those used in mammography, the blurring due to k fluorescence is also significant. The model of Que and Rowlands was used to calculate the MTF due to a-Se, $T_{Se}(f)$, in the DQE analysis. The two dominant factors contributing to $T_{Se}(f)$, oblique x-ray incidence and k fluorescence reabsorption, both occur before the conversion gain stage of a-Se, and follow the stochastic blurring process. Therefore using Eq. 4.2 for the stochastic blurring process, the x-ray signal spectrum $\Phi_2(f)$ and noise power spectrum $S_2(f)$ can be obtained as:

$$\Phi_2(f) = \left[ \int_0^{E_{\text{max}}} \eta(E) \Phi_0(E) dE \right] T_{Se}(f),$$

$$S_2(f) = \left[ S_1(f) - \Phi_1 \right] T^2_{Se}(f) + \Phi_1. \quad (4.9)$$

Substituting the values for $\Phi_1(f)$ and $S_1(f)$ shown in Eq. 4.8 into the expression for $S_2(f)$ in Eq. 4.9, the first term is zero and a spatially white spectrum for $S_2(f)$ is obtained:

$$S_2(f) = \Phi_1. \quad (4.10)$$

Eqs. 4.9 and 4.10 indicates that although the signal spectrum is blurred by the MTF of a-Se, the noise power spectrum remains white.
4.3.1.3 Amplification gain of a-Se

The mean amplification gain, \( \bar{g}(E) \), of a-Se, i.e. the mean number of electron-hole pairs collected after absorption of an x-ray of energy \( E \) (ignoring the small effect of \( k \) fluorescence escape), is given by:

\[
\bar{g}(E) = \frac{E}{W_{\pm}},
\]

where \( W_{\pm} \) is the energy required to free a electron in a-Se and depends on the electric field in a-Se. The value for \( W_{\pm} \) has previously been determined experimentally using a-Se pulse height spectroscopy as 50 eV at an electric field of 10 V/\( \mu \text{m} \) [133]. The variance associated with the conversion gain, \( \sigma_g^2(E) \), can be determined from the pulse height distribution of the gain \( g \) for absorption of a single x-ray of energy \( E \) in a-Se. The pulse height distribution, \( P(g) \), is the number of events detected as a function of \( g \). The noise associated with the gain stage is described by the Swank factor \( A_S \), which is defined in terms of the moments, \( m_0, m_1, \) and \( m_2 \), of \( P(g) \) [134, 135]:

\[
A_S = \frac{m_1^2}{m_0 m_2}.
\]

The first three moments of the gain are, respectively: \( m_0 = 1, m_1 = \bar{g}(E) \) and \( m_2 = \bar{g}^2(E) + \sigma_g^2(E) \) [120]. Therefore \( A_S(E) \) is related to \( \bar{g}(E) \) and \( \sigma_g^2(E) \) by [136]:

\[
A_S(E) = \frac{\bar{g}^2(E)}{\bar{g}^2(E) + \sigma_g^2(E)}.
\]

The pulse height distribution has been previously calculated for a-Se by taking into account \( k \) fluorescence escape and the Poisson distribution of the number of charge carriers released per x-ray [137]. The results predicted a Swank factor, \( A_S(E) \), ranging from 0.8 for \( E \) just above the \( k \)-edge \( E_k \) (12.7 keV) to practically unity (for \( E > 60 \) keV and \( E < E_k \)). Using Eq. 4.1, the signal spectrum after the gain stage is multiplied by \( \bar{g}(E) \) for each x-ray energy and then integrated over.
the entire x-ray energy spectrum:

$$\bar{F}_3(f) = \left[ \int_0^{E_{\text{max}}} \bar{g}(E) \eta(E) \Phi_0(E) dE \right] T_{\text{Se}}(f).$$

(4.14)

Substituting the expression for $A_S(E)$ in Eq. 4.13 into Eq. 4.1 for the propagation of NPS at the gain stage, the NPS can be calculated from known values of $A_S(E)$ using:

$$S_3(f) = \int_0^{E_{\text{max}}} \left[ \bar{g}^2(E) + \sigma^2_0(E) \right] \eta(E) \Phi_0(E) dE$$

$$= \int_0^{E_{\text{max}}} \frac{\bar{g}^2(E)}{A_S(E)} \eta(E) \Phi_0(E) dE.$$  (4.15)

### 4.3.1.4 Dark current shot noise

The dark current intrinsic to the a-Se layer is caused by thermal excitation of charge carriers in the a-Se layer. Due to the large bandgap energy ($E_g=2.2$ eV) and the relatively small density of states in the bandgap, the intrinsic dark current of a-Se is almost negligible. However, in order to make an x-ray imaging detector, electrode contacts have to be made to the surfaces of the a-Se, which can cause charge injection from the electrodes to be the dominant source of dark current in a-Se. The type of contact that minimizes charge injection is the blocking contact. An electrode is said to have a blocking contact if carriers generated within the a-Se layer are permitted to leave freely, but carriers from the electrodes are prevented, or blocked from entering the a-Se. Different blocking layers have been engineered for a-Se which can minimize the charge injection, but none are perfect. Practically, the dark current $I_d$ can usually be kept below $10^{-12}$ A mm$^{-2}$. In x-ray imaging applications, dark current generates an offset pixel charge which, provided it is small enough not to saturate the pixel response, can be eliminated through offset correction. However, the statistical fluctuation, or shot noise, in the dark current cannot be eliminated and always contributes to the total noise of the detector. The shot noise associated with the dark current $I_d$ has a white spectrum,
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and its variance $\sigma_d^2$ is given by [138]:

$$
\sigma_d^2 = \frac{I_d T_D}{e}, 
$$

(4.16)

where $e$ is the electronic charge, and $T_D$ is the time during which dark current accumulates on the pixel electrodes. In fluoroscopy, $T_D$ is the frame time $T_F$ between two subsequent readouts of the detector, i.e. 33 ms. The noise power spectrum, $S_d(f)$, associated with the dark current shot noise will be added to the total NPS from the previous stage shown in Eq. 4.15:

$$
S_d(f) = \int_0^{E_{\text{max}}} \frac{\bar{g}^2(E)}{A_5(E)} \eta(E) \Phi_0(E) dE + S_d(f).
$$

(4.17)

4.3.1.5 Integration of charge on pixel electrodes

When the charge created in $\alpha$-Se reaches the surfaces of the detector, it will be collected and integrated on each pixel electrode. If the pixel electrode is square and with width $a$, the two dimensional MTF associated with the aperture function of the pixel electrodes, $T_a(f_x, f_y)$, can be written as:

$$
T_a(f_x, f_y) = \text{sinc}(a f_x) \text{sinc}(a f_y), 
$$

(4.18)

where $\text{sinc}(a f_x) = \frac{\sin(\pi a f_x)}{\pi a f_x}$. The process of image charge integration on pixel electrodes is deterministic blurring [121], therefore using Eq. 4.3, the one dimensional signal spectrum $\Phi_5(f)$ is blurred by $T_a(f)$ which is a sinc function:

$$
\Phi_5(f) = \text{sinc}(af) \Phi_4(f),
$$

(4.19)

and the output two dimensional NPS is multiplied by the square of $T_a(f_x, f_y)$:

$$
S_5(f_x, f_y) = \text{sinc}^2(a f_x) \text{sinc}^2(a f_y) S_4(f_x, f_y).
$$

(4.20)

Since $S_4(f_x, f_y)$ is white, the shape of the output NPS, i.e. the presampling NPS, follows the square of $T_a(f_x, f_y)$. 
4.3.1.6 Aliasing due to undersampling

As described in Section 4.2, the use of presampling MTF and aliased NPS for the undersampled self-scanned a-Se detector provides the correct detector response for spatial frequencies up to the Nyquist frequency. According to Eq. 4.4 for signal and noise propagation through the aliasing stage, only NPS is affected. The one dimensional NPS, i.e. the central slice of the 2-D NPS with \( f_y = 0 \), after aliasing is given by:

\[
S_6(f) = S_6(f_x) = \sum_{n=-\infty}^{\infty} S_5(f_x - \frac{n}{d}, f_y - \frac{n}{d})|_{f_y=0} = S_4(f) \sum_{n=-\infty}^{\infty} \text{sinc}^2[a(f - \frac{n}{d})] \sum_{n=-\infty}^{\infty} \text{sinc}^2[a(f_y - \frac{n}{d})]|_{f_y=0}. \tag{4.21}
\]

It can be shown (Appendix A) that when \( a \leq d \) (which is always true for a pixelated active matrix detector), the aliasing of the presampling NPS (whose shape follows the square of a sinc function) produces a white spectrum in the frequency domain:

\[
\sum_{n=-\infty}^{\infty} \text{sinc}^2[a(f - \frac{n}{d})] = \frac{d}{a}. \tag{4.22}
\]

Therefore Eq. 4.21 becomes:

\[
S_6(f) = \frac{d^2}{a^2} \int_0^{E_{\text{max}}} \frac{\tilde{g}^2(E)}{A_S(E)} \eta(E) \tilde{G}_0(E) dE + S_d(f). \tag{4.23}
\]

As shown Eq. 4.23, the NPS after aliasing is spatially white, and the noise power density is inversely proportional to the pixel geometric fill factor \( F_P = a^2/d^2 \). This will result in the decrease of \( \text{SNR}^2 \) at the output of the detector by the factor \( F_P \).

4.3.1.7 Addition of readout electronic noise

During image readout \( q_n \), the rms electronic noise associated with the TFTs and the external charge amplifiers, will be added to the total noise power [84]. Since the electronic noise generated from
each pixel is independent from each other, the NPS \( \langle S_a(f) \rangle \) associated with the pixel electronic noise is spatially white, and will be added to \( S_6(f) \) and form the final NPS at the output of the detector:

\[
S_7(f) = \frac{d^2}{\lambda^2} \int_0^{E_{\text{max}}} \frac{g^2(E)}{A_5(E)} \eta(E) \Phi_0(E) + S_a(f) \right) + S_a(f). \tag{4.24}
\]

### 4.3.2 Detective Quantum Efficiency

#### 4.3.2.1 Propagation of detective quantum efficiency

\( DQE(f) \) represents the efficiency of x-ray utilization of the detector, and is unity for an ideal detector. For a practical x-ray imaging detector, \( DQE(f) \) is degraded through each detector stage. Therefore it is important to identify the stages that contribute most to the degradation in \( DQE \) so as to guide optimization of the detector design in order to maximize \( DQE(f) \). \( DQE(f) \) is defined as the ratio of the signal to noise ratio (SNR) squared at the output to that at the input of the detector: \( DQE(f) = \frac{SNR^2_{\text{out}}(f)}{SNR^2_{\text{in}}(f)} \). From the detector input signal and noise power spectra given in Eqs. 4.5 and 4.6, the value for \( SNR^2_{\text{in}}(f) \) is a constant equal to \( \Phi_0 \). Then the \( SNR^2_{\text{out}}(f) \) after the \( i \)th stage can be calculated from the propagation of signal spectrum \( \Phi_i(f) \) and the noise power spectrum \( S_i(f) \) up to the \( i \)th stage, and obtain \( DQE_i(f) \) using:

\[
DQE_i(f) = \frac{\Phi_i^2(f)}{\Phi_0 S_i(f)}. \tag{4.25}
\]

\( DQE_7(f) \) is the final \( DQE(f) \) of the entire detector and is obtained by substituting the expressions for \( \Phi_7(f) \) and \( S_7(f) \) into Eq. 4.25:

\[
DQE(f) = \frac{\frac{a^2}{d^2} \int_0^{E_{\text{max}}} \frac{g(E) \eta(E) \Phi_0(E) dE}{\Phi_0 [\int_0^{E_{\text{max}}} \frac{g^2(E)}{A_5(E)} \eta(E) \Phi_0(E) dE + S_a(f) + \frac{a^2}{d^2} S_a(f) ]} }{\frac{\eta T_{\text{eff}}^2(f) T_{\text{eff}}^2(f)}{\lambda_s + \frac{S_a(f)}{g^2 \eta \Phi_0} + \frac{a^2}{d^2} S_a(f)}}. \tag{4.26}
\]

For the special case of monoenergetic x-rays, Eq. 4.26 reduces to:

\[
DQE(f) = \frac{\frac{a^2}{d^2} \eta T_{\text{eff}}^2(f) T_{\text{eff}}^2(f)}{\lambda_s + \frac{S_a(f)}{g^2 \eta \Phi_0} + \frac{a^2}{d^2} S_a(f)} \tag{4.27}
\]
which, compared to Eq. 4.26, shows a more straightforward dependence of $DQE(f)$ on detector parameters (e.g. $F_p$, quantum efficiency $\eta$ and electronic noise). Eq. 4.27 shows that $DQE(f)$ is proportional to the quantum efficiency $\eta$ of the $a$-Se layer and the pixel fill factor $F_p$ of the active matrix. The shape of $DQE(f)$ follows that of the square of the MTF of the detector $[T_{q_e}^2(f) T_{a}^2(f)]$.

Various sources of noise, such as the Swank factor, the shot noise and the electronic noise, are in the denominator, and since $1/A_S$ is always greater than unity, these sources of noise all cause degradation in DQE.

4.3.2.2 $DQE(f)$ of detectors for x-ray imaging applications

The results of the DQE analysis were applied to detectors with parameters selected to be suitable for three specific x-ray imaging applications: mammography, chest radiography and fluoroscopy. The detector parameters used are identical to those chosen in the previous theoretical study of pixel signal and noise performance [84], and are summarized in Table 4.1. The dependence of $DQE(f)$ on imaging parameters, such as pixel fill factor $F_p$, electronic noise and x-ray exposures level, were analyzed for these three x-ray imaging applications.

4.3.3 Additional factors affecting $DQE(f)$

The DQE analysis so far is applicable to radiographic operations, where a single frame image is acquired after a short pulse of x-ray exposure during which the detector scanning is suspended. The same analysis is directly applicable to real-time fluoroscopic image acquisition if lag in the system, i.e. carry-over of image information from one frame to the subsequent frames, is negligible. Slight modification on the DQE analysis is needed if additional factors are considered. In this section, two additional factors that can affect $DQE(f)$ of a self-scanned a-Se detector will be discussed. The first factor is lag in real-time fluoroscopic operations. If there is lag in the system, the temporal
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<table>
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<th>Mammography</th>
<th>Chest Radiography</th>
<th>Fluoroscopy</th>
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</tr>
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<td>2.5 mm Al</td>
<td>2 mm Al</td>
</tr>
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<td>$2.12 \times 10^5$</td>
<td>$1.47 \times 10^5$</td>
</tr>
<tr>
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<td></td>
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<td>0.001</td>
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<td>0.0001 - 0.01</td>
</tr>
<tr>
<td>thickness of $\alpha$-Se $d_{se}$ ($\mu$m)</td>
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<td>500</td>
<td>500</td>
</tr>
</tbody>
</table>

Table 4.1: Detector and x-ray parameters used in the calculation of detective quantum efficiency.
CHAPTER 4. THEORETICAL ANALYSIS OF DQE

integration caused by lag will reduce the NPS of the images. The second factor is the effect of alternative radiographic detector configurations which employ an insulator layer between the a-Se layer and the contact electrode [60,102].

4.3.3.1 Lag in fluoroscopic imaging

For a self-scanned a-Se detector, lag can arise from two different sources: photoconductive lag of a-Se, and readout lag of the active matrix [84]. The origin of photoconductive lag is trapping of charge carriers at defects in a-Se. The subsequent release of these trapped charge exhibits single or multiple exponential decay depending on the energy distribution of the traps. The origin of readout lag is the incomplete transfer of charge from the pixels to the external charge amplifiers, and is determined by the ratio of the pixel readout time constant $\tau_{on}$ to the pixel readout time $T_{on}$ [84]. The effect of lag on NPS has been studied for an x-ray image intensifier /video system by evaluating the lag induced reduction in temporal bandwidth [139]. This approach is used here to calculate the reduction of NPS by lag in the a-Se detector.

For a detector without lag, the detector temporal frequency response $R_0(\nu)$, where $\nu$ is the temporal frequency, is determined by the frame integration time $T_F$ (1/30 s) as:

$$R_0(\nu) = \text{sinc}(T_F\nu).$$

(4.28)

When lag is introduced, which has a temporal frequency response of $R_L(\nu)$, the resulting response for the detector, $R(\nu)$, is given by:

$$R(\nu) = R_0(\nu) R_L(\nu).$$

(4.29)

Therefore the noise reduction by lag can be represented by $r_l$, which is the ratio of the total noise
power for a detector with lag to that without lag:

$$r_l = \frac{\int_0^\infty R^2(\nu)d\nu}{\int_0^\infty R_0^2(\nu)d\nu}. \quad (4.30)$$

The effect of lag on NPS reduction can be regarded as an extra stage after stage 4 (i.e. addition of Se shot noise), and before stage 5 (i.e. pixel charge integration). Therefore the NPS with lag, $S_4a(f)$, is related to $S_4(f)$ (given in Eq. 4.17) by:

$$S_4a(f) = r_l S_4(f). \quad (4.31)$$

Now an example is used to demonstrate the effect of lag on NPS, where the time response of lag, $l(t)$, exhibits a single exponential decay:

$$l(t) = e^{-\frac{t}{\tau_0}}. \quad (4.32)$$

Then its temporal frequency response $R_L(\nu)$ is given by [139]:

$$R_L(\nu) = \frac{\frac{1}{\tau_0}}{\sqrt{\frac{1}{\tau_0} + (2\pi\nu)^2}}. \quad (4.33)$$

Substituting Eq. 4.33 and the expression for $R_0(\nu)$ given by Eq. 4.28 into Eq. 4.30, as shown in Appendix B, $r_l$ for this example can be obtained as:

$$r_l = 1 - \frac{\tau_0}{T_F}(1 - e^{-\frac{T_F}{\tau_0}}). \quad (4.34)$$

For lag with a decay time constant of $\tau_0 = 0.5T_F$, the image charge generated at the beginning of one frame is carried over to the subsequent frame by 13.5%, and the value for $r_l$ is 0.57. Although this noise reduction is significant, it will introduce extra blurring for images of moving objects. Therefore the noise reduction by lag has to be considered together with the blurring of moving objects.
4.3.3.2 Radiographic self-scanned a-Se detectors

In the radiographic self-scanned a-Se detector, there is an insulator layer between the top bias electrode and the a-Se, shown in cross-section in Figure 4.3 (a). The thickness of this insulator, \( d_i \), is typically one fifth of that of the a-Se, \( d_{Se} \) [60]. During x-ray exposure, holes are driven downwards to the bottom surface where they are collected on the pixel electrodes. The electrons are driven upwards and trapped at the interface between the insulator and the a-Se layer. These electrons have the same spatial distribution as the holes driven to the bottom surface. Due to their distance from the active matrix, these trapped electrons will create a negative blurred image of a smaller amplitude on the plane of the pixel electrodes (bottom a-Se surface) and be subtracted from the original sharp positive image. The MTF due to field spreading in parallel plane geometry has been solved by Schaffert [140]. For the detector configuration shown in Figure 4.3 (a), the MTF is \( T_{f1}(f) \), which can be written as [40]:

\[
T_{f1}(f) = \frac{(\epsilon_i d_{Se} + \epsilon_{Se} d_i) \sinh(2\pi f d_i)}{d_i [\epsilon_{Se} \sinh(2\pi f d_i) \cosh(2\pi f d_{Se}) + \epsilon_i \sinh(2\pi f d_{Se}) \cosh(2\pi f d_i)]},
\]

(4.35)

where \( \epsilon_i \) and \( \epsilon_{Se} \) are the dielectric constant of the insulator and a-Se, respectively. The ratio of the amplitude of the blurred image to that of the original sharp image, \( r_{f1} \), is related to the thicknesses of the insulator and a-Se by [40]:

\[
r_{f1} = \frac{\epsilon_{Se} d_i}{\epsilon_i d_{Se} + \epsilon_{Se} d_i}.
\]

(4.36)

This effect of charge blurring is stochastic and can be treated as an extra stage after stage 4, and its MTF is given by:

\[
T_{4b}(f) = 1 - r_{f1} T_{f1}(f).
\]

(4.37)

As shown in Eq. 4.37, the MTF of this radiographic detector configuration is similar to unsharp masking [141]. According to the propagation of signal and noise through stochastic blurring stage
given in Eq. 4.2, the signal spectrum, $\Phi_{4b}(f)$, is given by:

$$\Phi_{4b}(f) = T_{4b}(f)\Phi_4(f). \quad (4.38)$$

The NPS, $S_{4b}(f)$ is given by:

$$S_{4b}(f) = T_{4b}^2(f)S_4(f) + [1 - T_{4b}^2(f)]\Phi_4. \quad (4.39)$$

Another possible radiographic detector configuration is shown in Figure 4.3 (b), where the insulator is placed between the $a$-Se layer and the active matrix. In this configuration the charge image is formed at the interface between the $a$-Se layer and the insulator. The image read out on the plane of the active matrix is a blurred version of the sharp image formed at the interface. The MTF, $T_{f2}(f)$, due to this blurring effect is identical to Eq. 4.35 except that the subscripts $i$ and $Se$ are reversed throughout:

$$T_{f2}(f) = \frac{(\epsilon_i d_{Se} + \epsilon_{Se} d_i)\sinh(2\pi f d_{Se})}{d_{Se}[\epsilon_i \sinh(2\pi f d_{Se})\cosh(2\pi f d_i) + \epsilon_{Se} \sinh(2\pi f d_i)\cosh(2\pi f d_{Se})]} \quad (4.40)$$

The ratio of the amplitude of the blurred image to that of the original sharp image, $r_{f2}$, is given by:

$$r_{f2} = \frac{\epsilon_i d_{Se}}{\epsilon_i d_{Se} + \epsilon_{Se} d_i}. \quad (4.41)$$

This source of blurring is stochastic and the signal spectrum, $\Phi_{4b}(f)$, is given by:

$$\Phi_{4b}(f) = r_{f2}T_{f2}(f)\Phi_4(f). \quad (4.42)$$

The NPS, $S_{4b}(f)$, is given by:

$$S_{4b}(f) = r_{f2}^2T_{f2}^2(f)S_4(f) + [1 - r_{f2}^2T_{f2}^2(f)]S_4(f)]\Phi_4. \quad (4.43)$$

As shown in Eqs. 4.14 and 4.17, $S_4(f)$ varies with $\bar{g}^2(E)$ and $\Phi_4$ varies with $\bar{g}(E)$, therefore when $\bar{g}(E)$ is sufficiently large, $S_{4b}(f)$ given in Eq. 4.43 approaches $r_{f2}^2T_{f2}^2(f)S_4(f)$ which is the
Figure 4.3: The cross-section of the radiographic self-scanned a-Se detector: (a) the insulator is between the top electrode and the a-Se layer; (b) the insulator is between the a-Se layer and the active matrix.
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NPS of deterministic blurring [142]. Under this condition, there is negligible degradation in the presampling \(\text{DQE}(f)\) by this blurring stage. However this radiographic detector configuration, as shown in Figure 4.3 (b), can be used as a presampling filter for the high resolution image formed by \(a\)-Se and reduces aliasing in the system [143].

4.4 RESULTS AND DISCUSSION

4.4.1 Propagation of \(\text{DQE}(f)\)

Using the detector parameters for mammography shown in Table 4.1, the propagation of \(\text{DQE}(f)\) through the seven detection stages described in Section 4.3.1 were calculated. The MTF associated with the blurring stages was first calculated, and the results are shown in Figure 4.4. The MTF due to \(a\)-Se blurring, \(T_S(f)\), was calculated using the previously developed theoretical model [132]. The parameters used in the calculation were: an x-ray energy of 20 keV and an x-ray incident angle of \(\theta = 12^\circ\) which, for a 24 cm x 18 cm mammographic detector located 60 cm from the x-ray source, corresponds to the maximum x-ray incidence angle along the chest wall [144]. The blurring due to the aperture function of the pixel electrode, \(T_a(f)\), was calculated using \(a = 45 \mu m\). As shown in Figure 4.4, the total MTF of the detector is \(~50\%\) at 10 lp/mm.

Based on the detector MTF shown in Figure 4.4, \(\text{DQE}(f)\) at each detection stage was calculated using Eq. 4.25. The value for the dark current of \(a\)-Se \(I_D\) used in all calculations is \(10^{-12} \text{ A mm}^{-2}\) which can be easily attained for \(a\)-Se layers with blocking contacts [84]. The value used for the pixel electronic noise \(q_n\), unless otherwise specified, is 1000 e which has been shown to be attainable in flat-panel detectors [57]. Figure 4.5 is a surface plot showing the propagation of \(\text{DQE}(f)\) through the seven detector stages at the minimum mammographic x-ray exposure, where it is most difficult to maintain DQE at a high value. Figure 4.5 shows that the shape of \(\text{DQE}(f)\)
Figure 4.4: The modulation transfer function (MTF) of the α-Se self-scanned detector: the MTF due to oblique incidence of x-rays, $T_{Se}(f)$; the MTF due to the pixel electrode aperture function, $T_a(f)$; and the total detector MTF.
is changed by $T_{Se}(f)$ at the second stage, which is blurring due to a-Se. After the pixel charge integration at the fifth stage, there is no noticeable change in the shape of presampling $DQE(f)$ since both the signal and NPS propagate through the pixel aperture function $T_{a}(f)$. However at the sixth stage, which is the aliasing of NPS due to undersampling, the NPS becomes white and hence the output $DQE(f)$ is reduced further by $T_{a}^{2}(f)$.

Figure 4.5: The surface plot of the propagation of $DQE(f)$ through the seven detector stages. The mammographic detector parameters used in the calculation are: pixel width $\alpha=45 \mu m$; pixel electronic noise $q_n=1000 \ e$; and x-ray exposure $X=0.6 \ mR$.

In order to more clearly visualize the quantitative change in the value of DQE through the detector stages, and compare the difference in $DQE(f)$ calculated for an x-ray spectrum and monoenergetic x-rays, Figure 4.6 shows the propagation of DQE as a function of detector stage.
for both 30 kVp mammographic spectrum and 20 keV monoenergetic x-rays. Figure 4.6 (a) is a plot of DQE (extracted from the data shown in Figure 4.5) as function of detection stage at three spatial frequencies: 0, 5 and 10 lp/mm. It shows that the value of DQE is decreased by \( \sim 10\% \) at the a-Se amplification stage due to the Swank factor and the broad distribution of x-ray energies. 

\( DQE(f) \) is further degraded due to the fill factor \( F_P = 0.81 \) at the aliasing stage, and finally by the addition of pixel electronic noise \( q_n \). The addition of electronic noise by the dark current of a-Se is negligible for the values of parameters chosen in this calculation.

Shown in Figure 4.6 (b) is the DQE calculated for 20 keV monoenergetic x-rays, which is the mean x-ray energy for the 30 kVp mammographic spectrum used in Figure 4.6 (a). The propagation of \( DQE(f) \) for monoenergetic x-rays differs from that for the broad spectrum at the stage of a-Se amplification. For 20 keV monoenergetic x-rays, the decrease in DQE is \( \sim 6\% \) which is due to the Swank factor of \( A_S(E) = 0.94 \). This value is smaller than the decrease of 10\% observed for the corresponding spectrum.

The difference in DQE between an x-ray spectrum and monoenergetic x-rays increases for higher x-ray energies, such as those used for fluoroscopy and chest radiography. Shown in Figure 4.7 (a) and (b) are the DQE as a function of detector stage at 0, 1.25 and 2.5 lp/mm for a 120 kVp chest radiographic spectrum and the corresponding 50 keV monoenergetic x-rays. The calculation was performed using detector parameters and the minimum x-ray exposure for chest radiography shown in Table 4.1, and the pixel width \( \alpha \) is 195 \( \mu \)m. As shown in Figure 4.7 (a), the values for DQE decrease by 10\% at the amplification stage for the 120 kVp spectrum. However in Figure 4.7 (b), the decrease in DQE is only 2\% for 50 keV monoenergetic x-rays (where \( A_S = 98\% \)).
Figure 4.6: The DQE at the spatial frequencies of 0, 5 and 10 lp/mm as a function of detector stage for a mammographic detector at the minimum exposure: (a) calculated using a molybdenum 30 kVp spectrum; (b) calculated using 20 keV monoenergetic x-rays. The electronic noise $q_n = 1000$ e, and pixel width $a = 45 \mu m$
Figure 4.7: The DQE at the spatial frequencies of 0, 1.25 and 2.5 lp/mm as a function of detector stage for a chest radiographic detector at the minimum exposure: (a) calculated using a 120 kVp spectrum; (b) calculated using 50 keV monoenergetic x-rays. The electronic noise $q_n = 1000 \text{ e}$, and pixel width $\alpha = 195 \text{ \mu m}$
4.4.2 DQE of detectors for x-ray imaging applications

Using detector parameters shown in Table 4.1, $DQE(f)$ was calculated for the whole range of x-ray exposure used in the three x-ray imaging applications. The calculation of $DQE(f)$ was performed as a function of the following three parameters: (1) pixel fill factor $F_P$; (2) x-ray exposure; and (3) pixel electronic noise. In order to highlight the effects of these three parameters on $DQE(f)$, $T_{Se}(f) = 1$ was assumed for all three x-ray imaging applications. This is a reasonable assumption for normal x-ray incidence [132].

Figure 4.8 shows the surface plots of $DQE(f)$ calculated as a function of pixel fill factor $F_P$ (for a square pixel) for the mean exposures used in (a) mammography; (b) chest radiography; and (c) fluoroscopy. Shown in Figure 4.8, $F_P$ has two effects on $DQE(f)$: (1) it changes the value of $DQE(0)$ due to aliasing of NPS; and (2) it changes the shape of $DQE(f)$ by the square of the pixel aperture function $\text{sinc}^2(af)$. As $F_P$ decreases, $DQE(0)$ decreases linearly. At the same time, the shape of $DQE(f)$ flattens as $a$ decreases. This decrease of $DQE(0)$ as a function of $F_P$ observed from $DQE(f)$ analysis is consistent with a simple zero spatial frequency analysis: Since the x-rays incident on the detector outside the region of pixel electrodes do not give rise to charge signal detected by the pixels, they do not contribute to the final image, and hence the square of the detector output signal to noise ratio $SNR^2_{out}(0)$ is proportional to $F_P$.

Figure 4.9 shows surface plots of $DQE(f)$ calculated as a function of x-ray exposure in the range typically encountered in each x-ray imaging application. Figure 4.9 (a) shows that $DQE(f)$ retains the maximum values, i.e. x-ray quantum noise limited, for most of the exposure range used in mammography, except for the region of exposure $X < 1 \text{ mR}$ where $DQE(0)$ is reduced to 0.76 from the maximum of 0.9. As shown in Figure 4.9 (b), $DQE(f)$ retains the maximum value throughout the entire range of exposure used for chest radiography. However as shown in
Figure 4.8: The dependence of $DQE(f)$ on pixel fill factor calculated for the mean x-ray exposure level used in the three x-ray imaging applications: (a) x-ray exposure $X = 12$ mR for mammography; (b) $X = 0.3$ mR for chest radiography; and (c) $X = 0.001$ mR for fluoroscopy. The pixel electronic noise $q_n = 1000$ e.
Figure 4.9 (c) for fluoroscopy, the detector only reaches the maximum $DQE(f)$ as the exposure approaches the high end of the x-ray exposure range. Therefore the detector falls short of quantum noise limitation especially below the mean exposure for fluoroscopy.

Figure 4.10 shows surface plots of $DQE(f)$ calculated as a function of pixel electronic noise $q_n$ at the minimum x-ray exposure for each x-ray imaging application where the detector is most sensitive to electronic noise. As shown in Figure 4.10 (a) for mammography, $DQE(0)$ can be maintained close to the maximum value of 0.9 if $q_n < 1000$ e is satisfied. As shown in Figure 4.10 (b) for chest radiography, $DQE(0)$ is at the maximum value of 0.5 for $q_n$ up to 3000 e. However, as shown in Figure 4.10 (c) for fluoroscopy, it is practically impossible to achieve the maximum $DQE(0)$ value of 0.7 at the minimum exposure of 0.0001 mR, since the value for $q_n$ would have to be less than two hundred electrons. However $DQE(0)$ can be kept above half the maximum value if $q_n$ is below 600 electrons, which will provide a good signal to noise performance at the lowest exposure. This has been shown to be possible by using low noise charge amplifiers [57,58]. The results in Figure 4.10 indicates that chest radiography is the application where the detector is most tolerant to electronic noise while fluoroscopy is least tolerant to electronic noise.

### 4.5 MAXIMIZING PIXEL FILL FACTOR

As shown in Figure 4.8, $DQE(0)$ of the detector decreases linearly as a function of pixel fill factor $F_p$. This arises from the lack of charge collection capability of the active matrix for image charges that land in the gap between pixel electrodes, which results in loss of x-ray detection due to the very high resolution of a-Se. The effect of $F_p$ on $DQE(f)$ is therefore different from that in the flat-panel detectors using the indirect conversion method. In the indirect approach, due to the blurring caused by phosphor screens, the loss of light photons due to $F_p$ will not usually result in
Figure 4.9: The $DQE(f)$ calculated for the whole x-ray exposure range used in: (a) mammography; (b) chest radiography, and (c) fluoroscopy. The pixel fill factor $F_P$ was assumed unity and the pixel electronic noise $q_n = 1000$ e.
Figure 4.10: The $DQE(f)$ as a function of added pixel electronic noise $q_n$ for the minimum x-ray exposure in the three x-ray imaging applications: (a) $X = 0.6$ mR for mammography; (b) $X = 0.03$ mR for chest radiography; and (c) $X = 0.0001$ mR for fluoroscopy. The fill factor $F_p$ was assumed unity.
a complete loss of x-rays [143]. Therefore while gaining the advantages of higher image resolution and simpler active matrix structure using the direct conversion method, increasing $F_P$ becomes more important in order to maximize the DQE.

In practical design of active matrices, a high pixel fill factor $F_P$ can be achieved by building the pixel electrodes above the rest of the active matrix structure (e.g. TFTs, gate and data lines), as shown in Figure 4.11. This method of building pixel electrodes is often referred to as a *mushroom* structure, and has been used in a radiographic self-scanned $a$-Se detector [60]. In this case the pixel electrodes occupy most of the pixel area and separated by a minimum gap (5 μm for the current design rules). With this design, $F_P$ can be higher than 95% for the detector parameters proposed for chest radiography and fluoroscopy, which is quite satisfactory. For mammography with pixel pitch of 50 μm, $F_P$ will be 81% where improvement is still desirable.

Further improvement of fill factor may be possible by manipulating the electric field in $a$-Se above the gap between the pixel electrodes so that essentially all the charge carriers will be driven to the pixel electrodes, as shown in Figure 4.11. Using this method, an *effective* fill factor $F_E$ that is much higher than the geometric fill factor $F_P$ can be achieved. This can be accomplished by *conditioning* the detector. During conditioning, a uniform x-ray (or light) exposure will generate holes and guide them to the bottom surface of the $a$-Se layer. At the interface between the $a$-Se and the insulator (usually silicon dioxide) in the gap between pixel electrodes, some holes will be permanently trapped. When this trapping process reaches equilibrium, the electric field in the $a$-Se layer above the gaps will be reduced to zero thereby forcing holes generated subsequently by radiation to be driven to the pixel electrodes, as shown in Figure 4.11. This approach can possibly increase $F_E$ to unity. In order for this concept to work properly, the design of the insulator layer between pixel electrodes has to be optimized in terms of thickness, dielectric strength and interface
trapping properties.

Figure 4.11: The cross-section of a conditioned self-scanned a-Se detector. The trapped holes at the a-Se / insulator interface between pixel electrodes force the electric field line to be diverted to pixel electrodes, thereby increasing the effective pixel fill factor.

The effect of pixel fill-factor $F_p$ on $DQE(0)$ can also be reduced using the radiographic detector configuration with an insulator between a-Se and the active matrix, as shown in Figure 4.3 (b). The extra blurring stage (introduced by the distance between the image formation plane and the readout plane) acts as a presampling filter which reduces aliasing, and hence the loss of image charge due to $F_p < 1$ will not lead to complete loss of x-rays. Unfortunately, this approach cannot be used for real-time x-ray imaging.

4.6 CONCLUSIONS

The spatial frequency dependent DQE of the real-time, flat-panel self-scanned a-Se detector has been analyzed based on a cascaded linear systems model. It was found that the aliased NPS of
the detector is white and the shape of the $DQE(f)$ follows that of the square of the MTF of the detector. The value for $DQE(0)$ is linearly dependent on the pixel fill factor $F_p$, and the added electronic noise reduces DQE at low x-ray exposure levels, especially for fluoroscopy. With typical detector parameters, the $DQE(0)$ of the self-scanned $\alpha$-Se detector is approximately 0.5, 0.7 and 0.9 for x-ray imaging applications of chest radiography, fluoroscopy and mammography, respectively.
Chapter 5

Construction and Evaluation of a
Prototype Real-time X-ray Detector

5.1 INTRODUCTION

The concept of making a flat-panel, real-time, digital x-ray imaging detector using self-scanned readout of a layer of x-ray photoconductor has previously been proposed by us [67,68]. As shown in Figure 1.9, x-ray detection is achieved by a photoconductive layer of amorphous selenium (a-Se) deposited on the surface of a large area, two dimensional array of thin film transistors (TFTs), or active matrix, which is used for readout. An electric field $E_{Se}$ is applied across the a-Se layer through a bias electrode deposited on the top surface of the a-Se layer. When x-rays are absorbed in a-Se, electrons and holes are created in pairs and drawn to the appropriate surfaces by $E_{Se}$. With positive potential on the bias electrode, holes are driven to the active matrix and collected by the pixel electrodes where they are stored on individual capacitors at each pixel. The charge

\footnote{This chapter has been submitted as a paper to Med. Phys., entitled "Digital radiology using active matrix readout of amorphous selenium: Construction and evaluation of a prototype real-time detector."}
image is then read out in a self-scanned manner using the active matrix. This is accomplished by turning on the TFTs one row at a time by means of a common gate line. The image charge on the pixels of the selected row is transferred to the corresponding vertical (data) lines and integrated by the charge amplifiers. A multiplexer then converts the amplified charge signals to a serial output for subsequent digitization. The image acquisition of the detector can be in real-time, i.e. 30 frames per second, therefore can potentially be applied in both fluoroscopy and radiography.

Self-scanned detectors based on indirect conversion of x-rays using scintillator (e.g. CsI) and an active matrix array of TFTs (or switching diodes) and photodiodes have been proposed by various investigators [56-59]. The advantages of the direct over the indirect approach are: (1) the higher image resolution provided by electrostatic image formation, and (2) simpler active matrix structure similar to that used for liquid crystal displays (LCDs), compared to a complex array structure with integrated photodiodes at every pixel required by the indirect approach. Recently, a variation of the self-scanned a-Se detector was proposed in which a dielectric layer was added between the a-Se and the top bias electrode [60]. Images obtained with a prototype of this concept have demonstrated high image resolution [102]. However, due to the procedure necessary to remove trapped charge at the interface between the a-Se and the dielectric layer, this variation is restricted to radiography and therefore cannot be applied to real-time x-ray imaging.

In this chapter, the investigation of the feasibility of a real-time, self-scanned a-Se detector will be presented. First, the design and construction of a prototype detector will be described. Secondly, the methods for evaluating the imaging performance of the prototype, e.g. x-ray sensitivity, modulation transfer function (MTF), noise power spectrum (NPS) and the frequency dependent detective quantum efficiency (DQE), will be described and finally, the experimental results will be presented and compared with the results of our theoretical analysis [65]. Digital radiographs
obtained using the prototype will also be shown.

5.2 MATERIALS: PROTOTYPE CONSTRUCTION

5.2.1 Sensor construction

5.2.1.1 Active matrix design

The active matrix for the prototype detector is made with CdSe TFTs and built on glass substrates using large area integrated circuit technology through collaboration with Litton Systems Canada Limited. Its design is based on an existing active matrix for LCD applications, which has 360 \times 480 pixels with 160 \mu m pixel pitch. Figure 5.1 is a micrograph showing a top view of four pixels of the active matrix built for the prototype. The horizontal lines are the gate lines for scanning control and the vertical lines are the data lines for transferring image charge from pixels to external charge amplifiers. The TFT is built on top of the gate lines at the corner of each pixel, with channel width and length of 20 and 50 \mu m, respectively. The pixel storage capacitor is formed by overlapping the pixel electrode with the adjacent gate line, and the value for each storage capacitor, \( C_p \), is 0.7 pF. The pixel geometric fill factor, \textit{i.e.} the ratio between the area of each pixel electrode and the pixel pitch squared, is 70 %.

Figure 5.2 is a schematic diagram showing the cross-sectional view of one pixel on the imaging active matrix. The active matrix sensor array was made using the same photolithographic processing as that used in making LCD active matrix arrays, and six photolithographic masks were used [145]. The gate metal (chromium) was first deposited on the glass substrate and delineated, followed by the deposition of a 360 nm thick gate insulator (SiO\(_2\)) layer using plasma enhanced chemical vapor deposition (PECVD). Then the 50 nm thick CdSe semiconductor layer was deposited using thermal evaporation and subsequently recrystallized under moderate temperature to form a
Figure 5.1: Micrograph of four pixels of the active matrix made for the prototype detector. The pixel pitch is 160 μm.
polycrystalline structure. On top of the CdSe is a passivation layer of SiO₂. Then the drain and source metal was formed and connected to the CdSe layer through etched vias in the passivation layer. At the same time as the drain and source electrode formation, an additional gate (top gate) was formed which is connected to the bottom gate through a via in both the top and the bottom insulator layers. The additional top gate is the main difference between the TFTs used in the imaging active matrix and that used in LCDs, and its purpose is to shield the TFT channel from the electric field in the a-Se layer.

![Diagram of pixel structure](image)

Figure 5.2: The cross-sectional view of one pixel of the imaging active matrix.

5.2.1.2 TFT characteristics

CdSe TFTs operate as enhanced mode n-type metal-oxide-semiconductor field effect transistors (MOSFETs). The characteristics of dual-gate TFTs were studied using individual test devices designed and fabricated along the edge of the imaging active matrix. The geometry of the test devices is identical to those on the imaging active matrix, but were made with isolated top and bottom gate and separate connections to all electrodes. The transfer characteristic curves of such a dual-gate TFT measured with drain-source voltage \( V_{DS} = 0.5 \) V, and under two gate bias conditions are shown in Figure 5.3. One curve shows the channel current \( I_D \) as a function of the voltage on the bottom gate, \( V_{BG} \), with the top and bottom gates connected, i.e. \( V_{TG} = V_{BG} \). This is the
operational condition for the dual-gate TFTs on the prototype detector. Also shown is the transfer characteristic curve with the top gate floating, which simulates the characteristics of an ordinary single gate TFT structure. The comparison of the dual-gate and the single gate configurations shows that the former provides a larger ON current and a steeper slope in the subthreshold region of the transfer characteristics than the latter, and therefore the switching characteristics of a dual-gate TFT is slightly better than that of the standard (single gate) TFT.

![Transfer characteristic curve](image)

Figure 5.3: The transfer characteristic curve of a dual-gate TFT with the top gate connected to the bottom gate, compared with that with the top gate floating (equivalent to a single gate TFT).

5.2.1.3 Selenium deposition

A 300 μm thick a-Se layer, i.e. \(d_{Se} = 300 \mu m\), was thermally evaporated onto the surface of the imaging active matrix through collaboration with Noranda Advanced Materials. The a-Se alloy and the deposition conditions are similar to those used in making Xeroradiographic plates [93].
After a-Se deposition, a uniform indium (In) metal layer was thermally evaporated to form the top high voltage bias electrode.

5.2.2 Image acquisition system

The active matrix x-ray sensor was connected to external electronics for image acquisition. Figure 5.4 is a block diagram showing the functionality of each sub-unit of the image acquisition system. The integrated circuits (ICs) containing the gate scanning control circuit and the charge amplifiers were wire-bonded to the gate and data lines of the active matrix. The array scanning and the image acquisition sequence are controlled by a personal computer (PC) based image acquisition system.

The active matrix sensor and its associated peripheral electronics on a printed circuit board (PCB) are shown in Figure 5.5. The gate scanning control circuit is on the left-hand side of the active matrix sensor and consists of three specialized LCD drivers, each with 120 output lines. A gate operational voltage of -5 V / 15 V (OFF / ON) was chosen which ensures an on resistance $R_{on}$ of 1 MΩ and an off resistance $R_{off}$ of $10^{13}$ Ω for the TFTs on the active matrix. The pixel readout time constant is given by the expression $\tau_{on} = R_{on}C_P$, and for the parameters of the prototype detector, $\tau_{on}$ corresponds to 0.7 μs, which permits a real-time rate without readout lag [84].

The data lines of the active matrix sensor were connected to integrated charge amplifier / multiplexer circuits (MB series products manufactured by EG&G Reticon) [146], each of which contains 64 input channels and are shown in Figure 5.5 below the active matrix sensor. Each charge amplifier has a resetable feedback capacitor $C_f$ of 16 pF, which results in a nominal gain of 100 electrons / μV, and it also incorporates correlated double sampling (CDS). Figure 5.6 (a) is a schematic diagram showing the charge amplifier and the CDS circuit, which consists of two parallel sampling circuits at the output of each charge amplifier. Figure 5.6(b) is a diagram showing the relative timing of the gate control pulse for the TFTs, the reset pulse for the feedback capacitor $C_f$, and...
CHAPTER 5. CONSTRUCTION AND EVALUATION OF A PROTOTYPE

Figure 5.4: A schematic block diagram of the functionality of each subunit of the prototype detector.

Figure 5.5: A photograph of the active matrix / selenium sensor and the peripheral electronic circuits for the prototype detector.
the voltage output of the charge amplifiers, and the two sampling pulses associated with the CDS. The first sample is taken shortly after resetting the feedback capacitor $C_f$ of each charge amplifier and before the TFT is turned on. The second sample is taken many ($\geq 5$) pixel time constants after the TFT is turned on so that the image charge is completely transferred from the pixel capacitance to $C_f$ of the charge amplifier. The multiplexer converts the parallel output of the two sampled signals to a serial output at a rate of 1 MHz. Then the two video outputs associated with the two samples were fed to an external differential voltage amplifier to achieve the CDS, i.e. to subtract the value obtained from the first sample from that of the second sample. The implementation of CDS eliminates low frequency components of the electronic noise arising in the charge amplifiers and more importantly, it eliminates the thermal noise associated with resetting the feedback capacitor $C_f$ which arises from the thermal noise of the resistance of the reset switch. While $C_f$ is reset by turning on the electronic switch in the feedback loop as shown in Figure 5.6, there is an uncertainty associated with the voltage across $C_f$ due to the thermal noise of the resistance of the switch. After the reset switch is turned off, the uncertainty of the amount of charge left on the capacitor $C_f$ is given by [147]:

$$q_{rn} = \frac{1}{e} \sqrt{kTC_f}$$  \hspace{1cm} (5.1)

where $k$ is Boltzmann’s constant and $T$ is the absolute temperature. With $C_f = 16$ pF, $q_{rn}$ is $\sim 1600$ electrons, which is larger than any other fundamental sources of electronic noise in the self-scanned α-Se detector [84]. However after incorporation of CDS, the exact value of $q_{rn}$ after each reset can be sampled and subtracted, and therefore be eliminated from the final signal.

The final output signal, after a further voltage amplification gain of 20, was digitized using an individual analog-to-digital converter (ADC) for each multiplexer. Each ADC digitizes at 1 M samples / second with a digitization depth of 12 bits. Therefore the whole image acquisition system
Figure 5.6: The implementation of correlated double sampling (CDS): (a) The circuit diagram of the charge sensitive amplifier and the two samples associated with the CDS; (b) The relative timing of the two samples, the charge readout and the reset of the charge amplifier.
is capable of producing real-time images for the instrumented 360 x 128 pixels of the prototype
detector. A real-time offset and gain correction was incorporated using image processing hardware
after the ADC and the resulting images were either displayed on a monitor or transferred to the
host PC.

5.3 THEORY AND METHODS

The imaging performance of the prototype self-scanned a-Se detector, including x-ray sensitivity,
presampling MTF, NPS and DQE, were measured.

5.3.1 X-ray sensitivity

The x-ray sensitivity of the prototype detector can be predicted from the x-ray energy absorption
and the conversion gain $g$ in a-Se, i.e. the number of charges released per x-ray absorbed in a-Se.
The x-ray energy absorption is proportional to the quantum efficiency $\eta$ which is a function of both
the x-ray energy $E$ and the thickness of a-Se, $d_{Se}$. The energy dependent $\eta$ is given by:

$$\eta(E) = 1 - e^{-\mu(E)d_{Se}},$$

(5.2)

where $\mu(E)$ is the x-ray linear attenuation coefficient of a-Se. In order to count for the fact that
not all attenuated x-ray energy is absorbed by a-Se, the quantum efficiency $\eta$ is multiplied by the
ratio of $\mu_{ab}(E)$, which is the energy dependent linear absorption coefficient of a-Se, and $\mu(E)$ in
order to determine the x-ray energy absorption. The conversion gain $g$ is inversely proportional to
the electric field dependent energy $W_\pm$ required to free an electron hole pair. Therefore the signal
charge generated by each pixel of the detector, $q_s$, is given by [84]:

$$q_s = \int_0^{E_{max}} \frac{\bar{\delta}(E)\eta(E)A_P \mu_{ab}(E)}{W_\pm \mu(E)} E \, dE,$$

(5.3)
where $\mathcal{F}_0(E)$ is the mean number of x-ray quanta incident per unit area with energy $E$, $E_{\text{max}}$ is the maximum energy of the incident x-rays, $A_P$ is the pixel area, and $F_P$ is the pixel fill-factor.

The x-ray system used in the experiments consists of a generator (Electromed) with a stable DC high voltage output, and an x-ray tube with a rotating tungsten anode. An x-ray spectrum of 70 kVp and a filtration of 2 mm aluminum were used during the x-ray sensitivity measurement. The source detector distance (SDD) was 115 cm and the exposure was measured using a Keithley dosimeter (model 35040) and ion chamber (model 96035).

The x-ray sensitivity was measured with the detector operated in fluoroscopic mode (30 frames/second). The pixel response was first measured as a function of x-ray exposure, which was changed by adjusting the x-ray tube current. At each exposure setting, 180 subsequent image frames were acquired, with the radiation turned on half way through the image acquisition. The image frames acquired prior to irradiation were averaged to form the offset (dark) image, which was then subtracted from each frame of image data acquired during the x-ray exposure. In order to eliminate the effect of gain nonuniformity, the offset-corrected image data from an area of 20 x 20 responding pixels were averaged to obtain the x-ray response of the detector at each exposure level. Then the pixel response as a function of x-ray exposure was fitted to a straight line in order to obtain the x-ray sensitivity of the prototype detector.

The x-ray response was measured as a function of the electric field across the a-Se layer, $E_{Se}$. At each $E_{Se}$ setting, the detector was first stabilized in the dark for approximately 10 minutes in order to permit transient changes in dark current to equilibrate before making sensitivity measurements.
5.3.2 Presampling modulation transfer function

5.3.2.1 Theoretical prediction of MTF

The modulation transfer function of an imaging detector could be obtained as the modulus of the Fourier transform of the line spread function (LSF), which is often derived from a slit image. The MTF has been used to characterize the resolution of conventional x-ray imaging systems, such as screen-films and x-ray image intensifier / video systems [148]. For a digital imaging detector, the LSF is sampled. If the presampling LSF does not have frequency component higher than the Nyquist frequency $f_{NY}$ of the sampling, the Fourier transform of the digital LSF will result in the detector MTF that has the same meaning as in traditional analog detectors. However if the presampling LSF of a digital imaging detector has frequency components higher than $f_{NY}$, the digital detector is undersampled and aliasing will occur. Since a-Se has very high intrinsic resolution, the self-scanned a-Se is inherently an undersampled digital imaging detector. The effects of various digital detector parameters, such as sampling aperture and sampling distance, on the MTF have been investigated by Giger et al. [126] As summarized by Dobbins, an undersampled digital detector is not a shift invariant system, the detector output signal varies depending on the position of the input signal (such as a slit) relative to the position of the pixels [128]. Therefore measurement and interpretation of the digital MTF of an undersampled detector become difficult.

The presampling MTF of a digital detector is the analog MTF before sampling occurs. It provides a means for understanding the effects of different detector stages on the resolution of the system. The usefulness of presampling MTF has been demonstrated by measurements on various digital x-ray imaging detectors, such as a slot-scan digital mammographic detector using a phosphor screen coupled to charge coupled device (CCD) imaging arrays through fiber optics [149], a photostimulable phosphor detector [150], and an indirect flat-panel detector using phosphor screen
and a-Si:H TFT/photodiode arrays [151].

For the self-scanned a-Se detector the presampling MTF can provide information on the factors affecting the image resolution before sampling occurs, such as the resolution of the a-Se layer and the blurring due to the aperture of the pixel electrode (aperture function). The presampling MTF of the self-scanned a-Se detector, \( T(f) \), is determined by the MTF of the a-Se layer, \( T_{Se}(f) \), and the MTF of the aperture of the pixel electrode, \( T_a(f) \):

\[
T(f) = T_{Se}(f) T_a(f).
\]  

(5.4)

The MTF of an a-Se layer has been established theoretically [132], and the results revealed that the major source of image blurring at diagnostic energies is the oblique entrance of x-rays. For low energy x-ray photons, such as those used in mammography, the blurring due to \( k \) fluorescence reabsorption is also important. For higher diagnostic x-ray energies (e.g. 50 keV) where the effect of \( k \) fluorescent reabsorption is negligible, the range of primary photoelectrons can be important. However, for spatial frequencies \( \leq 10 \, \text{lp/mm} \), the drop of MTF due to the range of primary photoelectrons is \(<5\%\) [132], thus for practical purposes \( T_{Se}(f) = 1 \) can be assumed when x-rays are at normal incidence to the a-Se detector. The aperture function of a square pixel with pixel width \( a \) is \( T_a(f) = \text{sinc}(af) \). However with our current active matrix design, the pixel electrode has a more complex shape as shown in Figure 5.7(a). The parameters \( W_n \) and \( H_n \) (\( n = 1-4 \)) are the width and height, respectively, of different sections of the pixel electrode. The LSF of such pixel electrode, for example in the gate line direction, can be determined from the response of the pixel to an assumed scanning slit oriented parallel to the data lines and scanned along the gate lines [152]. Shown in Figure 5.7(b) is the corresponding LSF. Each individual rectangular section of the LSF corresponds to a sinc function in the spatial frequency domain. Using the shift theorem for the Fourier transform [153], the aperture function \( T_a(f) \) can be obtained from the Fourier transform
of each individual section using:

\[
T_a(f) = \left| \frac{W_1 H_1}{A_P} \text{sinc}(W_1 f) + \frac{W_2 H_2}{A_P} e^{-j\pi(W_1+W_2)f} \text{sinc}(W_2 f) \right.
+ \left. \frac{W_3 H_3}{A_P} e^{-j\pi(W_1+2W_2+W_3)f} \text{sinc}(W_3 f) + \frac{W_4 H_4}{A_P} e^{-j\pi(W_1+2W_2+2W_3+W_4)f} \text{sinc}(W_4 f) \right|.
\]  

5.3.2.2 Experimental measurement of MTF

The slanted edge method was used to measure the presampling MTF of the prototype. This approach allows us to obtain an oversampled edge spread function (ESF) [154]. A 450 \(\mu\)m thick tantalum edge was placed at an angle of two degrees from the data line direction of the active matrix, and 2 cm from the surface of the a-Se top bias electrode. An x-ray image of the edge was taken with an x-ray spectrum of 40 kVp, and a SDD of 115 cm. The position of the edge on the image was determined by a straight line fitting for the pixels that crossed the edge. According to the percentage of pixel area crossed by the edge, the image data for each line of the edge image were placed into ten different bins. The image data from the ten bins were interleaved to provide a ten times oversampled ESF. Then the data for pixels away from the edge, i.e. the regions of the ESF curve that are flat, were fitted to a straight line in order to reduce the noise for subsequent differentiation of the data. The oversampled LSF of the detector was then obtained by differentiation of the ESF, and the presampling MTF is obtained as the Fourier transform of the LSF.

5.3.3 Noise power spectrum

Since self-scanned a-Se detectors are undersampled, the noise power spectrum available to us for measurement is the aliased NPS [128]. It has been shown that the aliased NPS for a self-scanned a-Se detector is white [65], which can be more intuitively understood as due to the complete isolation
Figure 5.7: The aperture function of an asymmetrical pixel electrode: (a) Schematic showing the shape and geometry of the pixel electrode used in the prototype detector; (b) The corresponding line spread function of the pixel electrode in the horizontal (gate line) direction.
of neighboring pixels resulted from the essentially perfect resolution of α-Se.

The NPS was measured using the synthesized scanning slit method, which provides a central slice through the two dimensional NPS [127,155]. The NPS was measured with an SDD of 115 cm and an x-ray spectrum of 70 kVp with 2 mm aluminum filtration. Four exposure levels were used, ranging from 0.94 mR to 3.77 mR, all at an applied electric field of \( E_{5e} = 5 \text{V/μm} \). At each x-ray exposure setting, an offset image and a bright image were first obtained by acquiring and averaging 100 frames of images under dark and flood-field exposure. The gain correction factor for each pixel was then obtained by subtracting the offset image from the bright image. Then the NPS was measured using an image sequence of 100 frames acquired under uniform x-ray exposure. For each image, a region with 280 gate lines and 100 data lines was chosen for the determination of NPS. The images were corrected for gain and offset nonuniformity, and defective pixels were replaced with image values of neighboring pixels. A pixel is regarded as defective when its offset-corrected image value is less than one third of the average of all the pixels. From each two-dimensional, corrected digital image data, slit images were synthesized by averaging the data along one direction for a given slit length. Using the gate line direction NPS as an example, a slit length of 22.4 mm was used, i.e. 140 lines of data were averaged to form one slit image. This slit length ensures that the resulting NPS corresponds to a thin central slice of the two-dimensional NPS. As a result, 200 synthesized slit images (one dimensional noise data) were obtained from the image sequence of 100 frames. Each slit image was adjusted to have zero mean by subtracting the mean of each slit image. Then the modulus of the Fourier transform of each slit image (100 points) was computed and squared to form one NPS sample, \( S_i(f) \), which was then normalized by multiplying by the length of the slit and dividing by the width of the slit image [155]. The 200 NPS samples were averaged to obtain the final NPS, \( S(f) \).
In order to establish the degree of frame to frame correlation (i.e. lag) so as to determine its effect on NPS measurement, we also measured the temporal autocorrelation function of the detector using acquired image sequences. The temporal autocorrelation function was computed using 100 consecutive image frames acquired at a rate of 30 frames / second, and the final result was obtained by averaging the measurements from 25 pixels.

5.3.4 Detective quantum efficiency

5.3.4.1 DQE analysis using a cascaded systems model

A cascaded linear systems model has been developed in order to analyze the spatial frequency dependent DQE of the self-scanned a-Se detector [65]. In this model, the detector is divided into seven stages: (1) x-ray attenuation by a-Se; (2) stochastic blurring due to the resolution of a-Se; (3) gain stage associated with conversion from x-rays to electron-hole pairs by a-Se; (4) addition of electronic noise due to dark current shot noise in a-Se; (5) integration of image charge on pixel electrodes; (6) aliasing due to undersampling; (7) addition of electronic noise due to the TFTs and the charge amplifiers. The propagation of signal and noise power spectra through each stage can be calculated and the overall \( DQE(f) \) for the complete detector can be written as:

\[
DQE(f) = \frac{F_P [\int_0^{E_{\text{max}}} \frac{\mathcal{P}(E)}{\mathcal{A}_S(E)} \eta(E) dE] T_a^2(f) T_a^2(f)}{\mathcal{A}_S \int_0^{E_{\text{max}}} \frac{\mathcal{P}(E)}{\mathcal{A}_S(E)} \eta(E) dE + S_d(f) + F_P S_a(f)}.
\]  

(5.6)

where \( S_d(f) \) and \( S_a(f) \) are the white noise power spectra associated with the dark current shot noise and the readout electronic noise, respectively. The former can be determined from the dark current \( I_d \) of a-Se and the latter from the pixel rms electronic noise \( q_n \). The Swank factor \( A_S \) describes the noise associated with the conversion gain of the a-Se layer. The prototype detector and x-ray parameters required to calculate \( DQE(f) \) using Eq. 5.6 are summarized in Table 5.1.
## Table 5.1: Detector parameters used in the theoretical calculation of the aperture function and the detective quantum efficiency (DQE) of the prototype self-scanned α-Se detector.

<table>
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<th>Description</th>
<th>parameter</th>
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<tr>
<td>pixel fill factor</td>
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<td>$W_2$</td>
<td>$\mu$m</td>
<td>52</td>
</tr>
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<td>155</td>
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<td>$\mu$m</td>
<td>93</td>
</tr>
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<td>kVp</td>
<td></td>
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</tr>
<tr>
<td>x-ray beam filtration</td>
<td>mm Al</td>
<td></td>
<td>2</td>
</tr>
<tr>
<td>electric field in α-Se</td>
<td>$E_{Se}$</td>
<td>V/$\mu$m</td>
<td>5</td>
</tr>
<tr>
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<td>$\theta$</td>
<td>degree</td>
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<td>energy needed to create an e/h pair</td>
<td>$W_e$</td>
<td>eV</td>
<td>87</td>
</tr>
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<td>added pixel electronic noise (rms)</td>
<td>$q_n$</td>
<td>electrons</td>
<td>18,000</td>
</tr>
<tr>
<td>selenium dark current</td>
<td>$I_d$</td>
<td>A mm$^{-2}$</td>
<td>$10^{-10}$</td>
</tr>
<tr>
<td>input x-ray fluence</td>
<td>$\Phi$</td>
<td>quanta mm$^{-2}$ mR$^{-1}$</td>
<td>$1.47 \times 10^5$</td>
</tr>
</tbody>
</table>
5.3.4.2 Experimental determination of DQE

The experimental $DQE(f)$ of the prototype detector can be computed from the measured x-ray sensitivity, the presampling MTF, $T(f)$, and the noise power spectrum $S(f)$ using [155]:

$$DQE(f) = \frac{k^2 T^2(f)}{\overline{f}_0 S(f)},$$

(5.7)

where $k$ is the pixel x-ray response (in electrons) at a given x-ray exposure, and $\overline{f}_0$ is the mean incident x-ray fluence integrated over the entire energy spectrum (in quanta mm$^{-2}$) at a given x-ray exposure. The value for $k$ can be obtained by the product of the pixel x-ray sensitivity (in electrons mR$^{-1}$) and the x-ray exposure (in mR).

5.3.5 Radiographs of phantoms

In order to examine the overall resolution and noise performance of the prototype detector, images of an x-ray bar pattern and a human hand phantom were taken at radiographic exposure levels. The instrumented section of the prototype is approximately 5 cm x 2 cm (360 gate lines x 128 data lines). When imaging the hand phantom, 14 subimages were acquired with the phantom moving seven approximately equal steps along two rows. Then the subimages were stitched offline to produce an image of the whole hand.

5.4 RESULTS AND DISCUSSION

5.4.1 X-ray sensitivity

The x-ray sensitivity can be calculated using Eq. 5.3. At $E_{Se} = 5$ V/µm, the value for $W_\pm$ in $\alpha$-Se is 87±2 eV according to measurements made using pulse height spectroscopy [133]. With $d_{Se} = 300$ µm, and a fill-factor $F_p$ of 70%, the theoretical calculation (Eq. 5.3) predicts a pixel
x-ray sensitivity of $9.9 \pm 0.2 \times 10^5$ electrons mR$^{-1}$. Figure 5.8 shows the measured x-ray signal per pixel of the prototype as a function of radiation exposure level. The best linear fitting of the measured data results in a pixel x-ray sensitivity of $9.7 \times 10^5$ electrons mR$^{-1}$. This is in agreement with the theoretical calculation.

![Figure 5.8](image)

**Figure 5.8:** The measured pixel response (in electrons) of the prototype detector as a function of x-ray exposure.

Figure 5.9 (a) shows the pixel x-ray signal, measured at an x-ray exposure level of 0.94 mR, as a function of $E_{Se}$ ranging from 2 to 7 V/μm. Previous studies [133] showed that the energy required to generate an electron-hole pair in α-Se, $W_\pm$, is related to $E_{Se}$ through the empirical expression:

$$W_\pm \propto E_{Se}^{-\gamma},$$ (5.8)

where $\gamma$ is an experimentally determined constant. The inverse of the measured signal shown in Figure 5.9 (a), which as shown in Eq. 5.3 is proportional to $W_\pm$, is plotted as a function of $E_{Se}$ in
Figure 5.9 (b) on a logarithmic scale. The best straight line fitting of the data predicts a value for $\gamma$ of 0.68. This is within the range of values found previously ($\gamma = 0.8 \pm 0.2$) [133].

5.4.2 Presampling modulation transfer function

The MTF due to pixel aperture blurring, $T_a(f)$, in the gate line direction was calculated using Eq. 5.6 and the geometry for different sections of the pixel electrode summarized in Table 5.1, and the result is shown in Figure 5.10. Because of the complicated shape of the pixel electrode, $T_a(f)$ shown in Figure 5.10 is not a sinc function as a regular square (or rectangular) pixel electrode. However, it shows that $T_a(f)$ has a first minimum at the spatial frequency of $f = 8$ lp/mm.

The measured presampling MTF in the gate line direction of the prototype detector using the slanted edge method is also shown in Figure 5.10. The presampling MTF of the entire detector closely follows the MTF of the aperture function of the pixel electrode for spatial frequencies up to the first minimum, which indicates that the a-Se layer has very little degradation effect on the resolution of the self-scanned detector up to 8 lp/mm.

5.4.3 Noise power spectrum

The dark noise power spectrum as well as the noise power spectra at four different x-ray exposures measured from the prototype detector are shown in Figure 5.11. The NPS is independent of the spatial frequency, i.e. white as expected, and the quality of fit to a white spectrum is demonstrated using a representative result (obtained at 0.94 mR). The noise power density measured in the dark is $10.0 \text{ ke}^2\text{mm}^2$, which corresponds to a pixel noise of $q_n = 18,000$ e rms. Figure 5.11 indicates an increase of noise power as a function of x-ray exposure. The increase is expected to be linear since the x-ray quantum noise is proportional to the number of incident x-ray quanta. Thus plotted in Figure 5.12 is the noise power density (obtained from the white noise fitting shown in Figure 5.11)
CHAPTER 5. CONSTRUCTION AND EVALUATION OF A PROTOTYPE

Figure 5.9: The pixel x-ray sensitivity of the prototype detector as a function of the electric field, $E_{Se}$, in a-Se: (a) The pixel signal as a function of $E_{Se}$ at an x-ray exposure of 0.94 mR; (b) The inverse of the pixel signal plotted as a function of $E_{Se}$ in log-log scale and the best straight line fitting for the data.
Figure 5.10: The calculated pixel aperture function (shown with a solid line), in comparison with the measured presampling modulation transfer function of the prototype detector (shown as filled squares).
as a function of x-ray exposure. It shows that the measured noise power density increases linearly as a function of x-ray exposure with an intercept with the ordinate corresponding to the electronic noise power. Figure 5.12 indicates that the electronic noise is comparable to the x-ray quantum noise at the lowest x-ray exposure levels, e.g. 0.94 mR, used in the measurement. Therefore it is not far from being adequate for those radiographic imaging applications where mean exposures of the order of 1 mR are used. However in order to use the detector for lower exposure examinations, e.g. in fluoroscopic imaging applications where the mean x-ray exposure to the detector is 1 μR, the electronic noise needs to be reduced by more than an order of magnitude to ~1000 e [65,84].

Figure 5.11: The noise power spectra as a function of spatial frequency for the prototype detector measured at exposures of 0 (dark), 0.94, 1.42, 2.36 and 3.77 mR. (for clarity, data is plotted only for 0.94 mR)

Figure 5.13 shows the measured temporal autocorrelation function of the detector, i.e. the normalized autocorrelation as a function of frame number. The result shows that the autocorrelation decreases to the noise level for a shift of one frame, demonstrating that there is no significant
Figure 5.12: The noise power density of the measured white NPS as a function of the x-ray exposure to the prototype detector.

temporal correlation. Therefore it can be concluded that the effect of image lag on the measurement of NPS is negligible.

5.4.4 Detective quantum efficiency

Shown in Figure 5.14 is the result of the theoretical DQE analysis obtained using the cascaded linear systems model and the parameters summarized in Table 5.1. In this surface plot, $DQE(f)$ is shown up to the Nyquist frequency (i.e. 3.1 lp/mm) as a function of x-ray exposure (ranging from 0.5 to 4 mR). $DQE(f)$ increases as a function of x-ray exposure, and starts to saturate after 3 mR, where the added system electronic noise becomes less important compared to x-ray quantum noise. In order to demonstrate all the important factors contributing to DQE degradation, in Figure 5.15, the values for DQE at 4 mR are shown as a function of detector stage for three spatial frequency values: 0, 1.55 and 3.1 lp/mm. It shows that $DQE(0)$ is 0.67 after the x-ray attenuation stage due
Figure 5.13: The measured temporal autocorrelation function of the prototype detector. It was obtained by averaging the autocorrelation function of 100 consecutive frames for 25 pixels.

to the quantum efficiency of a 300 μm a-Se layer for a 70 kVp x-ray spectrum. At the gain stage, due to the Swank factor of a-Se and the broad x-ray spectrum, $DQE(0)$ is further degraded to 0.6. Then $DQE(0)$ is decreased by the pixel fill factor of $F_p = 0.7$ at the aliasing stage and becomes 0.42. Finally $DQE(0)$ decreases to 0.38 after the addition of electronic noise.

From the measured x-ray sensitivity, presampling MTF and NPS, the spatial frequency dependent DQE of the prototype is calculated using Eq. 5.7. The value for $\bar{\omega}_0$ used in the calculation was $1.47 \times 10^5 \text{ mR}^{-1} \text{mm}^{-2}$ [156]. Shown in Figure 5.16 are the $DQE(f)$ for four different x-ray exposures calculated from the fitted white noise power spectra shown in Figure 5.11. Since the NPS is white, the shape of DQE follows that of the square of the presampling MTF. At 0.94 mR, $DQE(0)$ is just below 0.3. As x-ray exposure increases, electronic noise becomes less important, and $DQE(0)$ increases. The increase of $DQE(0)$ starts to saturate at 3.77 mR,
Figure 5.14: The calculated spatial frequency dependent detective quantum efficiency (DQE) as a function of x-ray exposure for the prototype detector, obtained using the cascaded systems model.
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Figure 5.15: The propagation of DQE as a function of detector stages for the prototype detector at three spatial frequencies: 0, 1.55 and 3.1 lp/mm, obtained using the cascaded systems model.

where $DQE(0) = 0.38$. The values for $DQE(f)$ shown in Figure 5.16 is in close agreement with the results of theoretical analysis shown in Figure 5.14. To illustrate the agreement between the experimental measurement and the theoretical analysis, Figure 5.17 shows comparison between the $DQE(f)$ measured at 3.77 mR (using raw NPS data) and the $DQE(f)$ obtained from the theoretical analysis for the same exposure.

5.4.5 Radiographs of phantoms

Digital radiograph of the x-ray bar pattern is shown in Figure 5.18 (a). The bars are aligned approximately along the gate line direction of the active matrix. The group of bars with spatial frequency of 2.82 was clearly resolvable. Aliasing starts to be visible at 3.19 lp/mm, and becomes severe for spatial frequency of 3.54 lp/mm and above. This is consistent with the sampling Nyquist
Figure 5.16: DQE plotted as a function of spatial frequency up to the Nyquist frequency for the prototype detector obtained using Eq. 5.7 and experimental measurements of the presampling MTF and the NPS (fitted white NPS) for x-ray exposures shown.

Figure 5.17: DQE as a function of spatial frequency for the prototype detector and an exposure of 3.77 mR. Experimental measurements are shown as filled squares, and the theoretical analysis is shown with a solid line.
frequency of 3.1 lp/mm by 160 μm pixel pitch. If the bar pattern is placed at 45 degree angle with respect to the gate and data line directions, the highest spatial frequency bars that can be resolved will be increased by the square root of two. Figure 5.18 (b) is an image of a finger phantom obtained at 40 kVp and 2.2 mR, which is a clinically relevant exposure level for radiography. The image appears to be x-ray quantum noise limited, and features such as the trabeculae and bone cap are clearly visible. Figure 5.19 shows a hand phantom image that has been assembled offline from 14 subimages acquired under similar conditions to those used for the finger phantom image.

Figure 5.18: The phantom images obtained using the instrumented section of the prototype detector: (a) Radiograph of a x-ray bar pattern; (b) Radiograph of a finger phantom obtained at 40 kVp and 2.2 mR.

5.5 CONCLUSION

A prototype self-scanned a-Se detector was built based on an active matrix with CdSe TFTs. Real-time image acquisition was achieved and digital radiographs were obtained at clinically relevant x-ray exposure levels. The quantitative image performances of the prototype detector, i.e. x-ray sensitivity, presampling MTF, NPS and DQE, were studied both theoretically and experimentally. The measured x-ray sensitivity agrees with the theoretical calculation. The presampling MTF of the prototype closely matches the aperture function of the pixel electrodes indicating that the a-Se layer has very little degradation effect on the image resolution, as expected from previous
theoretical calculations. The measured noise power spectrum is white and the noise power of the 
x-ray quantum noise increases linearly as a function of exposure, both as expected. Because of 
the high electronic noise, the prototype detector is x-ray quantum noise limited only for exposure 
greater than 3 mR. In order to achieve quantum noise limitation at fluoroscopic exposures, the pixel 
electronic noise of the prototype needs to be reduced by more than an order of magnitude. However, 
the lack of temporal correlation, i.e. lag, shown by the measurement of temporal autocorrelation 
function demonstrates the ability of real-time readout. The experimental determination of detective 
quantum efficiency from the measured x-ray sensitivity, presampling MTF and NPS agrees within 
experimental error with the theoretical analysis using a cascaded system model. This confirms 
our understanding of all the important factors contributing to DQE degradation: (1) the quantum 
efficiency of a-Se; (2) the pixel fill-factor; (3) the Swank factor of a-Se and (4) the electronic noise. 
It provides a basis for maximizing $DQE(f)$ for future detector design.
Chapter 6

Summary and Future Work

6.1 Summary of the Thesis

In Chapter 1 of the thesis, I reviewed the current analog radiological imaging technologies, i.e. screen-film radiography and XRII / video fluoroscopy, and their limitations. The rationale for developing digital radiology was outlined. Digital radiological imaging detectors that have been developed, or currently under development were reviewed, and their advantages and disadvantages compared. The flat-panel digital detectors, which are based on the technology of large area active matrices, can perform both radiography and fluoroscopy. Our approach for making a flat-panel detector, which is the direct method using a-Se, has both fundamental and practical advantages over the indirect method using phosphor screens or CsI.

6.1.1 TFT design for a reliable flat-panel detector

In Chapters 2 and 3 of the thesis, I described two factors that can affect the reliability of the direct, flat-panel detectors: radiation hardness and protection against high voltage damage, respectively. In Chapter 2, the radiation hardness of the CdSe TFTs used in our investigation was studied
experimentally at radiation doses up to 1000 Gy, which is 20 times the expected life-dose (50 Gy) of a diagnostic x-ray imaging detector. It was found that the dominant effect of radiation is the negative shift of threshold voltage $V_T$ after irradiation, which makes it more difficult to turn off the TFT. The mechanism for negative $V_T$ shift is radiation generated holes in the gate insulator being permanently trapped. The change in $V_T$ for existing CdSe TFTs is less than 1 V after the life-time dose for diagnostic x-ray imaging, which does not affect the normal operation of the TFTs thus ensuring an adequate radiation hardness. However if the existing CdSe TFTs are applied to imaging applications that require higher doses, *e.g.* portal imaging in radiation therapy which has an estimated life-time dose of $10^5$ Gy, the magnitude of $V_T$ shift is sufficiently large to prevent the TFTs from being turned off with normal gate operational voltage. For these imaging tasks, the radiation hardness of the CdSe TFTs would have to be improved. Sufficient radiation hardness can be achieved by using a thinner gate insulator, preferably with less sensitivity to radiation (*e.g.* Si$_3$N$_4$ instead of the current SiO$_2$ gate insulator).

In Chapter 3, a novel concept for protecting the TFTs from possible high voltage damage was proposed. When a high voltage is applied to a TFT, it can cause irreversible damages in two mechanisms: (1) dielectric breakdown of the gate insulator; and (2) collapse of the TFT on current. When the detector is operated in normal fluoroscopic or radiographic mode, the pixel potential does not rise above 10 V and will not cause damage to the TFTs. High voltage buildup on the pixel (*e.g.* 100 V) that can cause irreversible damage to the TFTs is only possible under an abnormal combination of the following two conditions: (1) a suspended detector scan; and (2) an x-ray exposure more than 10 times higher than that normally used in diagnostic imaging. Although this fault condition is unlikely, it is perhaps to be expected occasionally during the lifetime of the detector and thus a high voltage protection mechanism is necessary. The concept proposed by us
CHAPTER 6. SUMMARY AND FUTURE WORK

for high voltage protection employs a dual-gate TFT structure with a high voltage protection gate in addition to the normal scanning control gate. The protection gate turns the TFT on before the pixel potential reaches a damaging value regardless of the voltage on the control gate, thereby preventing possible high voltage damage. However under normal x-ray exposures, the control gate operates the TFTs reliably and the protection gate has little effect. The advantage of the dual-gate design over the approach of adding a second TFT at each pixel for high voltage protection is that the same TFT is used for both scanning control and protection, thereby minimizing the complexity of each pixel and allowing for higher production yield and lower cost. The TFT characteristics predicted by our dual-gate TFT model showed that the saturation value for the pixel potential \( V_{sat} \) depends on both the ratio between the thicknesses of the two gate insulators and the current generated in a-Se per pixel, \( I_{Se} \). For each x-ray imaging application, the ratio between the two gate insulator thicknesses can be established based on the maximum value of \( I_{Se} \) (under the fault conditions). Then the dynamic range of the detector can be ensured by adjusting the value of pixel capacitance \( C_p \). Using detector and x-ray parameters for mammography and chest radiography as examples, it was shown that practical gate insulator thicknesses of less than 1 \( \mu \)m and a standard pixel capacitance of \( C_p = 1 \) pF are adequate for providing reliable high voltage protection as well as the dynamic range required by these imaging applications.

6.1.2 Theoretical analysis of \( DQE(f) \)

In Chapter 4, a theoretical analysis of the spatial frequency dependent DQE of the self-scanned a-Se detector was performed. The major contribution of the study is to take into account the effect of aliasing in the analysis of \( DQE(f) \), since there has been no previous study on this subject for flat-panel detectors. The results of the analysis showed that the main factors contributing to the loss of \( DQE(f) \) are: (1) the quantum efficiency of a-Se, \( \eta \); (2) the pixel fill-factor \( F_p \) of the
active matrix; (3) the Swank factor $A_S$ associated with the gain of $a$-Se; and (4) the electronic noise $q_n$. The first two factors ($\eta$ and $F_P$) are both proportional to $DQE(f)$, and the last two factors ($A_S$ and $q_n$) both of which are noise sources cause degradation in $DQE(f)$. The effect of $q_n$ is especially severe at low x-ray exposure levels (e.g. in fluoroscopy). Possible strategies for maximizing DQE were discussed for different x-ray imaging applications. For fluoroscopy, the most important strategy is to minimize $q_n$ so that the detector can provide high DQE at very low x-ray exposure rate (i.e. $10^{-4}$ mR/image). For mammography, due to the small pixel pitch (50 $\mu$m), the most important strategy is to increase $F_P$ by building a mushroom structure and possibly by manipulating the electric field in $a$-Se.

6.1.3 Construction and evaluation of a prototype real-time detector

In Chapter 5, the construction and evaluation of a prototype real-time detector are described. The active matrix of the prototype has $360 \times 480$ pixels, out of which $360 \times 128$ pixels were instrumented with readout electronics, and the pixel pitch is $160 \mu$m. Aspects of the imaging performance measured were: x-ray sensitivity, presampling MTF, NPS and $DQE(f)$. The measurement of x-ray sensitivity of a pixel of the prototype agrees with our theoretical calculation. The presampling MTF of the detector closely matches the MTF of the pixel aperture function, indicating that as expected, $a$-Se has very little degradation effect on image resolution. The measured NPS is spatially white, which is consistent with the lack of correlation between pixels due to the very high resolution of $a$-Se. The experimental determination of $DQE(f)$ using the measured x-ray sensitivity, presampling MTF and NPS agrees with the theoretical calculation obtained using the model developed in Chapter 4, thus confirming the validity of our DQE analysis. Images of a finger phantom were obtained with radiographic exposure levels using the prototype. They are x-ray quantum noise limited, and features such as the trabeculae and bone cap are clearly visible.
However, although the prototype was operated in real-time, x-ray quantum noise limited images cannot be obtained at fluoroscopic exposure rate due to the high electronic noise $q_n$ in the prototype detector. The value of $q_n$ needs to be reduced from the current 18,000 electrons (which is mainly due to the charge amplifier noise) to less than 1,000 electrons before the detector can be used at fluoroscopic exposure rates.

### 6.2 Future Work

The present work has established all the major factors affecting $DQE(f)$ of the self-scanned α-Se detector, and forms a guideline for detector design in different x-ray imaging applications. It also provides solutions to practical problems affecting the reliability of the detector, such as radiation hardness and protection against high voltage. Therefore the focus of the future work is to apply these knowledges and experiences to the development of self-scanned α-Se detectors to each x-ray imaging application.

Development of flat-panel detectors are being actively pursued by industry. For example, a fluoroscopic detector using the indirect method is being commercialized by the collaboration of Xerox dpiX and Varian. The main difficulty involved in this development is to keep the electronic noise level below 1,000 electrons. Sterling Diagnostic Imaging is commercializing their radiographic direct, flat-panel detector for chest radiography [60]. Since an active matrix 14" x 17" in size is not yet available with the current flat-panel technology, the approach of tiling four active matrices has been used to form a chest radiographic detector [157].

This work has been carried out through our collaboration with Litton Systems Canada, who currently makes active matrices up to 10" x 14" in size for AMLCD. Since the strengths of our approach are the high resolution provided by α-Se and the real-time readout, our goal is to first apply
this technology in mammography and fluoroscopy. Here I will discuss a few practical considerations in how to make our detectors feasible and with high $DQE(f)$ in these two applications:

6.2.1 Mammographic detector

In mammography, a pixel pitch of 50 $\mu$m is required. The smallest pixel pitch that has been made for AMLCD is 90 $\mu$m [54]. Because the carrier mobility in CdSe is two orders of magnitude higher than that in $a$-Si:H, the choice of channel width to length ratio ($W/L$) in CdSe TFTs is very flexible thus allowing small TFTs to be made with desirable characteristics for pixel pitch of 50 $\mu$m. The first step of our plan is to make a prototype detector with 50 $\mu$m pixel pitch. Figure 6.1 shows the top view of the design of photolithographic masks proposed for a prototype active matrix with 50 $\mu$m pixel pitch. The $W$ and $L$ of the TFT are 10 and 25 $\mu$m, respectively, and the masks consist of five levels for the delineation of: (1) gate metal and ground lines for the pixel storage capacitors; (2) vias in the bottom gate insulator; (3) CdSe channel; (4) vias in the top gate insulator; and (5) drain/source metal as well as pixel electrodes. Although this mask design only provides a $F_P$ of 0.55, it will result in an active matrix process that is compatible with the existing technology for AMLCD. It allows us to investigate the characteristics for small TFTs, and explore the resolution performance of the detector (i.e. presampling MTF) at 50 $\mu$m pixel pitch.

Once the feasibility of making 50 $\mu$m pixels is established, we plan to investigate the practicality for: (1) incorporating the dual-gate TFT structure for high voltage protection, using the contact n-type doping method described in Chapter 3; and (2) building a mushroom structure. Based on the design shown in Figure 6.1, a mushroom structure as described in Chapter 3 can be made by adding two more mask levels for the delineation of: (6) a thick (∼1 $\mu$m) top passivation layer; and (7) the mushroom electrodes. Making the mushroom structure can increase the pixel fill-factor $F_P$ from the current 0.55 to 0.81. I believe that all these steps are feasible in practice
Figure 6.1: Top view of four pixels of the mask design for a mammographic detector with 50 μm pixel pitch. The width and length of the TFT channel are 10 and 25 μm, respectively.
and will result in high $DQE(f)$ as predicted in Chapter 4 of this thesis.

6.2.2 Fluoroscopic detector

As shown in Chapter 4, in order to retain the maximum $DQE(f)$ over the entire x-ray exposure range for fluoroscopy ($10^{-4}$-$10^{-2}$ mR), the electronic noise $q_n$ has to be less than 200 e which is almost impossible to achieve. The requirement for $q_n$ is less critical if the gain of the a-Se can be increased thus generating more charge for each absorbed x-ray. This can be achieved by increasing the electric field $E_{Se}$ in a-Se. However, since the current $E_{Se}$ of 10 V/μm already results in a high voltage bias of 5000 V, increasing $E_{Se}$ may not be a very practical solution.

It is known that other photoconductors, such as thallium bromide (TlBr), lead iodide (PbI$_2$), cadmium telluride (CdTe) and cadmium zinc telluride (CdZnTe), have higher gain than a-Se and are being proposed for x-ray imaging applications. TlBr and PbI$_2$ layers have been investigated as an x-ray sensor-target in a large area x-ray vidicon [158]. Crystalline CdTe or CdZnTe (e.g. Cd$_{0.9}$Zn$_{0.1}$Te) have also been investigated for industrial and medical x-ray imaging applications [159, 160]. The energy $W_\pm$ required to generate a charge pair is 50 eV in a-Se at $E_{Se} = 10$ V/μm, whereas $W_\pm$ is 6.5 and 4.5 eV in TlBr and CdTe, respectively [161]. Therefore detectors made with these photoconductors will generate much more image charge for each absorbed x-ray thus be more tolerant to electronic noise than with a-Se. Shown in Figure 6.2 are the plots of $DQE(f)$ of a fluoroscopic flat-panel detector with CdTe as the x-ray photoconductor calculated using the theoretical model developed in Chapter 4. Figure 6.2 (a) is a surface plot of the $DQE(f)$ as a function of the exposure range required for fluoroscopy. It shows that the maximum $DQE(0)$ of 0.78 is retained throughout the entire exposure range. Figure 6.2 (b) is a surface plot of the $DQE(f)$ as a function of the electronic noise $q_n$. It shows that $DQE$ is not degraded for $q_n$ up to 2,000 e. Even when $q_n = 5,000$ e, $DQE(0)$ is only reduced by ~0.2, therefore the detector is much
more tolerant to electronic noise than a-Se.

However, several key features of a-Se which make it a satisfactory x-ray photoconductor need to be engineered for TlBr, CdTe and CdZnTe layers before they can be used for flat-panel detectors: (1) low dark current; (2) capability of depositing thick layers in large area with low cost; (3) adequate mobility and lifetime for both electrons and holes; and (4) deposition at moderate or low temperature (<300°C), since once an active matrix is made, it cannot be subjected to temperature higher than 300°C which can alter the characteristics of the TFTs.

6.2.3 Summary of future work

The focus of our future work is to make direct, flat-panel detectors for mammography and fluoroscopy. I believe that with the current active matrix technology, it is feasible to make a flat-panel mammographic detector with 50 μm pixel pitch. Therefore flat-panel digital mammography with high DQE(f) may be the first imaging application that becomes reality using our approach. For fluoroscopy, it is not practically feasible with the current technology to make a detector with optimum DQE(f) for the low exposure level (10⁻⁴mR/image) used in fluoroscopy. In order to achieve the goal of flat-panel fluoroscopy, different approaches such as increasing the gain of a-Se or using other types of photoconductors with higher gain may be utilized. However they are more technologically demanding and may require a few more years of fundamental investigation before they can be applied to the manufacture of flat-panel detectors.
at the lowest x-ray exposure level in fluoroscopy (10 μm/image).

\[ \Phi(x) = \Phi(0) + \Phi(\text{noise}) \]

range encountered in fluoroscopy with \( \Phi(0) = 1000 \), and \( \Phi(\text{noise}) \) as a function of electronic noise.

200 μm thick layer of Crl 6 as the x-ray phosphor as a function of the exposure of the detector.

Figure 6.2: Theoretical analysis of DOE and spatial frequency of (a) the direct, and (b) the indirect, detector for fluoroscopy with a
Appendix A

Derivation of Aliased NPS

The NPS after aliasing is given by Eq. 4.21. In this appendix, it is shown that when the pixel width \( a \) and pixel pitch \( d \) satisfy \( a \leq d \), the aliased NPS is white, i.e. \( Q(f) = \sum_{n=-\infty}^{\infty} \sin^2[\alpha(f - \frac{n}{d})] \) is constant. Since \( Q(f) \) is periodic \((Q(f) = Q(f - \frac{n}{d}))\), only the case of \( 0 \leq f < \frac{1}{d} \) needs to be shown.

When \( f = 0 \),

\[
Q(0) = \sum_{n=-\infty}^{\infty} \frac{\sin^2(\frac{\pi a n}{d})}{(\frac{\pi a n}{d})^2} = 1 + 2 \sum_{n=1}^{\infty} \frac{\sin^2(\frac{\pi a n}{d})}{(\frac{\pi a n}{d})^2}
\]

Using the trigonometric series:

\[
\sum_{n=1}^{\infty} \frac{\sin^2(n x)}{n^2} = \frac{x(\pi - x)}{2}, \tag{A.2}
\]

Eq. A.1 can be reduced to:

\[
Q(0) = 1 + \left(\frac{d}{\pi a}\right)^2 \frac{\pi a}{d} \frac{\pi a}{d} = \frac{d}{a}. \tag{A.3}
\]

When \( 0 < f < \frac{1}{d} \),

\[
Q(f) = \left(\frac{d}{\pi a}\right)^2 \sum_{n=-\infty}^{\infty} \frac{1 - \cos[2\pi a(fd - n)]}{2(n - fd)^2}. \tag{A.4}
\]
Using the trigonometric series:

\[ \sum_{n=-\infty}^{\infty} \frac{1}{(n+a)^2} = \frac{\pi^2}{\sin^2(\pi a)}, \]  
(A.5)

and

\[ \sum_{n=0}^{\infty} \frac{\cos(nx + b)}{(n+a)^2} = \frac{1}{\Gamma(2)} \int_{0}^{\infty} \frac{te^{-at}[\cos b - e^{-t} \cos(x - b)]}{1 - 2e^{-t} \cos x + e^{-2t}} \, dt, \]  
(A.6)

where \( \Gamma(2) = \int_{0}^{\infty} te^{-t} \, dt = 1 \), Eq. A.4 can be reduced to:

\[
Q(f) = \frac{d^2}{2a^2 \sin^2(fd\pi)} - \frac{1}{2} \left( \frac{d}{\pi a} \right)^2 \sum_{n=0}^{\infty} \frac{\cos\left[\frac{2\pi a(n+1-\frac{f}{d})}{d}\right]}{[n + (1 - f)d]^2} + \sum_{n=0}^{\infty} \frac{\cos\left[\frac{2\pi a(n+f)}{d}\right]}{(n + f d)^2}.
\]

Using the definite integral of:

\[
\int_{0}^{\infty} \frac{x^a-1 \ln x}{x^2 - 2x \cos \gamma + 1} \, dx = \frac{\pi}{\sin \gamma \sin^2(\alpha \pi)} \gamma \sin(\alpha \pi) \cos[(\alpha - 1)(\gamma - \pi)] + \pi \sin[\gamma(\alpha - 1)],
\]  
(A.8)

Eq. A.7 can be further reduced to:

\[
Q(f) = \frac{d^2}{2a^2 \sin^2(fd\pi)} - \frac{1}{2} \left( \frac{d}{\pi a} \right)^2 \frac{\pi}{\sin^2\left(\frac{2\pi a}{d}\right) \sin^2(fd\pi)} \left[ -\frac{2\pi a}{d} \sin\left(\frac{2\pi a}{d}\right) \sin^2(\pi f d) + \pi \sin\left(\frac{2\pi a}{d}\right) \right]
\]

\[
= \frac{d}{a}.
\]  
(A.9)

Therefore Eqs. A.3 and A.9 show that the aliased NPS for the self-scanned a-Se detector is white and the noise power is increased by \( d/a \).
Appendix B

Derivation of the noise power reduction by lag

For lag that exhibits a single exponential decay, \( r_l \), which is the ratio of the total noise power for a detector with lag to that without lag, is given by:

\[
\frac{r_l}{\int_0^\infty R^2(\nu) d\nu} = \frac{1}{\int_0^\infty R_0^2(\nu) d\nu} = \frac{\int_0^\infty \frac{1}{\tau_0^2 + (2\pi \nu)^2} \text{sinc}^2(T_F \nu) d\nu}{\int_0^\infty \text{sinc}^2(T_F \nu) d\nu},
\]  

(B.1)

where \( R(\nu) \) and \( R_0(\nu) \) are the temporal response of the detector with and without lag, respectively, \( \tau_0 \) is the time constant for lag with a single exponential decay, and \( T_F \) is the frame time. From the definite integrals of:

\[
\int_0^\infty \text{sinc}^2(T_F \nu) = \frac{1}{2T_F},
\]  

(B.2)

and:

\[
\int_0^\infty \frac{1/\tau_0^2}{1/\tau_0^2 + (2\pi \nu)^2} \text{sinc}^2(T_F \nu) d\nu = \frac{1}{2T_F} \left[ 1 - \frac{\tau_0}{T_F} (1 - e^{-\frac{T_F}{\tau_0}}) \right],
\]  

(B.3)
Eq. B.1 can be reduced to:

\[ r_l = 1 - \frac{r_0}{T_F} \left( 1 - e^{-\frac{T_F}{r_0}} \right), \]  

(B.4)

as required.
Bibliography


IMAGE EVALUATION
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