Requirements on Amorphous Semiconductors for Medical X-Ray Detectors

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Abstract

Solid state X-ray detectors based on large-area amorphous semiconductors such as amorphous silicon or selenium have been developed. The requirements for various applications in medical diagnosis determine the boundary conditions for different detectors. Key parameters are image receptor size, spatial resolution, image frequency, signal-to-noise ratio, and long-term stability. Conventional thorax radiographs are 43 × 35 cm² in size. Therefore, to replace film, large detectors are required. Mammograms need to display microcalcifications of some 100 µm in detail. Fluoroscopic images are taken at a rate of 30 images per second or even faster. Moreover, all detectors need to deliver optimum image quality at the lowest tolerable dose. The lifetime should be at least 10 years. In this paper we discuss different types of detectors that could fulfill those needs. Two fundamental concepts are compared: a scintillator coupled to a photodiode and a direct converting semiconductor, both being arranged on an amorphous silicon switching matrix.

Keywords: X-ray, Detector, Amorphous semiconductors, Medical imaging

1. Introduction

Today, medical X-ray diagnosis is mostly still based on conventional techniques such as film-screen combinations, storage phosphor plates, or an X-ray image intensifier-TV system (XRII-TV). But digital solid state detectors have begun penetrate the market.

There is still a debate on which type of X-ray detector is best to fulfill the needs of medical applications. For those not familiar with the boundary conditions put up by the field of projection X-ray imaging, this paper will give an overview on the requirements for this kind of image receptors. Of course, in the different fields of application, different solutions will be optimal. On the other hand, multi-purpose detectors are desirable which demand some compromises.

Often, the decision to select a system is governed by price or availability on the market. These aspects are beyond the scope of this paper.

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2. Key parameters

When developing a medical X-ray detector, the important parameters which are required for its application must be considered. The human body and the needs of the radiologists define the boundary conditions and affect the physical properties of the detector.

2.1. Image receptor format

When scatter is disregarded, X-rays always propagate along straight lines. Furthermore, radiation with a broad spectrum used for medical purposes cannot be deflected or focused. Therefore, in a medical application the radiation receptor has to be of at least the same size as the human body or the organs under investigation.

This fact calls for a detector based on large-area electronics containing extended semiconductor layers. Hydrogenated amorphous silicon (a-Si:H) has proven to be the best material available for this purpose [1]. Other thin film materials such as amorphous selenium (a-Se) or cesium iodide (CsI) are used for X-ray absorption. The areas required for different applications are listed in Table 1.

Table 1

Application	approximate format [cm²] or [cm Ø]	Technology
Dental (intra-oral)	2 × 3	film, CCD
Mammographic biopsy	5 × 9	film, CCD (OPDIMA®)
Card angiography	17 / 23 Ø	XRII-TV
Mammography	18 × 24	film, storage phosphor
Neurology, fluoroscopy	27 / 33 Ø	XRII-TV
Angiography	40 Ø	XRII-TV
Radiography, thorax	35 × 43	film, storage phosphor
Radiography, thorax	43 × 43	a-Si (PIXIUM 4600®)

If large formats are required, in conventional radiology film-screen systems or storage phosphor plates up to 35 × 43 cm² are used. XRII-TV systems cover the range from 17 to 40 cm diameter input screen.

If rather small areas are sufficient, CCDs can be used. A CCD sensor is coupled to a fiber-optic plate (FO) which serves as a radiation shield. The FO is covered with a scintillator, for instance CsI. Such a detector with 40 µm pixel size is used for dental imaging. However, CCD technology is limited in size. The largest CCD of the world produced in quantity today is 49 × 86 mm² in size (i.e. 4K × 7K pixels of 12 µm). It is used in a sensor for mammographic biopsy [2].

Because most of the applications mentioned above require larger detectors than CCDs, a-Si based detectors have been developed. Companies such as Canon [3], dpiX [4], EG&G [5], Sterling [6], TriXell [7], and Varian [8] have demonstrated X-ray detectors up to 43 × 43 cm² in size.

2.2. *Spatial resolution*

The required performance in spatial resolution depends on the kind of objects a radiologist wants to investigate. Soft tissue can be imaged with 1 - 2 lp/mm, fine bone structures require > 3 lp/mm to be precisely imaged. Resolutions > 5 lp/mm are demanded for mammography detectors (where microcalcifications of some 100 µm shall be imaged) or for dental applications.

The spatial resolution which can be achieved by a detector depends on different physical processes. The primary interaction of X-rays with matter leads to Compton scattering, the cross section depending on X-ray energy and on the K-edge of the absorber. Scattered quanta are absorbed at a certain distance from the primary trajectory, thus degrading the spatial resolution.

Real detectors are not idealized planes, but have a non-zero thickness. Therefore, X-rays impinging the detector under an angle different from the normal direction are smeared laterally.

If a scintillator is used as an absorber, light propagation in the scintillator as well as the optical coupling between scintillator and photodiodes have also to be considered. For radiographic applications, CsI with a thickness of 400 to 600 µm has proven to be a good compromise. If a very high resolution is required, 100 to 200 µm are more suitable.

The advantage of a directly converting semiconductor is the extremely high spatial resolution achievable. Since no optical effects disturb the image, only lateral diffusion of charge carriers can reduce resolution. With the data for a-Se, even a 500 µm thick a-Se layer has, applying normal operation voltage, a drift length exceeding by far the diffusion length. Hence, the a-Se layer is suitable for spatial resolutions even > 10 lp/mm.

The array structure of the solid-state detectors introduces another reduction of spatial resolution. The signal created is averaged over the active part of the pixels, i.e. the area of the photodiodes or the electrodes of the directly converting semiconductor layer. Moreover, sampling is performed at discrete points which may lead to aliasing effects [9]. High spatial frequency objects are imaged with a so-called pseudo sharpness, even Moire patterns may appear.

2.3. *Frame rate*

Solid state detectors operate in real time. Real time in cardiac imaging means 30 Hz (or even 60 Hz in pediatry). Fluoroscopy is performed at 15 to 30 Hz, but can be slowed down to 6 to 12 Hz in order to reduce X-ray dose. Digital subtraction angiography (DSA) is done at 2 to 8 Hz. In single-image applications such as radiography, mammography, or dental imaging, even one image every 30 s can still be called real time, compared to the time required for film exposure, transport, and development.

Two main intrinsic factors limit the performance of an imager panel in the time domain: image lag and incomplete charge transfer. The latter is due to the on-resistance of the readout switches which leads to some 10 µs required to assure that at least 99% of the signal can be transferred to the readout amplifiers.

Image lag has its origin in both, scintillator and a-Si readout matrix. The scintillator afterglow can be reduced to an acceptable level by selecting the appropriate material and carefully controlling deposition parameters, doping level, and post deposition treatment (annealing) [10].

Lag from a-Si is caused by trap filling and emission of charge carriers from traps. The number of traps involved can be reduced by minimizing the geometrical device volume. The a-Si bulk density of states has already reached a concentration of $\approx 10^{15}\ cm^{-3}$ in state-of-the-art material. Lag effects can be reduced notably by keeping the traps always filled. This filling can be accomplished by a flashed backside illumination of the detector. Especially long-term lag can thus be quenched by more than a factor of 10. Moreover, the lag can be reduced by a short integration time prior readout.

Also extrinsic factors limit the achievable frame rate. Small signal charges (<< 1 pC) have to be fed through long (50 cm) wires with high (100 pF) capacitances. Signal charges have to become completely transferred to a sample-and-hold circuit. Data have to be transmitted with rates of more than 100 MB/s. Moreover, image conditioning has to be done in real time, i.e. offset subtraction and gain correction for each pixel, and interpolation over defective pixels. Some medical investigations require also more sophisticated post-processing, for instance image averaging (time domain), spatial filtering (edge enhancement), or motion detection. Finally, images have to be rapidly displayed and refreshed on high-resolution monitors. All data have to be stored and retrieved in real time.

2.4. Signal-to-noise ratio

Medical applications require imaging of small and low contrast details. Therefore, a high signal-to-noise ratio (SNR) is needed. The signal is proportional to the X-ray dose (as solid state detectors show a linear response), the absorbed fraction of the incident radiation, and the sensitivity.

The total noise consists of several contributions. X-ray quantum noise is determined by the input dose, the detector sensitivity, and the statistical fluctuations in the conversion / gain processes inside the detector. Electronic noise from pixel reset, leakage currents, line resistances, readout amplifiers, and the analog chain can be dominant, especially at low dose. The non-ideal analog-to-digital conversion can also be regarded as a kind of noise.

Signal and noise propagation through the detector is described by the detective quantum efficiency (DQE),

$$DQE(\nu) = \left(\frac{SNR(\nu)_{out}}{SNR(\nu)_{in}}\right)^2$$

which expresses the X-ray quanta utilization compared to that of an ideal detector. ν denotes the spatial frequency.

The contrast-detail detectability depends, of course, on the size of the object and its contrast. Therefore, the X-ray spectrum has to be adapted to the matter under investigation by optimizing tube voltage, anode material, and beam filtering. The atomic numbers and the K-edges of the materials involved have to be considered.

Having in mind that the human eye and visual perception system are also part of the imaging chain, care has to be taken for an optimum monitor or light box without interfering ambient illumination. The eye of the observer is very sensitive to correlated effects. Therefore, line-correlated noise has to be suppressed carefully.

In general, the signal-to-noise ratio in the image depends on the DQE, the applied X-ray dose, and the input beam spectrum. New detectors should not need more dose than those already in use today. For fluoroscopy, an average (system) dose of 10 nGy per image in the region of interest (ROI) must be sufficient. In dark parts of the image, the dose will be about a factor of 3 less, 10 times the system dose has to be processed linearly. Direct radiation hitting the detector may provide a dose exceeding the system dose by a factor of 100 or more. This dose should neither destroy the detector nor cause artifacts from long term image lag. Solid state detectors under

development today (CsI or a-Se on a-Si) allow a S/N ratio about a factor of 3 to 8 less than required for fluoroscopy.
System doses for radiography are 2.5 µGy, according to a film-screen system with a sensitivity class 400, or less. In film mammography 70 µGy are used. A DSA image needs some 10 µGy. For these doses, a-Si technology provides a better S/N ratio than conventional systems.

2.5. Long Term Stability

Medical equipment should operate for at least 10 years. During this time, the operational parameters of a detector have to remain constant or should be easily adjustable.
If semiconductors are irradiated (with visible or X-ray photons), defects are created. In a-Si, where it is known as Staebler-Wronski effect, only few defects are created with the small doses applied and under normal reverse bias operating conditions. The defects increase recombination and reduce the Schubweg. In a-Si thin-film transistor (TFT) switches a small shift of the threshold voltage due to space charges in the dielectric is observed which may affect proper operation. However, life cycle investigations of a-Si detectors could prove their outstanding radiation hardness.
A selenium layer used as an X-ray absorber shows a fatigue effect. The charge delivered by a given dose decreases by some percent after several images have been taken.
Scintillators show an alteration in light output during aging. CsI is known to increase its conversion yield by an order of one percent after intense irradiation. This effect can persist for days. CsI is also sensitive to moisture and has therefore to be encapsulated very effectively.

3. Alternative concepts

There are two fundamentally different approaches for an X-ray detector, (1) a scintillator coupled to an array of photodiodes, or (2) a directly converting X-ray sensitive semiconductor. Both make use of a readout matrix consisting of switches, i.e. TFTs or switching diodes.

3.1. Scintillator with photodiode

By using a scintillator to absorb the incident X-rays the absorber can be individually optimized. An evaporated layer of cesium iodide doped with thallium (CsI:Tl) is widely used, because it has a microstructure consisting of thin (5 - 10 µm diameter) needles. This structure has a non-isotropic propagation of the scintillating light resulting in excellent spatial resolution. Commercially available screens with Gd_2O_2S as scintillating material (Lanex) are a cost effective alternative, but yield a poorer performance.
The generated light is coupled to a readout matrix consisting of a photodiode and a switch per pixel. In the reversely biased photodiode, electron hole pairs are created, integrated on the photodiode capacitance, and subsequently transferred via the switch to the readout amplifier. A potential drawback of this concept is the limited fill factor (i.e. the ratio of the photodiode area to the pixel area) which becomes a problem at very small pixel pitch (< 70 µm). An approach to solve this dilemma would be to arrange the photodiodes as a separate layer above the switching matrix.

3.2. Directly Converting Semiconductors

Directly converting semiconductors avoid the limitations of two subsequent quantum conversion processes. On the other hand, one single material has to fulfill all requirements at once, high X-ray absorption and charge carrier generation, low lateral diffusion and recombination rate, and fast charge transport to the electrodes.
A suitable semiconductor for this purpose is amorphous selenium (a-Se) because it can be deposited on large areas in good quality. Since the semiconductor is deposited as a continuous layer on top of the switching matrix, generally, smaller pixels can be produced without a loss of fill factor.
Seeking for new materials, the following demands have to be considered. The DQE should be > 70%. Therefore, the X-ray absorption has to be also > 70%. This can be accomplished by a semiconductor containing elements with high atomic number deposited in an adequate thickness.
The thickness d, however, should be less than 1 mm to avoid a reduction of spatial resolution

due to quanta with oblique incidence. The Schubweg $\lambda=\mu\tau E$ should be large compared to the thickness, therefore, carrier mobility μ, lifetime τ, and electric field E have to be high enough. With $\lambda=10d$ nearly optimum charge collection can be achieved.

A problem can arise if the voltage needed to produce the required field E becomes too high. Voltages above 5 kV are difficult to handle in vicinity of very sensitive readout circuits. Moreover, the dark current might rise to undesirable magnitudes. It has to be kept as low as possible, because it is a source of noise. For that reason, the Fermi level in the semiconductor should be located close to midgap and blocking contacts have to reduce the dark current as far as feasible. Depending strongly on temperature, larger dark currents call for frequent offset calibration which is unfavorable. A comparison of some semiconductors is displayed in Figure 1 with respect to thickness, dark current, and bias voltage.

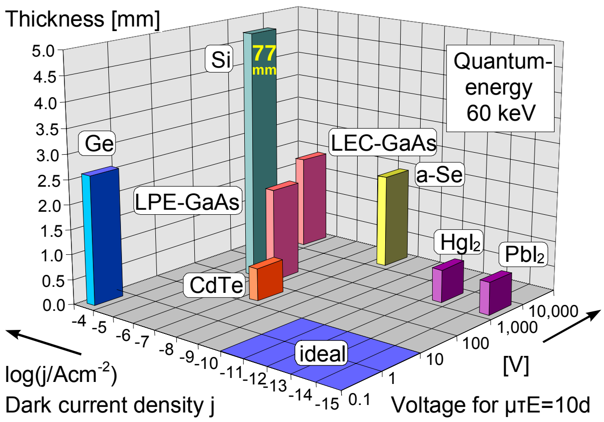

Fig. 1: Thickness required for 90% absorption at 60 keV. Different semiconductors (Ge, Si, Se, GaAs [11], CdTe, HgI_2, PbI_2) are shown with the dark current densities and voltages needed for good charge collection [12].

4. Discussion

For the readout matrix, a-Si has proven to be a favorable semiconductor for making switches, i.e. TFTs or diodes. Photodiodes also made from a-Si with CsI as a scintillator have been demonstrated by several manufacturers to be a good solution.

Direct converting semiconductors still offer a wide variety of new approaches. They often have, like a-Se, a very small diffusion length compared to the drift length, leading to an excellent spatial resolution. High-Z materials like PbI_2 [13] or HgI_2 with a high mass density effectively absorb X-rays even in thin layers. The low pair creation energy assures a high signal which, together with a low dark current, is prerequisite for a good signal-to-noise ratio. Deposition conditions have to be optimized to assure high carrier mobility, little recombination, and a low density of states. The latter is mandatory for a fast response and a low lag detector.

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