Comparison of Amorphous Selenium and Amorphous Silicon Flat-Panel Detectors

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Abstract

Digital radiographic flat-panel detectors may operate with direct or indirect manners of converting x-ray photon energy into spatial energy signals. Advances in both technologies have led to radiographic work areas that are faster and lend toward lower patient radiation doses. Superior image quality with an abundance of post-processing capabilities is a benefit as well. Whether based on amorphous selenium or amorphous silicon, detector hardware including the energy absorbing semiconductor components are remarkably small and highly engineered. There are clear benefits of using one detector type over the other and there are areas where comparative research needs to be further examined. At this time, it appears optimal selection of either type depends on selecting the proper detector for the intended application.
Introduction

Interventional and diagnostic digital flat-panel x-ray detectors were first used in radiography and mammography. Today, they are used in angiography, fluoroscopy and general radiography (Spahn, 2005). They are commonly seen contained within table buckys, wall buckys, portable imaging equipment, mammography, and interventional equipment (Feuerbach, Hamer, Strotzer, & Volk, 1998).

Digital x-ray-radiation acquisition equipment allows obtainment of diagnostic images from the general radiography x-ray energy spectrum. Digital radiography includes a vast number of detector technologies (Bassir et al., 2003). In this discussion, only flat-pans based on amorphous silicon (a-Si) and amorphous selenium (a-Se) will be reviewed. Both fall within a category of energy integrating detectors in digital radiography, as opposed to photon counting detectors (Acciavatti & Maidment, 2010). Also, terms such as indirect and direct are equally vast in radiography. For example, film-screen systems may loosely be considered as indirect. Again, for the sake of this discussion, both terms are used to describe a-Se and a-Si detectors and the general class they fall into.

Literature Review

Detector Requirements

Regardless of generalized type, any medical imaging x-ray detector needs to efficiently and without large geometric distortion convert a volley of x-ray photons into an image signal. It needs to ultimately provide a two-dimensional representation of the spread of these photons traveling from an irradiated object. That two-dimensional diagnostic image is created using distinct signal units as bits and distinct spatial units as pixels (Del Guerra, 2004). Flat-pans perform this task with high spatial resolution independent of the dose applied, a huge
technological step from earlier film screen methods (Spahn, 2005). The chest x-ray is the perfect example of a very common exam needing the most improved contrast characteristics where fine detail is required for the detection of the very smallest of pathologies (Bassir et al., 2003). Regardless of the exam, however, a medical detector needs to provide the highest quality images to diagnose injury and pathology.

Flat-Panel Characteristics

Two general types of flat-panels detectors exist. They are direct capture and, more commonly, indirect capture detectors. Both provide the capability to reduce dose when used properly. They allow faster work-flow in radiographic exam rooms due to quick acquisition and display of images. Both also promote increased image quality. They have a high sensitivity to radiation and the ability to provide an instant image. Both types of detectors also provide high contrast capabilities, largely distortion-free images and large dynamic ranges. An added benefit for certain flat-panels is a high compatibility with powerful magnetic fields; this increasing their interventional capabilities near magnets and lends to the creation of new interventional techniques (Spahn, 2005).

Flat-panel hardware needs parts incredibly small to allow them to be placed within a pixel, or the smallest photosensitive element of the detector. This placement must be obtained without limiting the acquisition capabilities of that pixel. The end objective is for the detector to provide a digital image that looks like an analog image to the human eye. In addition to all the technical specifics, any direct or indirect detector needs to be inexpensive enough to be affordable. Perhaps most importantly, a medical patient must expect that their imaging exam is using as little radiation as is necessary to provide that diagnostic image (Carroll, 2011).

Indirect Capture
In indirect capture flat-panel detectors, a scintillator is used for the conversion of x-ray photons into an electrical charge. Most often, an a-Si matrix is united with an intermediate cesium iodide (CsI) scintillation layer (Feuerbach et al., 1998). (See Figure 1) This scintillation layer converts the photons into visible light. Commonly, photodiodes, but sometimes cameras, are then used to change the light into an electrical signal charge (Bassir et al., 2003).

The indirect conversion process using scintillators utilizes attenuated x-ray quanta to create highly energized electrons. These electrons lose energy within the scintillation material because they create electron-hole pairs. Light is produced when these electron-hole pairs join together (Spahn, 2005).

Optical converters typically used in a-Si indirect detectors are thallium-doped CsI scintillators, or gadolinium oxysulfide phosphors (Del Guerra, 2004). CsI is grown in a needle-shaped crystalline structure that because of its shape, projects its green-colored light to the photodiode with very little scatter. The photodiode matrix array is designed to operate most effectively with the color light produced by its accompanying scintillator material (Spahn, 2005). The resulting current produced is directly proportional to the fluorescence emitted by this scintillator layer (Del Guerra, 2004). The photodiode covers about 70% of a pixel approximately sized 150 µm across. A single absorbed 60 keV x-ray photon would generate a charge of about one thousand electrons in a typical a-Si indirect detector. In the thin film transistor (TFT)-based pixel design, each TFT is coupled to one a-Si sensor (Spahn, 2005). The result is a pixel detector with a high sensitivity to light but low sensitivity to x-ray photons (Del Guerra, 2004). The photodiode acts as a capacitor, and as such, it holds a full bias voltage before an x-ray exposure is even made. Electron-hole pairs are created when, during the x-ray exposure, the photodiode receives the light photons from the scintillator. During acquisition a negative charge accumulates...
because of the drifting apart of these electron-holes on the sensor capacitor. The TFT switches to an “on” position allowing charge flow to the photodiode which resets it by discharging the photodiode. Electronics measure these charges which become the effective signals. Each row in the matrix with all of its pixels and the pixels’ associated signals are collected simultaneously. Electronic circuits amplify the received signal and measure its charge (Spahn, 2005). An analog-to-digital converter digitizes the signal from a multiplexer (Del Guerra, 2004). The resulting digital signal may then be further processed as needed.

A-Si over the past two decades has been researched as a semiconductor material for digital detectors. A-Si allows large matrices of single photodiodes to be produced. For example, matrices with several thousand by several thousand pixels allow for a detection surface of around 43 square cm in area, a size appropriate for all general radiography needs. A-Si detectors generally are combined with a phosphor or scintillator layer because a-Si has a low interaction probability (low Z) and is relatively thin. One large production consideration concerns the overall sensor area of an array. It is reduced by what is called dead area. This dead area in pixel cells is caused by the presence inside the cell of the transistor switch. Careful cell design will minimize this and maximize pixel fill factor (Del Guerra, 2004).

**Direct Capture**

A photoconductor, most commonly (a-Se), is used to convert x-ray photons into electrical charges in direct capture flat-panel detectors. This direct method of conversion occurs without producing visible light. Thus, the intermediary light-producing and light-absorbing steps are not required.

Direct conversion depends on several things to function. A photoconductor material with high x-ray absorption is needed. Also, with each x-ray photon, the number of associated charge-
carriers generated needs be maximized. Still also, dark current needs to be minimized (Spahn, 2005).

In direct systems an x-ray photon is absorbed by the photoconductor. (See Figure 2) An electron with a high charge is created when this occurs. This electron loses energy as it creates many electron-hole pairs. One absorbed x-ray photon may produce up to a thousand of these electron-hole pairs. A bias voltage is administered to the semiconductor causing electron-hole pairs to form in the substrate layer. Associated pixel electrodes collect these charges (Del Guerra, 2004). Deep trapping is a process that effectively steals away a portion of the charge generated. To minimize lost charge, otherwise known as dark current, a very high voltage of around 10kV is required to create a field of about 10 V/µm. Similar to indirect a-Si detectors, the TFTs are read line by line, with signals sent to electronics. The electrodes are recharged during this process to make them ready for the next exposure. An electric field administered to the photoconductor collects these generated charge-carriers. Every pixel in the a-Se detector array contains a TFT and an electrode. In similar manner to indirect detectors, signals are amplified and converted to a digital signal by a single or multiple analog-to-digital converters (ADC). This signal is then able to be further processed as needed.

Current research is investigating lead oxide, cadmium telluride, mercury iodide, and lead iodide and their uses as photoconductors in these direct systems. Generally though, as previously stated, the photoconductor material of choice is a-Se (Spahn, 2005). By an efficient evaporative process the a-Se semiconductor substrate is deposited onto a transistor array. A-Se is generally used because of its high resistivity, high atomic number, high density, and ability to be placed onto the TFT array in a thick and consistent manner (Del Guerra, 2004).

**Flat-Panel Semiconductors**
With regard to the semiconductors used in both indirect and direct systems, their ability to detect a specific energy range contributes greatly to their selection by engineers and to their performance. Semiconductors allow the manufacturing of very compact systems and provide good energy resolution. Interestingly though, flat-panels have not fully utilized their potential due largely to the difficulty to manufacture chemically pure semiconductor materials. This is dramatically changing, however, especially since the 1990s (Abbene et al., 2009).

Flat-Panel Compared

Inherent difficulties. Comparing different acquisition technologies and communicating differences found in a meaningful way is very difficult. Evaluative numbers do exist, however. These numbers may be used by engineers to further improve a technology, or by an imaging department manager making flat-panel equipment purchasing decisions out of the many available today. (See Table 1)

Comparison values. To follow are some of the more commonly used comparative values, albeit far from exhaustive. The image quality figure (IQF) is a number that attempts to depict image quality. Another is the detective quantum efficiency (DQE) which measures the signal-to-noise ratio (Bassir et al., 2003). DQE is effectively a performance value conveying how efficiently and accurately the input radiation a detector receives is changed into a diagnostic image output. The goal with new or improved imaging technology is to exceed the DQE of existing technologies, thus maximizing the IQF. Additive electronic noise is always engineered to be minimized. Other flat-panel technology design principles that directly influence the final image and need be selected based on the detector’s intended use are weight, exterior dimensions, active area size, pixel size, acquisition rate, and the dynamic range (Spahn, 2005).
**Indirect flat-panel advantages.** There are several advantages to using a-Si as opposed to using direct flat-panel detectors. It has shown stability and resilience to the damaging effects of x-ray radiation for long durations and with numerous exposures. A-Si has characteristics allowing it to be spread over large areas (Feuerbach et al., 1998). The resulting large active matrices of millions of pixels each can be built because plasma can be deposited over large surfaces. Indirect systems generally operate considerably better under lower temperatures (Feuerbach et al., 1998). DQE values favor indirect systems. With comparable thicknesses of CsI and a-Se, the selenium would absorb fewer photons due to its lower atomic number. Also, CsI detectors in a-Si indirect systems have lower additive noise compared to direct systems (Bassir et al., 2003). Still also, by indirect systems having the scintillation step, what results are two effective steps involved that are able to be adjusted independently for optimal performance. There is also higher expected detection efficiency for indirect-based units compared to direct-based (Del Guerra, 2004). With iodide and cesium having high atomic numbers that allow for very high x-ray absorption, indirect detectors have an advantage over direct systems when performing fluoroscopy, angiography and general radiography. This is due to the better handling of a wider range of peak kVp ranges (i.e. 45 to 120 kVp) compared to a-Se with a lower atomic number (Spahn, 2005).

**Direct flat-panel advantages.** Current a-Se direct flat-panel systems have benefits over indirect systems as well. Direct detectors have the benefit of higher spatial resolution. This is largely due to them not requiring an intermediary light-producing step. Distorting characteristics occur when light spreads and scatters in the indirect systems. A-Si indirect detectors try to limit this spread of light by limiting the thickness of scintillator layers in order to maximize detection efficiency as well as maximize the light used. Another benefit of a-Se is that it coincides far
better with numerous frame applications like cine-radiography. This is due to the slower time response associated with a-SI units, which may be as long as a few seconds (Del Guerra, 2004). Still another benefit is a-Se and its low atomic number. A-Se is very well suited for mammography because of this, as a 250 µm thick sheet of it is plenty for the kVp ranges seen in mammography. It should be noted that mammography is not restricted to direct systems as CsI allows for adequate mammography with flat-panels using the typical kVp range of 25-30 kVp (Spahn, 2005). A-Se has also been used in xeroradiography for some time now and is currently beginning to be used extensively in megavoltage x-ray detectors (Del Guerra, 2004).

Comparative Study

It is very difficult to compare direct and indirect digital technologies because the systems are so complex and phantoms do not act as real patients in studies. In a study, a Philips direct digital photodiode detector was compared to a GE cesium iodide/amorphous-silicon detector. The Philips used a selenium drum coated with a-Se, and the GE used pixel elements each consisting of an associated TFT and photodiode. In an attempt to normalize the two different technologies, the direct selenium system needed 2, 1.1, 0.8 and 0.5 mAs while the ScI indirect system needed higher values of 2.5, 1.6, 1.25 and 0.8 mAs to achieve the same phantom entrance doses. (See Table 2) At higher doses of radiation, both systems were very comparable looking at contrast-detail curves. At lower doses however, the ScI system showed superior performance over the selenium detector. IQF numbers showed better quality for the GE system at low dosages but equal numbers at higher dose settings. For the Philips system, IQF numbers appear to be dose-dependent, while they were largely constant for the GE system. DQE numbers showed more desirability for the GE system. It is unclear if the results of this study show differences
large enough to be visibly seen by the human eye but the data suggest indirect is the way to go for slightly lower dosage alone (Bassir et al., 2003).

Discussion

Digital x-ray-radiation flat-panel detectors are highly engineered to provide the best possible images for the diagnosis of injury and disease. A-Se direct capture and a-Si indirect capture flat-panels have clear pros and clear cons when compared. However, comparisons are only as good as the experimental designs used to compare. In any case, improvements to these technologies not only advance the technologies themselves. Other digital acquisition devices, not mentioned here, are pushed to outperform advances in competing technologies. For example, a prototype mammography photon counting detector recently outperformed a charge integrating flat-panel, similar to the ones discussed here, reducing dose by as much as fifty percent (Ducote, Le, & Malloi, 2010). New semiconductor materials are being manufactured and existing ones refined. Because of all of this, existing exams are improving and new exam possibilities are being explored. As long as costs can be kept reasonable and informed managers can use information to pick the imaging equipment that best suits their needs, patients will see unsurpassed image quality with lower radiation exposures.

Table 2. Entrance doses measured at differing mAs settings. Similar dose readings are in bold for comparison. Reprinted from “Comparison of indirect CsI/A:Si and direct a: Se digital radiography: An assessment of contrast and detail visualization,” Bassir et al., 2003, Acta Radiologica, 44(6), p. 618. Copyright 2003 by Acta Radiologica.
Figures


References


