John Yorkston, PhD

Digital Radiographic Technology[1]

In recent years, the digital revolution in projection radiography has slowly been gaining momentum. The combination of increasingly affordable computer power, low-cost memory, widely available high-bandwidth data-transfer infrastructures, and a more digitally accepting user population has helped to spur this advance. One essential component of a usable digital imaging system for general radiography is a robust digital detector with imaging performance that equals or surpasses that of conventional screen-film systems commonly used in radiology departments worldwide. There are a number of candidates for such a detector, which include *(a)* storage phosphor technology (described in the previous chapter by Ralph Schaetzing, PhD), *(b)* digital systems that are based on charge-coupled devices (CCDs), and *(c)* systems that are based on amorphous silicon (a-Si:H) flat-panel detectors. Both the CCD-based and a-Si:H–based systems have become known to the wider medical imaging community as digital radiographic (DR) detectors, and this chapter assumes the same definition.

To understand one of the main clinical advantages of DR systems, it is useful to look at the more traditional methods of x-ray acquisition. From the perspective of a user, image acquisition with a screen-film or storage phosphor system can be divided into two stages: *(a)* the exposure of the imaging plate to x rays and *(b)* the transfer of the film or phosphor plate to a separate reader that extracts the image information. Feedback to the technologist regarding the acceptability of the image can be delayed by many minutes, which may compromise the efficient use of the x-ray equipment. The work-flow efficiency of this acquisition process can be improved by using a DR detector that is self-scanning and can generate the digital image quickly without user intervention.

A number of different approaches designed to achieve this goal have been introduced into the marketplace in the past decade. One approach is to use single or multiple CCD cameras to view the output of a phosphor screen. These systems have been available for a number of years and have been applied in such areas as megavoltage imaging, mammography, and general radiography. The CCD-based approach will be discussed briefly, but the main focus of this chapter will be the recently introduced, commercially available digital x-ray imaging systems that are based on amorphous silicon (a-Si:H) flat-panel technology. These compact devices have the potential to improve the work-flow constraint imposed by traditional screen-film and storage phosphor systems while improving the image quality of the final radiograph.

One alternative approach to a fully electronic readout configuration is a new arrangement of the traditional storage phosphor system that incorporates imagewide

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solid-state line scanning within a dedicated unit (the Fuji Velocity CR system, available only outside the United States at this writing). From a user's perspective, this system functionally achieves fast image capture (cycle time, ~10 seconds) with no user interaction other than firing the x-ray exposure. The core technology behind this system is essentially identical to that described in the previous chapter on storage phosphors by Ralph Schaetzing, PhD, and thus will not be discussed further here.

A future technology that also deserves mention, because it may eventually rival the current generation of DR detectors, is the photon-counting approach being pursued by various companies (Sectra Imtec AB, Linköping, Sweden, among others) (1). This method counts individual x-ray photons and has the potential to provide the highest-quality image of any digital system, along with the possibility of dose reduction and excellent scatter rejection. Its implementation, however, is extremely challenging technologically, and the first clinical research systems have only begun to be evaluated in mammography. They will therefore be excluded from this review, although they are an exciting prospect for the not-too-distant future.

To understand the fundamental configurations of the different types of DR systems currently available, it is useful to divide the x-ray image formation process into three stages: *(a)* the intensity modulation of the incident x-ray field by the patient, *(b)* the x-ray detection by a material that absorbs the incident x rays and produces a response with an amplitude that is related to their intensity, and *(c)* the measurement of this response. The first stage is common to all x-ray imaging systems, old and new. In a DR detector, the second stage involves a phosphor (such as gadolinium oxysulfide [Gd_2O_2S] or cesium iodide [CsI]) or a photoconductor (such as amorphous selenium [a-Se]). The third stage is achieved by optically or electrically coupling the x-ray absorber to an electronic device that quickly measures the response of the absorber. This method can involve either a CCD or an a-Si:H flat-panel detector. The imaging performance of a DR detector is therefore determined by the capabilities of both the x-ray detection material and the CCD or flat-panel detector.

For both CCD and a-Si:H–based systems, the x-ray detection materials commonly used are not particularly new, although they continue to undergo improvements. The real step forward is the ability to record their output in an automatic quasi-real-time manner with a compact device that has no mechanical moving parts (other than the traditional Bucky mechanism). In addition, the efficiency with which the third-stage detector records the output of the x-ray absorption material is one of the most important determinants of the fundamental imaging performance of the different DR system designs. This issue will be revisited later.

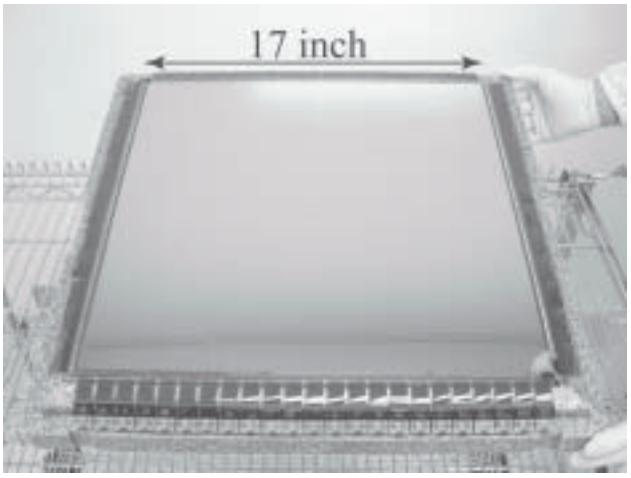


Figure 1. Single-piece 17 × 17-inch a-Si:H array with 150-μm pixels. This array is coated with a layer of a-Se approximately 1,000 μm thick. (Image courtesy of B. Polischuk, PhD, Anrad Corp, Montreal, Quebec, Canada.)

The next section of this chapter describes a few important features of the different x-ray detection materials used in currently available commercial DR systems. The design and performance of CCD-based systems is then discussed briefly. The rest of the chapter deals with the fabrication, pixel design, and operation of the new flat-panel x-ray detectors that are based on amorphous silicon technology. The fundamental reasons for their improved image quality are discussed, along with some differences among the various designs currently available in the marketplace. The requirements for a complete imaging system will be described, as well as the reasons why a-Si:H–based systems are having a profound effect on advanced clinical applications. Most of the chapter deals with general projection radiography, but other applications, such as fluoroscopic imaging and mammography, are mentioned where appropriate. The chapter concludes by reviewing some of the new developments in a-Si:H technology that may improve the performance, reduce the cost, and increase the robustness of flat-panel DR detectors.

X-RAY DETECTION MATERIALS AND IMPROVED IMAGING PERFORMANCE

DR detectors have been classified into two types, depending on the choice of x-ray absorber. Systems that use a phosphor as the x-ray detection material are known as *indirect detectors*, while those that use a photoconductor are known as *direct detectors*. The rationale for this nomenclature is that the systems that use phosphors convert the x-ray energy into electrical charge indirectly through an intermediate stage of light photons, while those that use photoconductors convert the x-ray energy directly into electrical charge

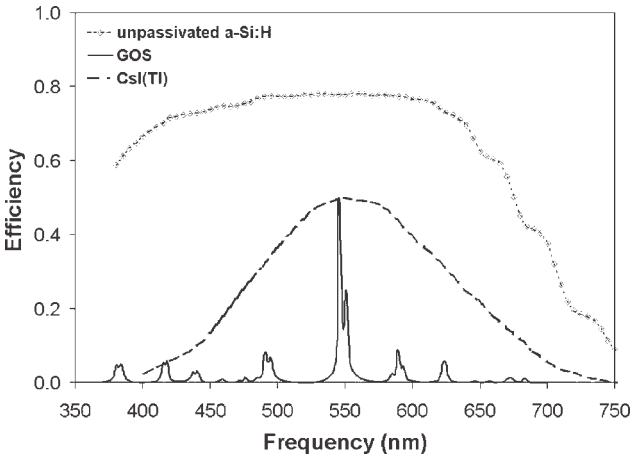


Figure 2. Absorption efficiency of unpassivated a-Si:H plotted along with the output spectra of CsI:Tl and Gd_2O_2S:Tb *(GOS)*. The a-Si:H response is well matched to both phosphors. (Data on a-Si:H absorption courtesy of R. Weisfield, PhD, dpiX, Palo Alto, Calif.)

Figure 3. Configuration of a DR detector that is based on an a-Si:H flat-panel array, showing the protective housing, the x-ray absorber (in this case, a-Se) in direct contact with the flat-panel array, and the peripheral electronics on the edge of the glass substrate. (Image courtesy of K. Schwartz, Direct Radiography Corp, Newark, Del.)

without the intermediate stage. The fundamental quantity of measurement in both approaches is electrical charge, but differences in how this charge is produced have important consequences for system design and performance.

One advantage of the new a-Si:H flat-panel x-ray detectors is that quantitative measurements of their imaging performance (see the next chapter, "Performance of Digital Radiographic Detectors: Quantification and Assessment Methods," by Ehsan Samei, PhD) indicate that such detectors should provide extremely high image quality data from the incident x-ray distribution. It is interesting to ask why this should be the case. As previously mentioned, the commercially available indirect flat-panel detectors use either CsI:Tl or Gd_2O_2S, while the direct detectors use a-Se. The medical imaging community has used these materials to detect x rays for many years. Gadolinium oxysulfide is the material still used for a large fraction of the intensifying screens in conventional screen-film systems, CsI has been used in image intensifier systems for many decades, and selenium was the detection medium used in xeromammographic systems. Amorphous selenium was also the basis of a commercial product used for chest imaging (Thoravision; Philips Medical Systems, Bothell, Wash), which was available until recently (2).

Why do these new x-ray detectors have such improved imaging performance compared with their predecessors? One important reason is that the a-Si:H arrays can be made on extremely large monolithic substrates (see Fig 1, which shows a single-piece, 17 × 17-inch a-Si:H detector coated with a contiguous layer of photoconductor). The arrays have the same physical area as the phosphor or photoconductor layers that interact with the x-ray field. One consequence of this large area is that the arrays are efficient at capturing the output of the overlying x-ray detection medium, whether it is a phosphor or photoconductor. In addition to this geometric consideration, for indirect systems there is an excellent match between the absorption efficiency of a-Si:H layers and the spectral output of the phosphor layer (Fig 2). Current indirect detector designs can capture about 50% or more of the light output from the phosphor layer, while direct detectors claim more than 98% collection efficiency for the charge generated in the photoconductor layer. Compared with about 1% efficiency for film in a traditional screen-film system or compared with the efficiency of less than 10% for the collection of light emitted from a storage phosphor screen, the flat-panel array is a highly efficient collector for the output of the x-ray detection layer. This efficiency helps to maintain the image quality through this stage of the imaging chain.

In a DR detector, the detection medium and the sensor (the CCD or a-Si:H array) are permanently enclosed inside a protective housing (Fig 3). This physical protection allows improvements that increase the imaging capabilities of the system. Optimizing the phosphor content of the Gd_2O_2S layer by reducing the fraction of binder material and removing the protective overcoat enhance the performance of the layer. These modifications are not feasible with traditional Gd_2O_2S screen-film configurations because the phosphor layer must be physically robust to survive continual abrasion from the insertion and extraction of the film from the cassette (the same is true for conventional storage phosphor systems).

This requirement for robustness is also a major problem for the relatively malleable and highly hygroscopic CsI:Tl phosphor layers and probably explains why this type of material (actually CsI:Na) has so far been limited to use in enclosed image intensifier systems. The

physical protection afforded by the detector housing allows the use of CsI:Tl as the x-ray detection medium. This is extremely important because of the tendency of CsI to grow in columnar structures, which mitigates the traditional trade-off between thickness and spatial resolution that plagues powdered phosphor–based layers such as Gd_2O_2S. The need for a thick layer of phosphor material to absorb as much of the incident x-ray beam as possible is normally offset by the reduction in spatial resolution caused by the increased amount of light scattering in the thicker powdered layer. The light-piping effect of the CsI:Tl needle structures reduces but does not completely eliminate this correlation between thickness and resolution. As a result, CsI:Tl layers that are at least approximately 500 μm thick have acceptable spatial resolution and excellent x-ray absorption properties. The higher effective packing density of the CsI:Tl layer also improves its x-ray absorption compared with that of powdered phosphor layers. Figure 4 illustrates these differences between the structure of powdered phosphors and needle-type CsI:Tl layers. The excellent x-ray absorption of thick CsI:Tl markedly improves the imaging performance of indirect systems that use it (3,4).

In the direct approach, the photoconductor layer must be "energized" by the application of an external voltage to make it sensitive to incident x rays. One consequence of this externally applied electric field is that the spatial resolution of the photoconductor is essentially independent of the thickness of the a-Se (5). The electrical charges created by the x rays are guided to the nearest pixel by the internal electric field lines, regardless of the thickness of the a-Se layer. Layers as thick as 1 mm have been reported in the literature (6), with spatial resolution determined roughly by the size of the individual pixels.

Although the previous arguments are certainly not exhaustive, they indicate some of the main reasons for the improved imaging capabilities of many current DR detectors. The physical protection provided by the DR housing allows optimization of the x-ray detection layer. This improves the extraction of the imaging information from the incident x-ray field. Furthermore, the large sensitive area and excellent spectral absorption match provided by a-Si:H technology allow efficient collection of this image information. This improvement in image quality, together with the real-time capabilities of the electronic readout, results in DR imaging systems that satisfy many, if not all, requirements of a usable clinical system.

CCD-BASED DR TECHNOLOGY

The configuration of a DR system that is based on CCD (or complementary metal oxide semiconductor) technology is conceptually simple. A CCD sensor is positioned to image the output of the light from an overlying phosphor layer. All currently available CCD-based DR imaging systems are of the indirect type. Because the operation of a CCD will be familiar to most readers, it will not be discussed further here. Informative descriptions of their capabilities and "bucket brigade" readout schemes can be found elsewhere (7).

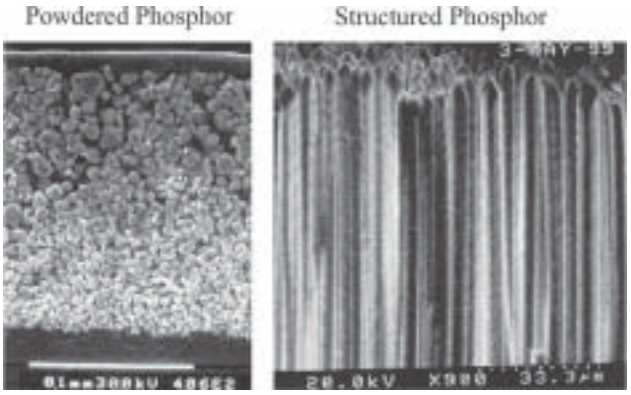


Figure 4. Scanning electron microscopic images. Left: Image of the structure of a powdered phosphor (Gd_2O_2S) shows the random nature of the phosphor particles, which causes considerable light spreading and resolution loss. Right: Image of the needle structure of a CsI:Tl phosphor, which helps to maintain the spatial resolution even for thick layers. Note the different scales of these two images. (Image at right courtesy of R. Weisfield, PhD, dpiX, Palo Alto, Calif.)

Because of the difference in physical size between the imaging area of a clinical phosphor (~35 × 43 cm for general radiography) and the active area of currently available, reasonably priced CCDs (currently limited to ~5 × 5 cm), different techniques involving mirrors, lenses, and fiberoptic components must be used to minimize the phosphor output image to the size of the imaging area of the CCD. A fundamental limitation of this minification is that the efficiency of light collection from the phosphor is extremely low (potentially <0.1%) (8,9), which results in what is known as a secondary quantum sink at this stage of the imaging chain, especially for lens-coupled CCD detectors. The consequence of this unavoidable limitation is reduced image quality. Independent quantitative measurements under clinically relevant conditions indicate that the performance of these systems can be inferior to that of traditional screen-film and storage phosphor systems (9). In addition, the image reduction optics require physical space, which increases the thickness of the detector housing and can cause problems for retrofitting these detectors into existing systems.

Notwithstanding this performance limitation, a number of commercially available systems use a single CCD. The IMIX 2000 system (Oy Imix AB; Tampere, Finland), the DX2000 (Wuestec; Mobile, Ala), and the Xplorer 1700 (Imaging Dynamics; Calgary, Alberta, Canada) are a few examples. In addition, mammographic systems that are based on CCD sensors have

been developed and are commercially available (eg, the SenoScan; Fischer Imaging, Denver, Colo). The smaller size of a typical mammographic image (~25 × 20 or 30 × 25 cm) eases some of the demagnification requirements, reducing the effect of the secondary quantum sink caused by this minification step.

To improve the performance limitation imposed by the small area of the CCD, systems that use more than one CCD have been developed. One commercially available family of products (ddR; Swissray, Hochdorf, Switzerland) uses four high-performance CCDs in conjunction with four high-quality lens arrangements to view overlapping quadrants of the input phosphor. This reduces the demagnification requirements and should improve the overall image quality produced by this system, but the results of quantitative performance evaluations have not yet been reported. The approach pursued by another manufacturer (Clarity 7000; Cares Built, Keyport, NJ) is to use a large number of smaller complementary metal oxide semiconductor sensors (actually a 20 × 20 two-dimensional array) placed close to the phosphor, with each viewing a small fraction of the output area. Although this design has the advantage of being extremely compact, the results of quantitative evaluation of its imaging performance have also not yet been published in the peer-reviewed literature.

In conclusion, although the currently available DR systems that are based on CCD technology are all affected to varying degrees by the low collection efficiency of the light output from their phosphor, they still provide prompt, clinically useful information to radiologists. For this reason, along with the numerous other benefits of digital acquisition, these DR systems remain attractive to certain cross-sections of the clinical community and will undoubtedly remain in the marketplace for a number of years to come. The next sections of this chapter describe in more detail the technology behind a-Si:H flat-panel detectors, how they are made, their individual pixel components, how they operate, and how they are configured into a complete imaging system.

AMORPHOUS SILICON–BASED FLAT-PANEL DR DETECTORS

Of all of the different approaches currently being pursued to configure a clinically usable DR imaging system, perhaps the most promising are those that use amorphous silicon (see references 10–12 for more details on a-Si:H technology and device performance, and see reference 13 for details of its application in radiology). Amorphous silicon technology is well established in the display industry and has been developing during the past couple of decades. The commercial interest in large-area display devices made from a-Si:H has driven a multibillion dollar investment into understanding the fundamental material properties of a-Si:H, improving its device performance, and creating fabrication facilities that can produce thousands of devices per day. The main focus of the commercial investment is the creation of flat-panel displays for laptop and desktop computer applications. Most of the world's laptop computers currently use active-matrix liquid crystal displays that are based on a-Si:H technology. The desire to reduce the fabrication costs of these a-Si:H displays has driven the construction of facilities that can create extremely large-area devices, with substrates approaching 1–1.5 m. Fortunately for the much smaller medical imaging market, this huge investment can be leveraged to produce devices suitable for medical x-ray imaging applications.

ARRAY FABRICATION

As previously described, an a-Si:H flat-panel x-ray detector is made up of an x-ray detection medium, either a phosphor or photoconductor, physically coupled to an a-Si:H flat-panel array. The a-Si:H array itself is fabricated by using a technique known as plasma-enhanced chemical vapor deposition. In this process, layers of amorphous silicon are deposited onto a thin glass substrate (typically ~0.7 mm) from a plasma. This process allows the deposition to be performed over extremely large areas. The addition of various different types of impurities controls the electrical properties of the different layers in much the same way as doping is used in conventional crystalline silicon fabrication. Hydrogen is also used to "neutralize" many of the dangling bond defects that occur in silicon layers created with this technique. The concentration and location of these dangling bonds determine many of the characteristic electrical properties of a-Si:H devices. The relatively high density of dangling bond defects in a-Si:H material also imparts high resistance to radiation damage, which is a particular advantage in high-dose applications such as megavoltage imaging.

Other techniques such as photolithographic patterning and etching are also borrowed from conventional integrated-circuit fabrication to create the active components on the array. In addition, a functional array has numerous metallic lines that allow readout and control of the electrical components of the device. Although the different layers are extremely thin (on the order of ≤1 μm), the array structure is actually three-dimensional, with many components (particularly control lines) passing over or under other elements of the array. This means that capacitive coupling between different components on the array is an important issue and a main concern of array designers.

FLAT-PANEL ARRAY DESIGN

A flat-panel array consists of a two-dimensional rectilinear distribution of imaging pixels, which can be

electronically addressed in a line-by-line manner by external electronics. Depending on the design of the flat-panel array and external electronics, this readout can be performed in a fraction of a second. The array design is illustrated in Figure 5, which shows the corner of an array specifically designed for use with an overlying photoconductor but also demonstrates the generic features of all array types. The horizontal and vertical data and control lines are visible, as well as the component parts of the individual pixels.

The construction of these individual imaging pixels is conceptually simple. Each pixel has a switching element and an element for signal sensing and storage. The switching element allows each pixel to be addressed and read out at the appropriate time, while the sensing element is designed to record the output signal from the particular x-ray detection medium of choice. Choices for the switching component are limited to a two-terminal device known as a switching diode (14) or a three-terminal device known as a thin-film transistor (TFT) (15). The device in Figure 5 incorporates a TFT. Both approaches have been implemented in commercial systems. For the indirect approach that uses a phosphor, the sensing/storage element is a photodiode. For the direct approach that uses a photoconductor, the sensing/storage element is a storage capacitor. Both direct and indirect approaches use a-Si:H flat-panel arrays, albeit of slightly different design. The common distinction between amorphous silicon and a-Se detectors is therefore based on a misunderstanding of the actual configuration of a flat-panel detector.

Figure 5 illustrates that each component on the array occupies a certain physical area. There are limitations as to how closely different components can be positioned. This spacing constraint is one of the many competing design rules that must be followed to guarantee a sufficient yield of functioning arrays. This array fabrication yield has a direct effect on the cost of manufacturing the arrays. One consequence of the details of these constraints becomes apparent when an array with small pixel pitch is designed. As the pixel pitch is reduced, more of the area of the pixel is occupied by the nonsensitive components of the array. This reduces the *fill factor* of the pixels, that is, the fraction of the pixel area that is sensitive to the signal from the x-ray detection material, and can eventually have a marked effect on the imaging performance of the array. This is analogous to the effect of low light collection inherent in the CCD approach described previously.

The issue of the fill factor is mainly a problem for the indirect detectors because electric field shaping inside a photoconductor tends to occur between pixels (5). This guides the charge from the bulk of the photoconductor onto the separate pixel contacts in a way that keeps their charge collection efficiency high.

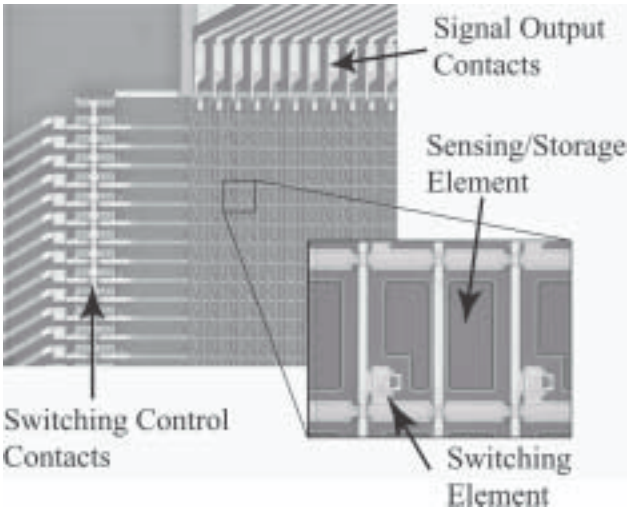


Figure 5. Corner of an a-Si:H array, showing the external contact pads, the switching and signal output connections, and the details of an individual pixel (~150-μm pixel pitch). This array was designed for use with a photoconductor and has a capacitive storage element and a TFT switching element. The vertical line running through the center of the pixel is a ground contact for the back plane of the storage capacitor. Note how the horizontal switching lines "neck down" where they pass over the signal output lines. This helps to reduce the capacitive coupling between these two metal layers and improves the performance of the array.

For applications such as mammography, for which pixels in the range of 50–100 μm are desired, this reduction in fill factor can become a major issue. New approaches, such as the use of continuous rather than pixellated layers of photodiodes, are being investigated but have not yet appeared in commercial products (16,17). In general radiographic applications with 150–200-μm pixels, the reduction in fill factor is not normally a major problem unless the signal levels are extremely low, as in low-dose fluoroscopy (~1 μR per frame [0.258 mC/kg]), in which it is important to collect as many light photons as possible.

As far as the pixel elements themselves are concerned, there are numerous ways to configure components that have the appropriate electrical properties (13). The main difference between them is the complexity of the layer structures used to achieve the desired functionality. More complex fabrication procedures tend to produce better-performing devices, with each pixel component optimized separately. However, the consequence of this increased complexity is usually lower fabrication yield and increased array costs. Alternatively, some arrays are deliberately designed with performance tradeoffs that simplify their fabrication, increase their yield, and, theoretically at least, reduce their cost. Many issues besides imaging performance drive the approach a manufacturer ultimately pursues. These issues include demands of the intended application, intellectual property considerations, and access to particular fabrication capabilities.

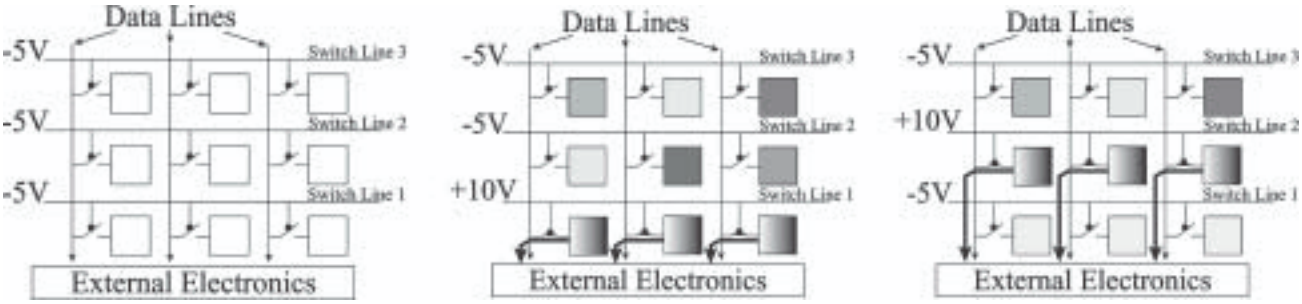
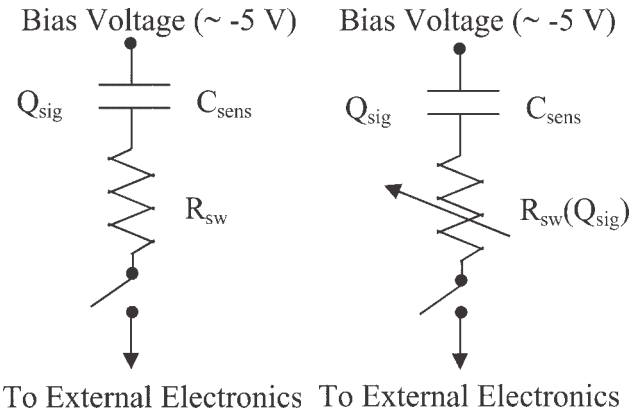


Figure 6. Schematic diagram of a generic flat-panel array readout scheme. Left: The array is in an initialized state ready for the x-ray exposure. Center: The x-ray exposure has been made, and switching line 1 is being read out. Right: Switching line 2 is being read out (see text for details).



a. **b.**

Figure 7. Simplified electrical circuit drawings for **(a)** an n-i-p photodiode coupled to a TFT switching element and **(b)** an n-i-p photodiode coupled to a switching diode element. Variable resistance of switching diode complicates readout of the pixel. C_{sens} = capacitance of sensing element, Q_{sig} = signal current, R_{sw} = resistance of switching element.

ARRAY OPERATION

The readout operation of an array is illustrated in Figure 6. This figure represents a 3 × 3-pixel section of a generic flat-panel array. All switching elements of the pixels along a row are connected to the same horizontal control line, and all signal output contacts from the pixels along a column are connected to the same vertical data line. Once the array has been put into a suitable "initialized" state, all switching elements are held in an "off" state by the appropriate control voltage (−5 V in this example) (Fig 6, left).

The x-ray exposure is made, and the pixels now contain the image information in the sensing/storage element. This image information is then read out one line at a time by changing the control line voltage (to +10 V in this example), such that all of the pixels along a single row become connected to their corresponding data line (Fig 6, center). The signal charge from this row of pixels is transferred along the data lines to the external electronics, where it is amplified, digitized, and stored. The controlling voltage for this line is then returned to the "disconnect" value (−5 V), and the next line is connected by switching its control line voltage to +10 V (Fig 6, right).

This process is repeated for each line of the detector. This line-by-line self-scanning is in contrast to the pixel-to-pixel transfer typical of the readout of a CCD device. For fluoroscopic real-time applications, this readout occurs extremely quickly, with the full detector (~2,000 lines or more) being addressed about every 30–50 msec. Static imaging typically occurs on time scales an order of magnitude slower.

This description of the readout of a flat-panel array is generic and describes the operation of all currently available static and fluoroscopic commercial systems. At the next stage of the readout process, when the array is readied for the next x-ray exposure, the major differences between the different array designs become apparent. The requirements for this reinitialization depend crucially on the design of the pixel components. These differences will be illustrated by two examples.

Figure 7a shows a simplified electrical circuit for a TFT + n-i-p photodiode pixel (*n*, *i*, and *p* refer to the type of doping added to the a-Si:H to create the light-sensitive structure). The photodiode effectively acts as a capacitive element, and the TFT switch acts as a resistor. Readout of this pixel is actually equivalent to recharging a capacitor through a resistor, which occurs with the characteristic time constant *RC*, where *R* is the resistance of the TFT in its "on" condition, and *C* is the inherent capacitance of the photodiode. The transfer of charge from the photodiode to the external electronics is achieved by keeping the switch open for a multiple number of *RC* time constants. In effect, the readout of the pixel is its reinitialization step.

Figure 7b shows the same simplified circuit for a pixel incorporating a switching diode instead of a TFT. For this type of switching component, the resistance of the switching diode is proportional to the voltage across it. This voltage is, in turn, related to the amount of signal charge contained in the photodiode. Because the signal is readout from the pixel, the voltage across the switching diode decreases, causing the switch resistance to increase. Consequently, it takes longer and longer to extract the last fraction of the signal charge

from the pixel. Within a fixed time, the fraction of charge readout from the pixel depends on the initial amount of charge stored in the pixel. This can cause the pixel to exhibit nonlinear behavior to the incident x-ray intensity. The situation is resolved by a complex combination of injected offset charge and application of light to the pixels (4). This process guarantees that the pixels are reliably put into the same initialized state before the next x-ray exposure, which improves the reproducibility of the behavior of the array and reduces the incidence of signal "carryover" from one image to the next. Because this reinitialization is one of the key steps in operating a usable clinical device, the exact details of the procedure are usually a closely guarded trade secret.

For direct systems, reinitialization also depends on the detailed design of the photoconductor layer. The next example illustrates the effect this can have on the reinitialization requirements and operation of an array. Figure 8a shows the simplified electrical circuit for a pixel with a TFT switching element, a capacitive storage element, and an overlying photoconductor. To energize the photoconductor layer and make it sensitive to x rays, a reverse bias of about 10 V/μm must be applied across the layer. For a 500-μm-thick layer of a-Se, this corresponds to +5,000 V of applied bias. Why this is a problem becomes clear when we consider the voltage at point A in Figure 8a. With the array in an initialized state, this voltage is close to ground potential. As the array collects charge with x-ray exposure (or even collection of reverse bias leakage current from the a-Se in the dark), the voltage at A moves toward the applied bias (ie, +5,000 V). However, the voltage of the TFT contact connected to the data line remains at or close to ground, and the TFT gate contact remains at about −5 V. At some point, the voltage between the different TFT contacts will increase beyond its breakdown limit, and the TFT may be permanently damaged.

This tendency for self-destruction is obviously an undesirable feature in an expensive device. To circumvent this problem, an additional dielectric layer is placed on top of the a-Se to act as a third capacitor (Fig 8b). Careful choice of the magnitude of this capacitance can ensure that even when the pixel reaches its saturation charge, the voltage across the TFT will remain within the safe range of operation of the switch. However, this arrangement results in a layer of trapped charges at the dielectric–a-Se interface. This layer of charges needs to be redistributed before the next exposure. (A similar issue must be addressed for the photodiode design of the metal-insulator-semiconductor approach commercialized by Canon [Irvine, Calif] [18,19].) This reinitialization is achieved with a combination of light exposure and manipulation of the applied bias voltage (20).

Reliable reinitialization with this technique may take multiple seconds. Although this is not really an issue for static projection radiography, it is unlikely

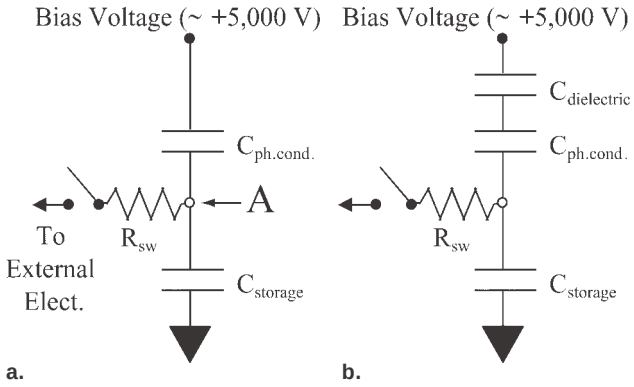


Figure 8. Simplified electrical circuit drawings for a photoconductor with capacitive storage element and TFT switching element. **(a)** Configuration with no high-voltage protection layer. **(b)** Configuration with dielectric high-voltage protection layer. Added capacitance of the dielectric layer ($C_{dielectric}$) complicates reinitialization of the pixel. $C_{ph.cond.}$ = capacitance of photoconductor, $C_{storage}$ = capacitance of storage element, *Elect* = electronics, R_{sw} = resistance of switching element.

that the design could be operated at readout rates sufficient for fluoroscopic imaging. To solve this problem, researchers have developed doped layers of a-Se that allow the application of a negative bias voltage across the photoconductor (21).

This seemingly innocuous change has major consequences. As the pixel collects charge, the voltage of the TFT contact connected to the storage capacitor element is driven toward −5,000 V (note the polarity change). When the voltage of this contact becomes more negative than the applied gate voltage, the TFT begins to turn on, and the excess signal charge can leak through the switch onto the data line. This leakage can affect the signal from other pixels connected to the same data line, but more important, it provides built-in high-voltage protection for the pixels on the array. Another approach is to incorporate a second protection diode in parallel with the TFT switch that essentially provides the same function by letting excess charge from the pixel flow into ground before damage can occur (22). Both approaches have been demonstrated in research devices and allow fluoroscopic operation of the array while providing high-voltage protection for the pixels.

These examples highlight the differences between various flat-panel detector designs. Success in this reinitialization step is crucial in determining the suitability of a detector design for clinical use. All a-Si:H flat-panel detectors exhibit a certain degree of charge carryover, or ghosting, from one frame to the next (23–28). This ghosting can be caused by *(a)* charge trapping and retention in the a-Si:H elements on the flat-panel array or *(b)* sensitivity variations in the photoconductor or phosphor. How well this charge carryover can be accounted for or reduced can determine the usability of a design for a particular application.

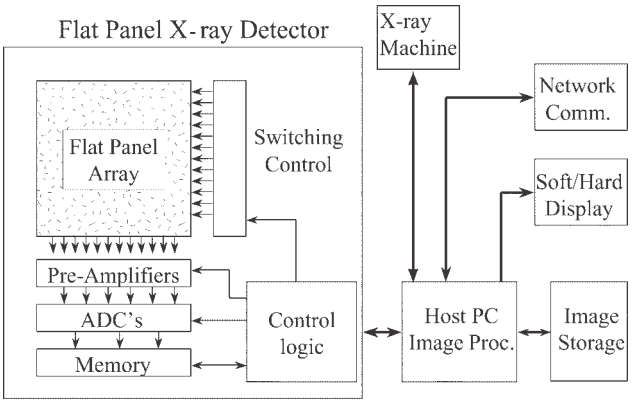


Figure 9. Schematic diagram of the configuration of a complete flat-panel detector imaging system. Synchronization between the x-ray exposure and the array readout allows many advanced acquisition procedures, such as dual-energy imaging and tomosynthesis. *ADC* = analog-to-digital converter, *Comm* = communication, *PC* = personal computer, *Proc* = processing.

DETECTOR CONFIGURATION

As mentioned previously, a flat-panel a-Si:H detector is configured from an a-Si:H array coupled to an x-ray detection medium (Fig 3). In addition, the array must be connected to peripheral electronics that amplify and multiplex the signals from the array, as well as synchronize the switching of the readout control lines and the digitizing of the signals. The noise performance of the front-end amplifier circuits can determine the low-signal performance of the system and is an especially important component of the electronic readout. These circuits are normally designed as application-specific integrated circuits (ASICs) specifically tailored to the signal characteristics of the array to which they will be connected. The use of ASICs that read many channels (typically ≥128) of the array within a single, extremely small piece of crystalline silicon is essential in the creation of a compact device.

Making robust, reliable connections between the closely spaced output contacts on the edge of the array substrate and these ASIC readout chips is a major technologic challenge. During final configuration of the x-ray detector, the peripheral electronics are typically folded beneath the glass substrate to reduce the area surrounding the active region of the detector. The external controlling electronics are then connected, and the whole assembly is enclosed in a protective housing. The detector produced in this way is extremely compact, but thermal management becomes a major issue of detector design. With such tightly confined electronics, thermal drifts can affect the stability of the calibration of the detector and ultimately its imaging performance.

SYSTEM CONFIGURATION

A complete DR detection system incorporates a number of components in addition to the detector itself (Fig 9). Their reliability and performance can have an important effect on the clinical utility of the system as a whole. These components include *(a)* a host personal computer that provides a control interface to the detector system and allows synchronization of the delivery of the x rays and the reinitialization and readout of the array, *(b)* image-processing software to perform image correction and optimization for display, *(c)* appropriate hard- and soft-copy display capabilities, *(d)* network communications to allow distribution of the final image, and *(e)* image storage capability to organize and archive the vast amounts of image information generated by these types of detectors. This whole configuration must work seamlessly if all of the promised productivity gains of the fully digital department are to be realized.

Perhaps the most underappreciated components of this complete system are the image-processing steps necessary to present an optimally rendered image to the viewer. Many aspects of this complex topic will be discussed in the chapter by J. Anthony Seibert, PhD, and in the chapter by Michael J. Flynn, PhD, on "Processing Digital Radiographs of Specific Body Parts," but a few aspects that are specifically relevant to detector performance will be mentioned here.

As with all imaging devices, a-Si:H detectors are not perfect systems, and each component exhibits variations in characteristic behavior that must be accounted for. These include variations in phosphor and photoconductor output caused by imperfections and thickness nonuniformities, as well as pixel-to-pixel response variations caused by switching and sensing element nonuniformities on the a-Si:H panel itself. Careful "flat fielding" and dark-current offset correction techniques can adequately account for most of these nonuniformities (29,30), but temporal variations in array behavior—caused by temperature changes, for example—can cause certain types of artifacts to reappear with undesirable frequency. These artifacts may be a serious issue for system design because they may require recalibration of the system more frequently than is clinically practical.

In addition to the normal statistical variation in pixel behavior, a certain number of pixels and/or lines on an array will inevitably malfunction. Arrays with 100% operational pixels are currently fabricated with such low yield that they would be too expensive for use in normal clinical environments. Single isolated pixel defects can successfully be removed with simple median filtering approaches. Even isolated line defects can be easily corrected with these methods.

When clusters of neighboring pixels or lines are defective, however, it becomes difficult to interpolate

reliably over the defective region. At this point, it is useful to divide these larger defects into two types: *(a)* cosmetic defects and *(b)* diagnostically relevant defects. Although the cosmetic defects are undesirable in an expensive device, they are unlikely to interfere with the diagnostic interpretation of the resultant images. Diagnostically relevant defects, on the other hand, are extremely important and must not be present on clinical systems. Exactly where the distinction lies between these two types of defects depends on the intended application of the detector.

The imaging community as a whole is still coming to terms with this issue, but it may be instructive to remember that all other x-ray imaging modalities exhibit artifacts in a clinical environment. Traditional film is prone to "pick-off," scratches, and processor marks, while storage phosphor plates can also exhibit artifacts caused by screen damage or scanner malfunction. Radiologists have become familiar with the different signatures of the artifacts characteristic of different modalities and can normally diagnose around them (as long as the defects are much smaller than the typical objects of interest). As the medical field becomes more familiar with DR detectors and their characteristic artifacts, the current concern about even small numbers of defective pixels will probably be approached more pragmatically. The concerns of radiologists might also be relieved by allowing them to easily identify where pixels have been corrected and replaced and to determine whether this area is in the region of suspicious clinical features on the image. The issue of pixel defects and clinical acceptability will undoubtedly warrant further discussion as these DR systems become more widely available.

One of the most exciting features of this new technology is the capability to synchronize the detector readout with the delivery of the x rays. The level of control provided by the integration of the detector, the controlling personal computer, and the x-ray machine allows previously impractical procedures, such as dual-energy imaging, cone-beam computed tomography, and conventional tomosynthesis, to be performed almost effortlessly. For example, the switching of beam energy, insertion of appropriate filters, and acquisition of low- and high-energy images can be performed in a single breath hold with computer control. This capability may make dual-energy examinations a realistic option in a busy clinical setting. Coupling this dual-energy soft- and hard-tissue information with software packages for computer-aided detection may advance the detection of many disease indications that are visible with radiographic x rays.

The three-dimensional information that can easily be acquired with a flat-panel system configured in a conventional tomographic setup will undoubtedly provide added benefit for radiologists. The coupling of this three-dimensional information with other modalities, such as ultrasound, is already providing enhanced combinations of imaging data in mammography, for example (31). Image-guided surgery is another area in which the real-time capabilities of flat-panel detectors can provide useful information. They can, for example, help define regions of interest for preoperative planning, enable intraoperative monitoring of the actual surgery, and provide postoperative feedback once a procedure is finished, all from a single imaging device. More details on these and other advanced applications will be given in the six chapters in the syllabus sections, "Digital Radiographic Methods for Tissue Discrimination" and "Digital Radiographic Methods for Depth Discrimination." Figure 10 provides information about some commercially available flat-panel detector systems and a few of the more common CCD-based systems.

FUTURE DIRECTIONS

The development of DR detectors can be divided into two areas: *(a)* the development and optimization of the x-ray detection materials and *(b)* the improvement of the flat-panel arrays themselves. Research into new and improved x-ray detection materials has been ongoing for many years and will undoubtedly continue. New methods of formation of phosphor materials and new types and formulations of photoconductors, such as lead iodide (PbI_2) and mercuric iodide (HgI_2), continue to be reported in the literature, and these hold promise for improving the imaging performance of DR systems even further. However, the most noticeable advances in system performance and functionality will probably come from developments in the fabrication and capabilities of the flat-panel arrays themselves.

One of the main limitations of current DR detectors is their fragility. It is extremely expensive to equip a room with multiple DR detectors in a table and an upright chest system for example. The reason why two separate detectors are needed is because current flat-panel detectors are deposited on a thin glass substrate that may shatter if a detector is dropped during movement from one system to another. This is not the case with screen-film or storage phosphor cassettes and prevents the use of DR detectors for bedside applications. To address this issue of fragility for the display market (and to reduce display weight), researchers have already demonstrated a-Si:H arrays deposited on thin flexible substrates (Fig 11). This development will allow the creation of truly robust bedside DR systems that can survive the demanding clinical environment of a busy modern hospital.

Other researchers are implementing on-substrate amplification and multiplexing circuitry, which may reduce the cost associated with ASICs and other external electronics and help achieve reliable, robust electrical connection to the thousands of peripheral

Array/ Detector Manufacturer.	Product Family	Size¶ (cm)	Pixel Pitch (μm)	No. of Pixels	X-ray Absorbing Material‡	Pixel Components #
a-Si:H Flat-panel Systems (Indirect)						
General Electric/ Perkin Elmer	Revolution XQ/I, XR/d™	41x41	200	2048x2048	CsI:Tl undisclosed thickness	a-Si:H nip photodiode + TFT switch
	Innova 4100™					
	Senographe 2000D™	18x23	100	1800x2304		
Trixell	Pixium 4600™ Siemens: Thorax/Vertix/Multix™ FD, Axoim Aristos™ FX Philips:Digital Diagnost™ Inflmed & Listem: Stingray DR™	43x43	143	3121x3121	CsI:Tl ~550μm	a-Si:H nip photodiode + switching diode
	Pixium 4800™ Philips: Integris Allura™ 9	25 diagonal.	184	960x960	CsI:Tl ~550μm	a-Si:H nip photodiode + TFT switch
Varian/ dpiX	PaxScan™ 2520 Hitachi*	20x25	127	1536x1920	Gd_2O_2S:Tb or CsI:Tl ~600μm	a-Si:H nip photodiode + TFT switch
	PaxScan™ 4030A/R Hitachi: FPD-DR*	30x40	194	1536x2048		
Canon	Canon: CXDI™ 40G Agfa: ADR Thorax™	43x43	160	2688x2688	Gd_2O_2S:Tb ~200μm	MIS photodiode + TFT switch
	Canon CXDI™ 50G (portable)	35x43		2208x2688		
	Canon CXDI™ 31 (portable)	23x29	100	2256x2878		
a-Si:H Flat-panel Systems (Direct)						
Direct Radiography Corp.	Hologic: Epex, Radex™ Eastman Kodak:DirectViewDR™. Fischer Imaging: VersaRad D™	35x43	139	2560x3072	a-Se ~500μm (dielectric HV protection layer)	Storage capacitor + TFT switch
	Lorad: Selenia™ Agfa: Embrace™ Siemens: FFDM-FD™	25x29	70	3584x4096	a-Se ~250μm (dielectric HV protection layer)	
Anrad Corp.	Toshiba: DynaDirect™*	22x21	150	1460x1440	a-Se ~1000μm (p-i-n HV protection layer)	Storage capacitor + TFT switch
		35x35		2304x2304		
	Instrumentarium: Diamond DX™*	18x24	85	2816x2048	a-Se ~200μm (p-i-n HV protection layer)	
Sharp	Shimadzu*	23x23	150	1536x1536	a-Se ~1000μm (also CdZnTe)	Storage capacitor + TFT switch
CCD Based Systems						
SwissRay	ddR™ systems	35x43	167	~2500x2000	CsI, undisclosed doping	4 overlapping CCDs
Wuestec	DX2000™	35x43	120	3072 x 3895	Gd_2O_2S:Tb	Single CCD
Imaging Dynamics	Xplorer 1700™	43x43	108	~4000x4000	Gd_2O_2S:Tb	Single CCD
Oy ImixAB	Imix 2000™	40x40	200	2000x2000	Gd_2O_2S:Eu	Single CCD
Fischer Imaging	SensoScan TrueView™	21x29	54	4096x5624	CsI:Tl	Line scanned TDI CCD
Delft Diagnostic Imaging	Thorascan™	44x44	162	2736x2736	CsI:Tl ~500μm	Line scanned TDI CCD
Cares Built	Clarity 7000™	43x43	60	7000x7000	Gd_2O_2S:Tb	20x20 CMOS array
Star V-Ray	Tradix 4000™	43x43	140	3080x3080	Gd_2O_2S:Tb	8x8 CMOS array
Other Systems						
Sectra Imtec AB	MDM™* (MicroDose Mammography)	24x26	50	not available	10mm thick crystalline silicon	Line scanned Photon counting
Edge Medical	Quix™* DR systems	43x43	120	3420x3420	a-Se, undisclosed thickness	Mechanically scanned line readout

¶ Approximate size, not necessarily active imaging area.
‡ HV = high voltage
# CMOS = complementary metal oxide semiconductor, MIS = metal-insulator-semiconductor, TDI = time delayed integration
* Works in progress

Figure 10. Commercially available detector systems.

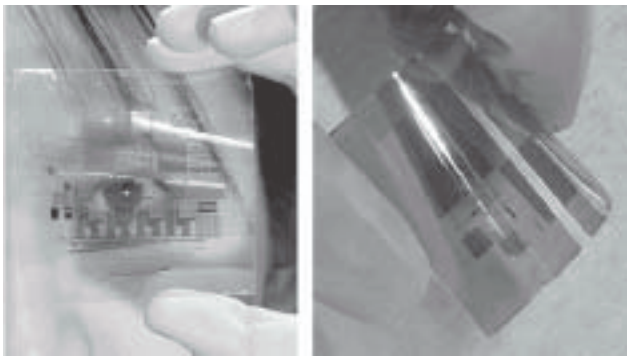


Figure 11. Examples of the use of flexible substrates, which have the potential to improve the robustness of flat-panel arrays. Left: Pentacene organic circuit on polymeric substrate. Right: a-Si:H active matrix gamma ray detector on polyimide substrate. (Images courtesy of T. Jackson, PhD, Pennsylvania State University, University Park, Pa.)

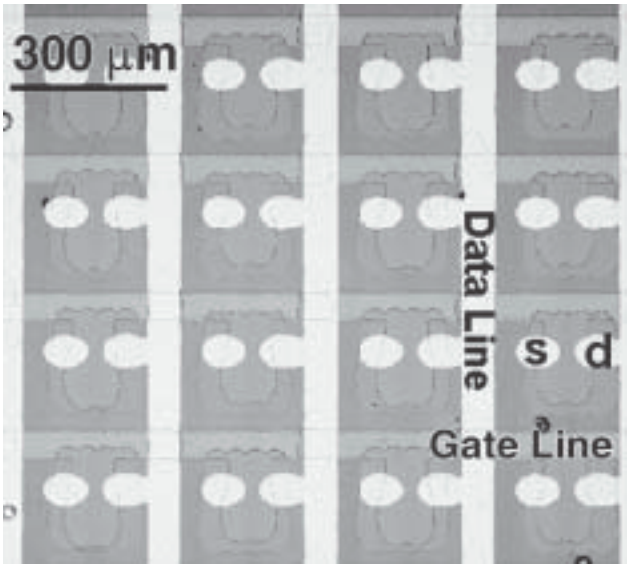


Figure 12. Part of a 64 × 64, 300-μm-pitch, a-Si TFT pixel array fabricated by using jet-printing techniques, which may considerably reduce the future cost of detectors; *s* and *d* indicate the source and drain of the TFT structure. (Image courtesy of R. A. Street, PhD, Xerox PARC, Palo Alto, Calif.)

contacts currently required by today's systems. Pixel-level circuitry has the potential to improve the low-signal-level performance of fluoroscopic systems, which are currently limited by electronic noise from the large data line capacitances characteristic of today's devices. Alternatives to a-Si:H (organic molecules, such as pentacene or naphthacene, and polymeric materials, such as polythiophene) are also being pursued, and their performance levels and stability continue to improve, although they are not ready to replace a-Si:H. Perhaps one of the most exciting but speculative developments is the research into methods of creating complex arrays of electrical components with advanced jet-printing technology. The use of this technique instead of plasma vapor deposition would lead to a major breakthrough in device cost. Figure 12 shows an example of an amorphous silicon TFT switching array that was fabricated by jet-printing techniques and exhibits promising electrical characteristics.

In addition to these generic flat-panel developments, advances in detector configuration specifically targeted toward medical imaging systems have already begun. One recent example is a detector design that incorporates the automatic exposure control capability into the flat-panel array itself (32). This is probably just the first of such developments that will improve the utility and function of the arrays used for medical imaging. Further specialization for specific applications will also occur, with hybrid systems capable of static and fluoroscopic multimode imaging developed and optimized for different clinical needs. Although most of the developments described previously are being pursued primarily to reduce the cost and improve the robustness and performance of flat-panel displays, these developments will undoubtedly lead to major advances in their utility and performance, much to the benefit of the medical imaging community.

In conclusion, DR detectors are already a clinical reality in many radiology departments throughout the world. In the past few years, DR detectors have moved from research curiosities to commercially available systems provided by a number of different companies for applications ranging from mammography, general radiography, and megavoltage imaging to real-time applications such as cardiology and angiography. The ability of these detectors to provide fast feedback to the technologist with regard to patient positioning improves patient throughput, while their compact size and the lack of moving parts in these detectors make their form-factor suitable for many different applications.

The two most prevalent configurations of DR systems are the coupling of either a CCD or an a-Si:H flat-panel array to a traditional x-ray absorber with properties that have been optimized for the specific detector and clinical application. The high efficiency of the a-Si:H arrays for collection of the output signal of the x-ray absorber is the main reason their performance surpasses that of the CCD-based approaches. This high efficiency is a consequence of their large area, which matches the anatomy of the patient and removes the need for image reduction and its accompanying limitations.

The DR systems that incorporate a-Si:H flat-panel arrays are divided into two different types: those that use phosphors and those that use photoconductors. Both designs provide better image quality than the more traditional screen-film and storage phosphor systems, and each has its benefits and limitations. It is likely that both approaches will continue to coexist, with each

having a number of applications for which its particular capabilities make it the more natural choice. Indeed, the same can be said for storage phosphor technology because there are numerous applications (eg, bedside imaging and imaging in the intensive care setting) in which DR systems cannot currently be used but storage phosphors already provide a solution. However, the introduction of a truly robust DR system, probably one based on flexible substrates, is only a matter of time.

The design of the pixels on all a-Si:H arrays incorporates a switching element and a sensing/storage element. The detailed design of these components can have a considerable effect on the operation of the detector, with the precise reinitialization scheme required after an exposure depending on many factors. The exact details of these reinitialization protocols are difficult to obtain but can limit a specific design to static imaging only. Proper implementation of appropriate refresh procedures is crucial in controlling the level of signal retention from one image to the next.

Research into the clinical implications of the improved image quality provided by these flat-panel detectors is well underway, with numerous articles already published (see Kotter and Langer [33] for a comprehensive review of the current literature). Many of these investigators have compared the detectability of simulated clinical features with different DR, screen-film, and storage phosphor systems. Other investigators have imaged actual patients. Early indications are that the DR systems can achieve equivalent performance at lower patient exposures because of their improved imaging capabilities. Whether this improvement is best used to reduce patient dose or to improve image quality at the "standard" dose will depend on the specific application.

The biggest effect of these new devices may be that they will enable the clinical implementation of advanced applications, such as dual-energy imaging and tomosynthesis. Improvements in array design and application-specific system optimization will undoubtedly bring major changes in many areas of clinical practice in the near future. The extent of the influence of these new devices may well depend on how their costs decline in the coming years and on developments in the flat-panel display marketplace.

We are at the beginning of what promises to be an exciting time for both detector development and clinical implementation of this new technology. Flat-panel detectors may be one of the technologies that will expedite major improvements in the clinical practice of many different specializations, from general projection radiography to advanced real-time applications. Different approaches will be necessary for many of these applications, and the marketplace will undoubtedly continue to evolve rapidly in the coming years.

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