The practice of digital radiographic imaging is poised to undergo dramatic change in the very near future owing to a rapid proliferation of electronically readable x-ray detectors. Although self-scanning, direct-readout digital detectors have been in use since the introduction of the charged-coupled device (CCD) almost 30 years ago, recent advances in manufacturing technology have made possible a new generation of large-area, flat-panel detectors with integrated, thin-film transistor readout mechanisms. The excitement surrounding this new technology is based on two factors—the promise of very rapid access to digital images wherever radiography with stationary x-ray equipment is performed and the anticipation of image quality that exceeds that of both screen-film receptors and photostimulable storage phosphor computed radiographic systems because of improvements in x-ray detector technology.

As digital radiography continues this rapid evolution, it is likely that radiologists will be inundated with information concerning a wide variety of large-area, flat-panel electronic detectors. Unfortunately, it is also likely that there will be a tendency to think of these devices as equivalent, interchangeable commodities because they will be similar in physical size, appearance, and targeted applications. It must be emphasized, however, that important differences exist among these detectors, and differences in digital image quality among the various systems are inevitable and may be quite large. To minimize confusion, therefore, it is important that radiologists have a working understanding of this emerging technology.

In this report, we provide an overview of digital electronic x-ray detectors, including two broad classes of detectors based on thin-film transistor arrays and the older, CCD-based designs. Computed radiographic systems based on photostimulable storage phosphors are omitted from this report because they do not contain integrated readout mechanisms [1]. Our goal is to provide a brief review of the basic methods, designs, and materials used in direct-readout radiographic systems and to emphasize important characteristics that may affect system performance and image quality. The advantages and disadvantages of the different detectors, as well as the important factors that should be considered when performing a critical analysis of these new digital imaging systems, are discussed.

**DIRECT VERSUS INDIRECT CONVERSION**

Electronic x-ray detectors can be divided into two classes—those in which direct methods are used to convert x rays into an electric charge and those in which indirect methods are used (Fig 1). (“Direct conversion” should not be confused with “direct readout,” which is a capability of all electronic detectors.) Direct-conversion detectors have an x-ray photoconductor, such as amorphous selenium, that directly converts x-ray photons into an electric charge. Indirect-conversion detectors, on the other hand, have a two-step process for x-ray detection; a scintillator is the primary material for x-ray interaction. When x rays strike the scintillator, the x-ray energy is converted into visible light, and that light is then converted into an electric charge by means of photodetectors such as amorphous silicon photodiode arrays or CCDs. In both direct- and indirect-conversion detectors, the electric charge pattern that remains after x-ray exposure is sensed by an electronic readout mechanism, and analog-to-digital conversion is performed to produce the digital image.
Recent advances in photolithography and electronic microfabrication techniques have enabled the development of large-area x-ray detectors with integrated readout mechanisms based on arrays of thin-film transistors. Unlike older, CCD-based detectors that require optical coupling and image demagnification (discussed later), thin-film transistor-based, flat-panel systems are constructed such that the pixel charge collection and readout electronics for each pixel are immediately adjacent to the site of the x-ray interactions. Several investigators (2,3) have reported prototypic thin-film transistor-based systems with x-ray-sensitive elements that achieve detective quantum efficiencies of 65% or more; this greatly exceeds the performance of photostimulable storage phosphor systems, which have efficiencies of 20%-35% (4), and of screen-film systems for chest radiography, which have nominal efficiencies of 25% (5), and is comparable to the performance of a commercial selenium-drum digital chest radiographic system (6).

Thin-film transistor arrays are used as the active electronic elements in both indirect- and direct-conversion, flat-panel detectors. Thin-film transistor arrays are typically deposited onto a glass substrate in multiple layers, beginning with readout electronics at the lowest level and followed by charge collector arrays at higher levels. Then, depending on the type of detector being constructed, x-ray elements, light-sensitive elements, or both, are deposited to form the top layer of this "electronic sandwich," and the entire assembly is encased in a protective enclosure with external cabling for computer connection.

**THIN-FILM TRANSISTOR-DIRECT CONVERSION**

Direct-conversion systems based on thin-film transistor arrays are constructed by adding an x-ray photoconductor as the top layer of the electronic thin-film transistor sandwich (Fig 2). Typically, amorphous selenium is used as the photoconductor material because of its excellent x-ray detection properties (6-8) and extremely high intrinsic spatial resolution (> 20 line pairs per millimeter [lp/mm] at 100 keV) (9). Before exposure, an electric field is applied across the amorphous selenium layer through a bias electrode on the top surface of the selenium. As x rays are absorbed in the detector, electrons and holes are released within the selenium, and owing to the electric field within the selenium, electric charges are drawn directly to the charge-collecting electrodes. Pixels are effectively separated by means of field shaping within the selenium layer, and the entire selenium surface is available for x-ray charge conversion. Thus, with properly designed charge-collection electrodes, effective fill factors approaching 100% are achievable (3,10).

Amorphous selenium is well developed technologically; it has been used for decades as a photoconductor in photocopiers and in an x-ray imaging technique known as xeroradiography (11,12). Because selenium is used in its amorphous form, selenium plates can be made by means of evaporation and thus can be made relatively easily and inexpensively.
In indirect-conversion detectors, the x-ray scintillator can be structured or unstructured. Structured scintillators, which typically are crystalline cesium iodide, reduce the spread of visible light; this improves spatial resolution and permits the use of thicker scintillator materials for improved quantum detection.

### THIN-FILM TRANSISTOR-INDIRECT CONVERSION

Indirect-conversion systems based on thin-film transistor arrays are constructed by adding amorphous silicon photodiode circuitry and a scintillator as the top layers of the thin-film transistor sandwich. These layers replace the x-ray photodiode layer that is used in direct-conversion devices. When x rays strike the scintillator, visible light is emitted proportional to the incident x-ray energy. Visible light photons are then converted into an electric charge by the photodiode array, and the charge collected at each photodiode is converted into a digital value by using the underlying readout electronics.

The scintillators used in indirect-conversion detectors can be either structured or unstructured (Fig. 3). With an unstructured scintillator, such as conventional fluorescent screens, visible light that is emitted in the material can spread to adjacent pixels and thereby reduce spatial resolution. To reduce this problem, some manufacturers now use a structured scintillator that consists of cesium iodide crystals that are grown on the detector. The crystalline structure, which consists of discrete and parallel “needles” approximately 5–10 μm wide, behaves similar to a bundle of light pipes and channels most of the signal directly to the photodiode layer. Because light spreading is greatly reduced in a structured scintillator, thicker layers of this material can be used in the detector; this increases the number of x-ray photon interactions and thus the available visible light. Although there are trade-offs and practical limitations to this approach (13), use of thicker layers of structured scintillators greatly increases the potential detectable quantum efficiency that is achievable by using the detector, especially at higher imaging frequencies. For this reason and because of its high x-ray absorption efficiency, most manufacturers of indirect detector systems appear to be moving toward the use of structured cesium iodide scintillator materials.

### CHARGED-COUPLED DEVICES

The basic CCD consists of a series of metal-oxide-semiconductor capacitors that are fabricated very close together on a semiconductor surface. Originally invented by Bell Laboratories in 1969 in an effort to create new computer memory devices, CCDs were quickly adapted for use as photodetectors when their sensitivity to visible light was recognized. Today, CCDs are used in a wide variety of indirect-conversion x-ray imaging devices, including large-area radiographic imaging systems and the familiar image-intensifier television system that is used in fluoroscopy.

The single most salient characteristic of CCDs with regard to digital radiography is that they are physically small—typically 2–4 cm², which is much smaller than typical projected x-ray areas. Because of this, cost-effective CCD-based radiographic systems must include some means of optical coupling to reduce the size of the projected visible light image and transfer the image to the face of one or more CCDs. Some CCD-based systems have an image intensifier that reduces the large x-ray field to the size of one CCD. Other systems are based on an array of CCD cameras, each of which is coupled to a scintillator by a lens or a fiberoptic taper. The lens system substantially reduces the number of photons that reach the CCD; this increases the appearance of image noise (ie, quantum mottle) and degrades image quality. In addition, optical coupling with lenses can introduce geometric distortions, optical scatter, and reduced spatial resolution. Fiberoptic tapers can be used in place of optical systems to mitigate light loss and optical scatter problems to some degree, but imperfections in the optical fiber bundles can introduce structure artifacts on the image. Finally, thermal noise within the CCD itself also can degrade image quality, although this is less of a factor with modern, cooled CCDs.

### SYSTEM CONSIDERATIONS

If recent professional radiology conferences are any indication, numerous digital radiographic systems based on direct-readout, flat-panel x-ray detectors will soon be available commercially. Because all of these detectors are compact, electronically readable devices that offer rapid image capture and direct connection to computers in an imaging network, it may appear to many radiologists that they are equivalent, interchangeable commodities. As indicated previously, this is not the case. Generally speaking, it is our opinion that digital radiographic systems based on thin-film transistor arrays will ultimately provide superior image quality relative to CCD-based detectors, particularly when large-area detectors are required. This is owing principally to the requirement of CCD systems for either image demagnification or a more costly design with many tiled CCDs. In comparing systems based on thin-film transistor arrays, however, it is not yet clear whether direct-conversion or indirect-conversion designs will ultimately provide the preferred solution. It may well be that optimum system selection will depend on the intended application (eg, static imaging or fluoroscopy), although it is premature to make such a determination at this time.

Evaluation and selection of a digital radiographic system should involve a thorough analysis of the complete imaging system, including the x-ray detector itself and the environment in which the...
systems (14), detective quantum efficiency is generally accepted as the best single objective indicator of image fidelity. Detective quantum efficiency combines spatial resolution (ie, modulation transfer function) and image noise (ie, noise power spectrum) to provide a measure of the signal-to-noise ratio of the various frequency components of the image. Higher detective quantum efficiency values suggest greater image quality, and the results should be evaluated at all frequencies to estimate the ability of the image to depict both small and large image structures.

**Spatial Resolution**

Image spatial resolution, often expressed as the modulation transfer function, can vary substantially, depending on physical detector characteristics. The intrinsic spatial resolution of amorphous selenium (with direct conversion) and of structured cesium iodide (with indirect conversion) is higher than that of unstructured scintillators, and the intrinsic resolution of selenium is much higher. Digital images can be processed to alter apparent image sharpness; however, excessive processing can lead to an increase in perceived noise. Thus, the inherent detector modulation transfer function, when expressed as a function of spatial frequency, is not as useful as a measure of overall system performance as is the detective quantum efficiency. (See the Image Quality section that follows.)

**Image Quality**

Although observer studies are the most conclusive methods of determining overall system performance, the difficulty in performing such studies for all clinical detection tasks makes it desirable to find a more suitable physical measurement that gives an indication of overall system performance. Although there is no physical measurement that correlates perfectly with perceived diagnostic quality, and there are unique difficulties associated with physical image assessment in digital systems,
Dynamic Range

One of the principal advantages of digital radiography is that image acquisition and image display are decoupled; this allows the detector to have a wide dynamic range and linear response to x-ray exposure. Radiologists should be aware, however, that for a given analog-to-digital converter (ie, a given pixel depth), there can be a trade-off between the detector's sensitivity range (ie, latitude) and contrast resolution (ie, bits per pixel), and some manufacturers may choose to limit the dynamic range of the detector to maximize the number of gray values on the digital image. If used with an automatic exposure control mechanism (ie, phototimer), however, even detector systems with limited latitude (ie, two to three decades) should have no operational problems in clinical use, because the photodetector will ensure that the exposure is appropriate for the available dynamic range. If no phototimer is incorporated into the system, it is recommended that the detector have a minimum of four decades of latitude, equivalent to the latitude of photostimulable storage phosphor systems, to avoid data loss owing to overexposure or underexposure.

Antiscatter Grid

Any digital imaging system that includes an antiscatter grid has the potential for interaction between the grid lines (ie, lead septa) and the rows of pixels that constitute the digital image. For example, grid line artifacts often manifest in a "corduroy pattern" of lines on the image owing to image aliasing; however, some high-strip-count grids do not leave apparent grid lines on digital images. Artifacts that can be produced by grid interaction are not always obvious and can degrade image quality, so caution and proper selection of the grid is advised.

CONCLUSION

Electronically readable, large-area x-ray detectors that promise rapid access to the image for diagnosis, improved image quality relative to that of screen-film and storage phosphor-based radiography, reduced patient examination time, a reduction of consumable agents, and possibilities for reduced patient exposure will soon be commercially available for digital radiography. While these devices will certainly create many new opportunities in diagnostic imaging, it is critical that radiologists recognize that these new self-scanning digital radiographic systems are not interchangeable commodities and can, in fact, produce images with very different image quality.

References