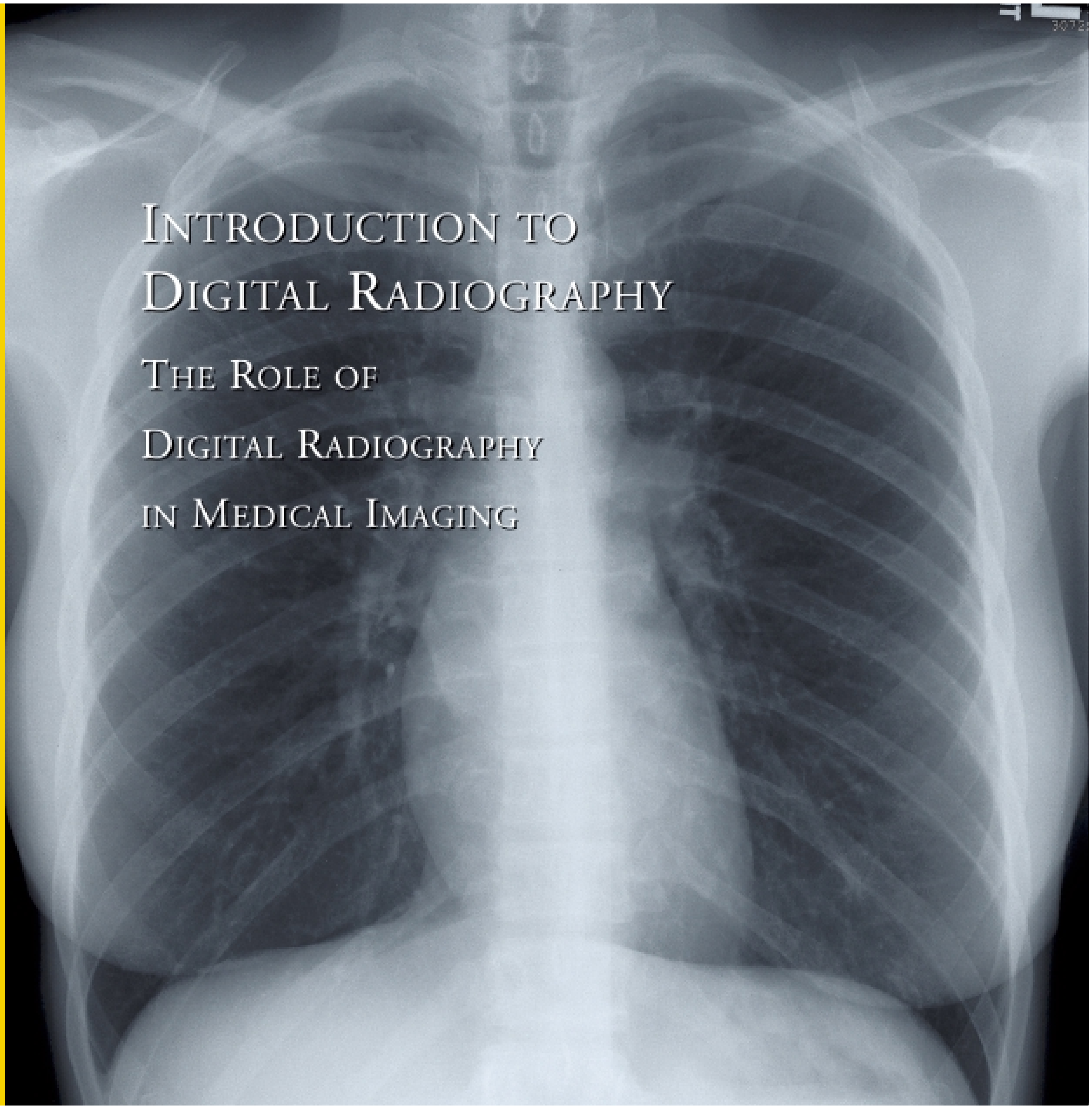

Introduction to Digital Radiography

The Role of Digital Radiography in Medical Imaging

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For nearly 100 years, photographic film has been used to record x-ray images. For over 60 of those years, intensifying screens have been used with x-ray film to provide high-quality images that are the standard for medical imaging because of their image quality, dose efficiency, and functional utility. Once the exposed film has been chemically processed to create a visible image, it can be displayed on a lightbox for diagnosis, easily transported wherever it is needed and kept as an archival record. In modern terminology, x-ray film-screen systems enable radiologists to acquire, display, communicate, and store image data.

Digital imaging modalities, such as computed tomography, ultrasound, and nuclear medicine, gained widespread acceptance in the 1970s. In the 1980s, magnetic resonance imaging, computed radiography, and digital subtraction angiography furthered the trend toward digital imaging. Even so, projection radiography using film-screen technology continues to account for an estimated 65% of all diagnostic examinations.

Recently, it has become technically possible and economically feasible for digital imaging technologies to challenge film for projection radiography. This was made possible by certain prerequisite technological advances, such as high-luminance and high-resolution display monitors, combined with high-performance computer/workstations, which, though still costly, are now readily available. Electronic image archives that can efficiently store and retrieve the massive amounts of image data generated by projection radiography are becoming increasingly cost-effective.

High-speed electronic networks with bandwidth adequate to transmit image files wherever and whenever needed are now accepted as an essential infrastructure component in health care.

Until now, storage-phosphor–based computed radiography has been the best alternative for acquiring digital projection radiography images. Computed radiography (CR) has the advantage of being fully compatible with existing x-ray equipment designed for film-screen imaging. However, it has the disadvantage of requiring readout and processing steps that take about the same time as conventional film to obtain a diagnostically useful image. In the past few years something radically different has entered the medical imaging market, offering a new standard for digital x-ray image capture: large-area, flat-panel, solid state detectors with integrated, thin-film transistor readout mechanisms. Together with the advances in digital display, archiving, and communication technology, digital image capture holds the promise of enhancing the quality and productivity of radiology departments. Experts say this new technology will bring about the most significant x-ray advances since the development of computed tomography (CT) scanners 25 years ago. Digital radiography (DR) systems will enable radiology departments to realize the improved patient care and productivity that the "fully integrated digital department" has long promised.

The excitement surrounding DR technology is twofold: it promises very rapid access to digital images wherever radiography with stationary x-ray equipment is performed, and it can provide image quality that exceeds that of both film-screen and computed radiography. While CR will continue to play an important role in many applications, Kodak DirectView DR systems now offer the most efficient method for obtaining higher-quality digital images with substantial productivity improvements in high-workflow departments.

The Ideal Digital Radiography System

It is useful to compare the attributes of existing DR systems with those of a hypothetical "ideal" DR system (Figure 1). The ideal DR system would produce images of higher quality than the very best conventional systems. It would have high spatial resolution, contrast resolution, and dose efficiency. The detector itself would be robust, solid state, and much like existing film-screen systems in size, weight, and ease of use, while completely eliminating cassette handling. Because the entire system is digital, it would interface well with the hospital information systems and conveniently output images to printers, archives, and workstations.

Although a discussion of the pros and cons of each detector on the market—its size, weight, dose efficiency, and image quality—is beyond the scope of this paper, it is appropriate to compare currently available technologies in terms of the attributes of the ideal digital radiography system.

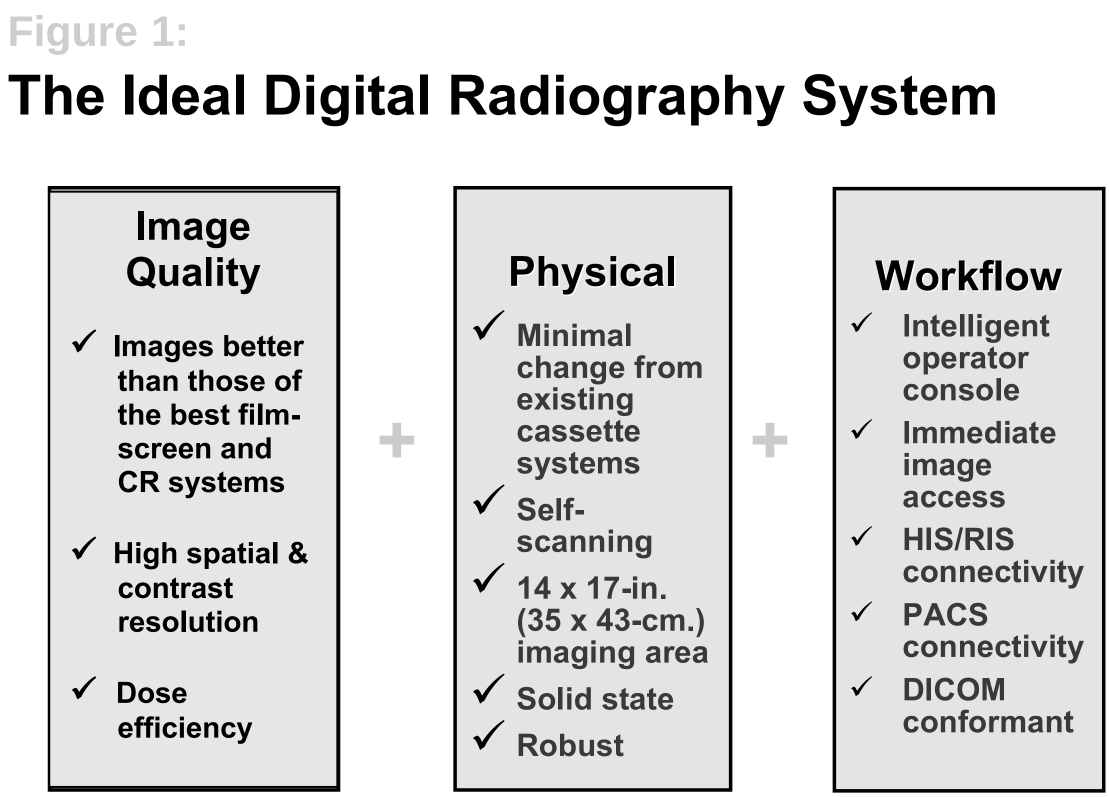


Fig. 1: Characteristics that contribute to the optimal digital system.

Digital Radiography Detectors

Many different digital radiography detectors are now being marketed, with more on the way. There may be a tendency to think of these devices as equivalent and interchangeable because of their similarity in physical size, appearance, and targeted applications. However, important differences exist among the available detectors, both in terms of how they capture the x-ray image and in the resulting image quality. These can be appreciated by distinguishing three distinct digital radiography technologies currently available.

This paper provides a brief review of the materials, design, and performance characteristics of each of these detector technologies. An additional detector technology, computed radiography (CR), based on photostimulable storage phosphors will not be discussed in detail here since it has been described elsewhere. Clinically, CR systems will continue to be an important complement to DR systems for applications in which workflow or examination conditions (such as portable examinations) are not suitable for DR. Where possible, CR and DR systems should provide technologists and radiologists with a common "look and feel."

Digital Radiography Detector Technologies

Application: Static Imaging for General Radiography

CCD* Camera – Indirect

X-ray scintillator / optics / CCD(s)

Gadolinium Oxysulfide / lens / CCD(s)
- Oy Imix
- Oldelft
- Swissray
- Trex
- Weustec
- Imaging Dynamics

Flat Panel – Indirect

X-ray scintillator / photodiode / TFT

Cesium Iodide / a-Si:H photodiode + TFT
- General Electric Medical Systems

Cesium Iodide / Silicon photodiode / Switching diode
- Trixell (a joint venture of Thomson Tubes, Siemens and Philips)

Gadolinium Oxysulfide / Silicon photodiode / TFT
- Canon
- Agfa

Flat Panel – Direct

X-ray photoconductor / TFT

a-Se / Storage Capacitor + a-Si:H TFT
- Direct Radiography Corp (Hologic, Inc.)
- Eastman Kodak Company

*CCD = charge-coupled device.

CCD Camera DR

Charge-coupled devices (CCDs) are now used as the image-acquisition component of cameras in video and digital photography. CCD digital radiography systems use these devices together with minification optics to image the light emitted by a scintillator, most often a conventional intensifying screen designed for film-screen radiography. CCD-based systems are available from a wide variety of manufacturers, including Swissray, Oy Imix, Trex, Odelft, Wuestec, and Imaging Dynamics.

The most important characteristic of CCDs with regard to digital radiography is that they are physically small—typically 2 to 3 cm^2, which is much smaller than typical areas of interest in projection radiography. Because of this, cost-effective CCD-based radiographic systems must include some optical means to reduce the size of the light image emitted by the intensifying screen while transferring that image to one or more CCDs. Lenses and fiber-optic tapers are commonly used. In either case, demagnification is very inefficient. The number of light photons that reach the CCD can be on orders of magnitude less than those emitted by the scintillator. This often results in a "secondary quantum sink," which is simply a signal-to-noise bottleneck that can substantially increase image noise and degrade image quality. Lenses and fiber-optic tapers both generally introduce geometric distortions, light scatter, and reduced spatial resolution, and imperfections in the optical fiber bundles can introduce structure distortions and artifacts into the image. Thermal noise within the CCD itself also can degrade image quality, although this is less of a factor with modern, cooled CCDs.

Most experts believe that CCD-based detectors are a transitional technology and that flat-panel digital radiographic systems will be the preferred source of digital projection images because of their superior image quality, intrinsic robustness, and compact design.

Flat-Panel Technology: Direct versus indirect conversion systems

Flat-panel digital radiographic x-ray detectors can be divided into two classes: **direct conversion detectors,** in which x-ray energy is converted directly into electric charge, and **indirect conversion detectors,** in which x-ray energy is first converted to light by an x-ray scintillator (Figures 2 and 3). The two most common scintillators are cesium iodide (used in x-ray image intensifiers) and gadolinium oxysulfide (used in conventional x-ray intensifying screens to expose film). Note that "direct conversion" should not be confused with "direct readout," which is a capability of all electronic detectors.

Figure 2:
Indirect Digital Radiography Systems

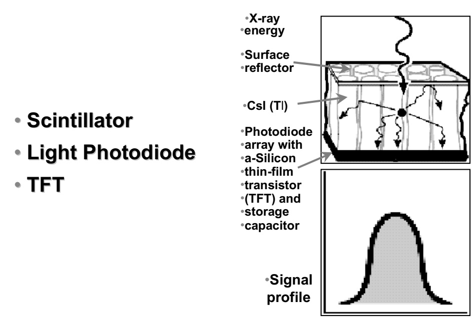
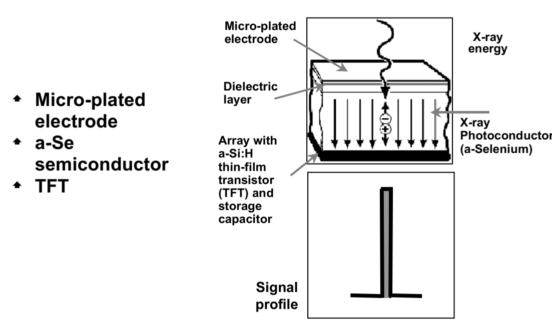


Figure 3:
Direct Digital Radiography Systems



Figs. 2 and 3: Comparison model of typical design and associated signal profile for indirect and direct radiography.

Direct-conversion flat-panel digital detectors, such as that in the Kodak DirectView DR system, use an x-ray photoconductor material, amorphous selenium, to directly convert x-ray quanta into electric charges. This is the simplest and most efficient method, since it requires no intensifying screens, intermediate steps, or additional processes.

Most indirect flat-panel digital radiography detectors use a light scintillator similar to that used in film-screen systems. Conventional film-screen intensifying screens are composed of a fluorescent material which absorbs the incident x-ray energy and converts it to light during an exposure. A substantial fraction of the emitted light never reaches the film. That which does has scattered many times, degrading image sharpness. The light creates a “latent image” on the film that is made visible by processing with photographic developer. The film is then fixed, rinsed, and dried to obtain a permanent archival image.

Like film-screen, indirect-conversion detectors require a two-step process for x-ray detection: a fluorescent material such as gadolinium oxysulfide or cesium iodide is used to capture x-ray energy and convert it to light. Some of the light energy is lost and the remainder scatters, degrading image sharpness before being converted to electronic charge by an array of thin-film photodiodes. Finally, the charge pattern is read out by an array of thin-film transistors (TFT).

X-ray image quality depends on maintaining a high signal-to-noise ratio throughout the entire imaging chain. Preserving the signal profile at the detector is a critical first step in that process. Because indirect-conversion systems rely on light, substantial scatter occurs before the energy is converted to charge. This reduces that amplitude and signal-to-noise ratio of diagnostically important details. The narrow response profile (Figure 3) that characterizes the direct-capture systems results from directly converting x-ray energy into electronic charges. The amplitude of features of all sizes in the detected image is preserved, thereby maintaining the signal-to-noise ratio of diagnostically important details.

In both direct- and indirect-conversion detectors, the electric-charge pattern is temporarily stored by the detector during x-ray exposure. After the exposure, the TFT-switching electronics on the detector direct this charge to amplifiers and analog-to-digital converters that produce the raw digital image. This is called "direct readout" and is an important feature of all electronic detectors.

Digital image processing of the raw image data is then required to produce images suitable for display and diagnosis. The raw image data produced by the Kodak DirectView DR systems covers a wider dynamic range than is required for even the most challenging examinations. As a result, fully automatic image processing is able to produce high quality images even when film-screen systems would have been over- or under-exposed. This, together with highly effective automatic exposure control, virtually eliminates retakes caused by incorrect exposure technique.

Large Area TFT Arrays

Recent advances in photolithography and electronic microfabrication techniques have enabled the development of large-area x-ray detectors with integrated readout mechanisms based on TFT arrays. Unlike CCD-based detectors, which require optical coupling and image demagnification, TFT-based flat-panel systems are constructed such that the charge collection and readout electronics are immediately adjacent to the layer in which the x-rays interact. This allows for a compact design and provides immediate access to the digital images.

TFT arrays are used as the active electronic switching elements in both direct and indirect digital radiography systems. TFT arrays are deposited onto a glass substrate in multiple layers, beginning with readout electronics at the lowest level, followed by charge-collector arrays at higher levels. Then, depending on the type of detector being constructed, x-ray-sensitive elements, light-sensitive elements, or both are deposited to form the top layer of this complex electronic structure. The entire assembly is then encased in a protective enclosure with external cabling for computer connections. As with all electronic devices, the more layers that are needed, the greater the complexity of the fabrication process. This generally lowers yield and reliability.

Direct-conversion detectors use the fewest layers and have the simplest design.

Flat-Panel Direct-Conversion Systems

The Kodak DirectView DR system detector is constructed by adding an x-ray photoconductor layer adjacent to the amorphous silicon thin-film

Figure 4:
Principle of Operation - Direct Detector

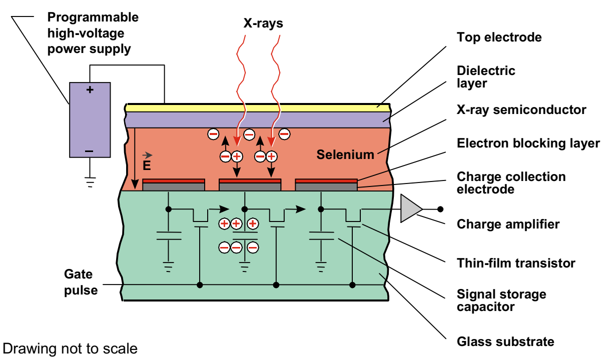


Fig. 4: X-ray capture and operative design of the direct-conversion detector.

transistor and charge storage capacitor (Figure 4). Amorphous selenium is used as the photoconductor material because of its excellent x-ray-detection properties and extremely high intrinsic spatial resolution. 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, electric charges are drawn along the electric field lines directly to the charge-storage-capacitor electrodes. The charge collected at each storage capacitor is amplified and quantized to a digital code-value for that pixel by the underlying readout electronics. Detector elements are effectively separated by electric-field shaping within the selenium layer. Therefore, the entire selenium surface is available for x-ray charge conversion. Charge-collection electrode design results in effective fill factors approaching 100%. Amorphous selenium is well developed technologically; it has been used for decades in photocopiers, photocells, and exposure meters for photographic use, as well as in solar cells. It is also used as a photographic toner and as an additive in the glass and stainless steel industries. Because selenium is used in its amorphous form, large area selenium plates can be made by vapor deposition, a highly reproducible and cost-effective technology.

Flat-Panel Indirect Conversion Systems

Indirect-conversion systems based on flat-panel TFT arrays are constructed by first adding an amorphous silicon photodiode, biasing electrodes, and then a scintillator as the top layer of the detector. These layers replace the single x-ray photoconductor layer that is used in direct conversion devices. When x-rays strike the scintillator, visible light proportional to the incident x-ray energy is emitted in all directions. Some of that light is absorbed in the scintillator while most of the remainder is scattered before reaching the photodiode. Those photons that reach the photodiode array are converted into an electric charge. The charge collected at each photodiode is amplified and quantized to a digital code-value for that pixel by the underlying readout electronics.

The scintillators used in indirect-conversion detectors can be either structured or unstructured. Unstructured scintillators are often simply conventional intensifying screens designed for film-screen radiography. In unstructured scintillators, scattering causes the light to spread laterally over distances equal to or greater than the thickness of the scintillator. This means that the light from a single x-ray can spread to a number of adjacent pixels, reducing spatial resolution. To address this problem, some manufacturers use structured scintillators that consist of cesium iodide crystals that are grown on or joined to the detector. These crystals are highly hygroscopic (they absorb water) and rapidly degrade if not completely sealed. The crystalline structure, which consists of discrete and parallel "needles" approximately 5 to 10 microns wide and as much as 600 microns long, behaves similarly to a bundle of fiber-optic “light pipes.” These help to channel light to the photodiode layer.

Light spreading is reduced, but by no means eliminated, in structured scintillators. In practice, the reduced light-spread allows thicker layers of structured scintillators to be used that have light-spread comparable to conventional x-ray intensifying screens. In this way, structured scintillators provide increased x-ray energy absorption, compared with unstructured scintillators having the same amount of light-spread. Although there are trade-offs and practical limitations to this approach, the use of thicker layers of structured scintillators increases the achievable image quality, compared with that of unstructured scintillators.

Comparing Technologies

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 system will be used. Beyond this, DR systems must provide connectivity to DICOM and HIS/RIS systems as well as to the image processing needed to produce high-quality display-ready images. The following considerations are offered to facilitate the evaluation of electronic digital radiographic systems.

Detector Size. The physical dimensions of the x-ray detector have an obvious influence on the radiographic examinations performed. It is critical that the detector be both sufficiently large to capture the desired anatomic views and sufficiently compact for the desired applications. Electronic detectors for standing radiographic examinations, for example, would ideally have an active detector that is at least 35 x 43 cm (14 x 17 inches) in size and allow both vertical (35 x 43-cm) and horizontal (43 x 35-cm) imaging orientations. With good mechanical design, a rectangular 35 x 43-cm (14 x 17-inch) detector that can be easily rotated may perform as well as a larger square (43 x 43-cm) detector. Smaller square formats, such as 41 x 41-cm designs, risk truncating diagnostically important areas of an image. Rectangular formats also correspond to workstations, monitors, and x-ray printers which normally provide rectangular output of the same size and aspect ratio. Image-capture devices that are square would need to have the image data minified or cropped to meet the requirements of these output devices.

Monolithic Panels versus Tiled Arrays. Because fabrication of full-size TFT-detector panels is technically challenging and can have relatively low yields, many manufacturers reduce costs by constructing detectors that consist of two or more smaller panels in a tiled configuration. In these detectors, digital image processing is used to "stitch together" the image sections to eliminate the appearance of the tile junctions.

Detector Element and Matrix Size. The maximum spatial resolution of an image is defined by detector-element size and spacing (ie, pitch). For example, spatial frequencies above 2.5 cycles per mm will not be properly recorded by a system with a 200-micron detector-element size. The limiting frequency for the Kodak DirectView DR system is 3.6 cycles/mm (139 microns), encompassing all the spatial frequency of interest for general radiographic imaging. This results in images that are 2560 x 3072 pixels. Each pixel carries unique information, since the direct detector has no source of presampling resolution loss. In contrast, the pixel values of indirect detectors are usually highly correlated, which means that each pixel is not carrying unique image information. This results in wasted storage space and network bandwidth. Comparing indirect detectors solely in terms of the number of detector elements can be misleading, since spatial resolution will often be limited by image blurring that results from light scatter in the detector.

Special applications, such as mammography, will require smaller detector-element sizes—probably in the range of 50 to 100 microns per element. However, detectors with significantly smaller detector elements and large pixel matricies have implications in practical implementation, including increased data volume per image, greater network traffic, increased archiving requirements, and the likely need for image reduction when images are displayed on video monitors. Such detectors are inappropriate for general radiography.

Spatial Resolution. Image spatial resolution can vary substantially, depending on physical detector characteristics. As already mentioned, limiting spatial resolution is determined by the pixel spacing in the detector. The frequency that characterizes this limiting resolution is known as the Nyquist frequency. It is simply the inverse of twice the pixel spacing. The vertical discontinuities in the curves in Figure 5 indicate the Nyquist frequencies of the

Figure 5

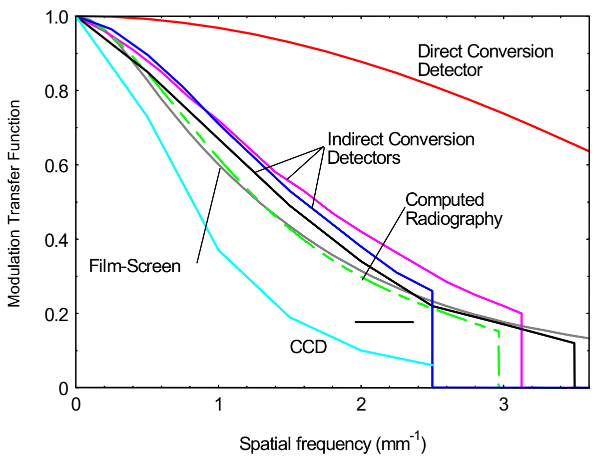


Fig. 5: MTF of Kodak DirectView DR system compared with other indirect digital systems, computed radiography and traditional film-screen systems. NOTE: Presampling MTF of digital systems shown below their respective Nyquist frequencies.

digital detectors. The 139-micron pixel spacing of the Kodak DirectView DR system results in the highest Nyquist frequency. But the Nyquist frequency is only part of the story. The best measure of image blur is the presampled modulation transfer function (MTF). Every image can be described in terms of the amount of energy in each of its spatial frequency components. MTF describes the fraction of each frequency component that will be preserved in the captured image. Figure 5 compares the MTF of the Kodak DirectView DR system with that of state-of-the-art indirect-conversion detectors, computed radiography, a 250-speed film-screen system, and a CCD-camera DR system. The MTF of all current technologies except the direct-conversion detector falls off substantially at higher spatial frequencies. By comparison, the MTF of the direct-detector system remains high up to the Nyquist frequency. The small observed drop in MTF for the direct conversion detector results only from the physical size of the pixel. Higher Nyquist frequency and MTF allow visualization of finer details and provide the basis for better imaging performance. It is well established that the finer the structure the higher the resolution needed for faithful image representations. For applications requiring visualization of fine detail, such as imaging lung parenchyma or trabecular and small bone detail, the Kodak DirectView DR system is superior to film-screen, computed radiography, and all indirect-detector digital systems.

Image blur prior to the sampling increases the "effective size" of the pixels as compared with their physical size. One result of the intrinsically high spatial resolution of direct conversion systems using amorphous selenium is that the physical and effective sizes of pixels are identical. However, for indirect detectors, the effective pixel size can be substantially larger than the physical pixel size. This can be appreciated by considering the "line-spread function" computed directly from the presampled MTF. The line-spread function describes the image blur that would occur for an infinitesimally narrow beam of x-rays incident upon a detector. Figure 6 shows the line-spread function of a state-of-the-art indirect detector. For this system, the pixel size is

Figure 6

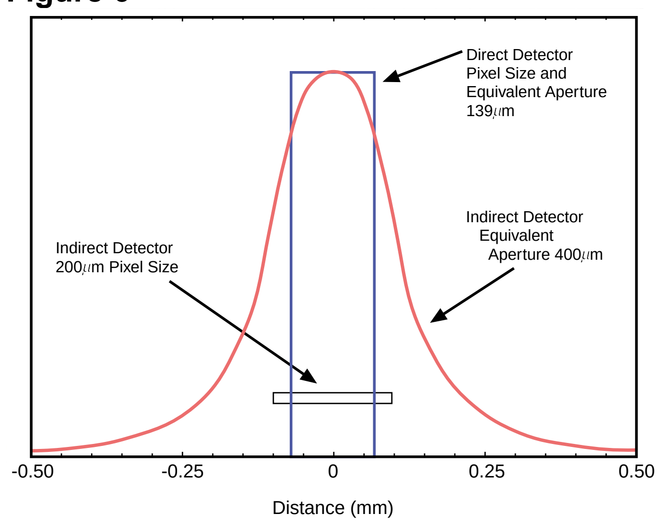


Fig. 6: Comparison of signal profile and line spread function of a direct and indirect detector, demonstrating the increase in image blur that occurs with an indirect detector system.

200 microns. However, the line-spread function is substantially wider. Comparing the line-spread function and the pixel size shows that x-rays incident on a pixel will make substantial contributions to the signal recorded by neighboring pixels. This can also be understood in terms of "equivalent aperture," a useful measure of the effective size of an imaging element first described

by Shade and refined by Burgess. The effective pixel size of this indirect 200-micron pixel is actually 400 microns, or twice its physical size.

By comparison, the line-spread function for the Kodak DirectView DR system detector corresponds precisely to the detector's 139-micron pixel pitch. As a result, the "equivalent aperture" and the physical size of the pixel are the same. Therefore, x-rays only contribute to the signal recorded by the pixel on which they were incident, which means that pixel data is independent and that there is no wasted network bandwidth or archive storage space.

Figure 7 compares the pixel size of the Kodak DirectView DR system detector (actual and effective pixel size are both 139 microns) with a CsI indirect detector in which the actual pixel size is 200 microns, but the effective pixel size is 400 microns. Any increase in effective pixel size is a direct contributor to image blur. Using smaller pixels makes little difference in the effective pixel size, as it is determined primarily by the amount of light-spread in the scintillator.

Figure 7

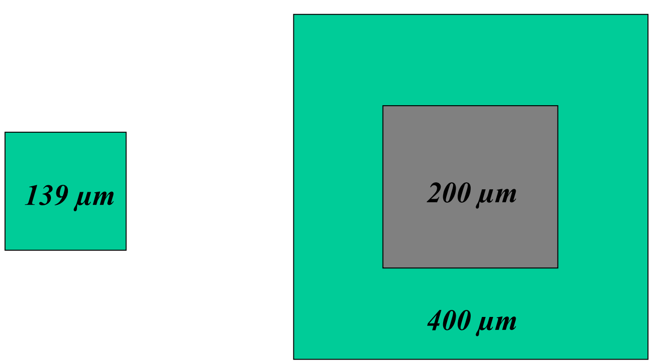


Fig. 7: Kodak DirectView DR system's actual (139 microns) and effective (139 microns) pixel size compared to an indirect system's actual (200 microns) and effective (400 microns) pixel size. This indicates the superior resolution of direct capture.

Image Quality. Observer studies using receiver-operating-characteristic (ROC) methodology are the most conclusive methods of evaluating overall system performance. However, the difficulty in performing such studies for a wide range of clinical detection tasks makes it desirable to find suitable surrogate measurements. Objective physical measurements, such as MTF and DQE are now widely accepted for this purpose. However, physical measurements of the detector alone must be used with caution, since research has shown that clinical imaging tasks are influenced by both clinical and imaging system factors not considered in these measurements. For example, recent experiments have shown that anatomical noise can be a limiting factor in the detection of clinically relevant findings. For some clinical tasks, anatomical noise may be a more direct predictor of performance than signal-to-noise characteristics of the detector. As another example, image processing has been shown to substantially influence even simple detection tasks.

Recognizing these limitations, detective quantum efficiency (DQE) is the best and most widely accepted overall measure of detector image-quality performance. DQE is simply the efficiency with which a detector captures the information present in an x-ray exposure. The information available in any image is limited by the finite number of x-ray quanta incident upon the imaging detector, which in turn is related to patient dose. The information limit results directly from the inherent statistical nature of x-ray quanta. An ideal imaging system accurately records every incident x-ray quantum and is characterized by a DQE of 100%. Real imaging systems, however, always have a DQE of less than 100% because of inefficiencies in detecting the incident x-ray quanta and internal sources of noise. More importantly, the DQE of real detectors does not have a single value, but varies depending on exposure technique (kVp), level (mR), and the spatial frequency. The maximum reported DQEs of film-screen systems and computed radiography systems are typically 15% to 25%. By comparison, the maximum DQE of the Kodak DirectView DR system, measured under similar conditions, is 46%. However, there is more to the story than simply the maximum DQE value, which generally occurs near zero spatial frequency.

Figure 8 shows the spatial frequency dependence of the DQE for the Kodak DirectView DR system detector as well as examples of several

Figure 8

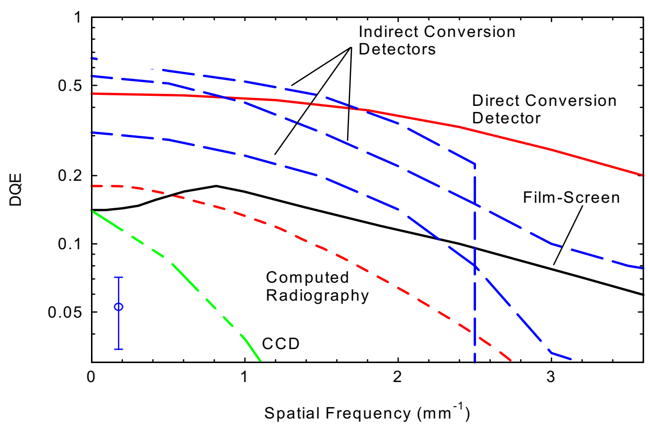


Fig. 8: DQE as a function of spatial frequency compares Kodak DirectView DR with many other available systems. This shows that DQE of the Kodak DirectView DR system remains high throughout the full diagnostic range of resolution, from gross to fine structure imaging.

other technologies. Direct-conversion digital radiography is clearly superior to film-screen, computed radiography, and CCD-camera-based DR. Figure 8 also shows three examples of state-of-the-art indirect-conversion technology. While the DQE of the indirect detector can be slightly higher at the lowest spatial frequencies, the DQE of the direct detector is generally substantially higher at the higher spatial frequencies.

Which frequencies are more important? Research has shown that low spatial frequencies are most important for low-order imaging tasks, such as detecting simple objects in uniform backgrounds. An example of a low-order imaging task would be the scoring of contrast-detail test objects. However, for higher-order imaging tasks that require localizing and characterizing complex objects in non-uniform backgrounds, high frequencies become more important. Many clinical tasks require localization and characterization of small anatomical structures. These are examples of high-order imaging tasks. Having high DQE over the entire diagnostic range of visual information (0.5 to 3.5 cy/mm) is necessary for optimal performance in the wide range of imaging tasks found in clinical practice.

How big a difference in DQE is important? While definitive studies have yet to be done for radiographic images, the smallest detectable DQE difference for pictorial images has been shown to be between 12% and 69%, depending on the complexity of the scene. Average scenes require a 35% change in DQE to produce a "just noticeable difference." The DQE change required for ±1 just noticeable difference is shown as the "error bar" in Figure 8. Comparing the direct and indirect conversion technologies shows that while the differences at low frequencies may not be detectable, at high spatial frequencies the direct-conversion detector is clearly superior.

Several factors strongly influence the DQE of imaging technologies. These include less than complete x-ray absorption, factors which reduce the amplitude of the signal profile (as measured by MTF), and additional sources of noise. Because of the light scatter inherent in indirect detectors, there is a trade-off between x-ray absorption and MTF. Increasing the thickness of the scintillator of an indirect detector may increase the low-frequency DQE, but the high-frequency DQE will be reduced by the lower MTF that results from increased light-scatter. Because of the reduced signal amplitude at high frequencies, it is often necessary to reduce the electronic noise in indirect detectors by cumbersome external cooling devices. Direct-detector technology is not limited by light scatter. X-ray absorption can be increased without loss of MTF. This results in the high-frequency DQE necessary for diagnostically important imaging tasks.

Indirect technologies also suffer several sources of added noise that are virtually absent in direct detectors. The brightness and lateral spread of the light burst produced by an absorbed x-ray quantum depends on the depth of interaction in the scintillator. Quanta absorbed closer to the light-sensing element are brighter and have less spread, while those absorbed closer to the front of the detector are less bright and have greater spread. These two sources of variability add noise to the image and substantially reduce the DQE, particularly at high frequencies. In a direct-conversion system the number of electrons produced and their lateral spread (virtually none) are both independent of the depth at which an x-ray interacts.

In summary, Kodak DirectView DR systems provide the highest DQE at spatial frequencies that affect the ability of the imaging system to render clinically essential detail contrast, such as the trabecular patterns in bone examinations.

Benefits of an All-Digital Department

Changing healthcare needs require tomorrow's diagnostic imaging service provider to rapidly produce the highest quality images, transmit them broadly, display them in alternative ways, and then archive and retrieve them efficiently. New digital radiography image-capture systems are a critical element in this all-digital vision.

Within the examination room, Kodak DirectView DR systems eliminate the need for processing protocols associated with film and CR. Since the new digital radiography systems capture and convert an x-ray image into digital format within seconds of the exposure, the technologist can quickly preview each digitized image for quality assurance prior to completion of the exam procedure. DR provides the opportunity for more expedient patient-exam flow, fewer repeat exams, increased room use, and reduced cost.

Within and between imaging facilities, digital radiography systems should enhance workflow through networked distribution of the diagnostic images. A digitized image can be transmitted electronically for diagnosis via reading on a workstation monitor or for printing on film, as well as for electronic storage.

Because of the wide dynamic range inherent in most digital images, a radiologist can adjust the image electronically at a workstation to optimize the view of the desired anatomy. Hardcopy images from digital printers can also be optimized to suit the viewer's preferences.

For healthcare providers, the benefits of the new digital radiography systems stem from the fast and efficient production of diagnostic-quality images. Bottom-line benefits include improved patient care, increased productivity of staff and equipment, and the potential to attract a greater number of referral patients and physicians.

With all-digital radiography, patients will be able to use the most convenient location in a network for their exam and be assured that the procedure will be accurately and quickly completed. Patient images can be forwarded wherever and whenever they are needed for diagnosis, with the results communicated to a primary care physician in the shortest possible time.

With digital image-capture systems, the image data sent to workstations, printers, and archives is always identical to the original.

With improved workflow and increased efficiency, the all-digital radiology department will help hospitals, imaging centers, private practices, and clinics realize the full benefits of a picture archiving and communication system (PACS).

Summary

Ultimately, x-ray image quality depends on many factors, but it begins with the signal profile that is captured by the x-ray detector. Indirect-conversion digital radiography systems rely on light which scatters, degrading sharpness and signal-to-noise ratio. The Kodak DirectView DR systems convert x-rays directly to electrical charge. Their performance in preserving both the sharpness and the signal-to-noise ratio of the absorbed x-ray quanta is unequaled by any other detector in the industry.

Like Kodak CR systems, the wide dynamic range and automated image processing of the Kodak DirectView DR system provide an opportunity to minimize retakes while providing the highest quality images. Add to this the product's ease of use for technologists, enhanced productivity capabilities, plus the DICOM digital image format and the result is a new standard for x-ray image capture ideally suited to digital-projection radiography.

Glossary

Amorphous Silicon (a-Si:H) – Amorphous materials make flat-panel detectors possible. Early semiconductor technology required single-crystal silicon to produce useful electronic devices. This limited the size of electronic devices to the largest size that a single crystal could be grown. The development of amorphous-silicon materials, which have the same structure as single crystals over short distances but are less ordered over larger distances, has enabled fabrication of flat-panel thin-film transistor (TFT) arrays large enough to be used as the basis for all flat-panel x-ray detectors.

Amorphous Selenium (a-Se) – Amorphous selenium layers have the same structure as single crystals over short distances but, are less ordered over larger distances. As a result, amorphous selenium layers can provide uniform x-ray detection over the large areas needed by flat-panel x-ray detectors. Direct-conversion detectors use amorphous selenium, a unique x-ray-sensitive photoconductor that can be deposited onto amorphous-silicon TFT arrays.

Contrast Resolution – The smallest exposure change that can be detected. Ultimately, this is limited by the exposure range and the quantization (number of bits per pixel) of the detector.

Detective Quantum Efficiency (DQE) – A measure of noise performance that is obtained by comparing the image noise of a detector with that expected for an "ideal" detector having the same signal-response characteristics. The only source of noise in an ideal detector results from the incident x-ray quantum statistics.

Detector Size – The detector size describes the useful imaging area of an imaging device.

Detector Element – A detector element is the smallest resolvable area in a digital imaging device.

Exposure Range – The range of exposures over which a detector can capture an image. Digital radiography and computed radiography are capable of capturing an image over a much larger range of exposures than film-screen. This has been shown to reduce the number of retakes that result from over- or under-exposure.

Image Noise – All images have unwanted fluctuations that are unrelated to the object being imaged. These are collectively described as image noise. In addition to the x-ray quantum noise, which cannot be avoided, imaging systems contribute additional noise to an image. The electronic components of all digital detectors add noise. Indirect-conversion detectors have additional noise sources caused by the conversion of x-ray energy to light and the varying degree to which that light spreads before being absorbed by the light-sensitive photodiode.

Limiting Spatial Resolution (LSR) – The highest number of line pairs that can be seen in an image or target consisting of a series of periodic bar patterns of increasing spatial frequency. LSR is an unreliable measure of performance because it depends on the contrast of the target, the number and length of the target patterns, as well as the exposure and display conditions. Modulation transfer function is a much more reliable measure.

Matrix Size – The matrix size of a digital detector is the number of detector elements. This is normally expressed in terms of the number of detector elements in two orthogonal directions.

Modulation Transfer Function (MTF) – A measure of the ability of an imaging system to preserve signal contrast as a function of spatial frequency. Every image can be described in terms of the amount of energy in each of its spatial frequency components. MTF describes the fraction of each component that will be preserved in the captured image.

Nyquist Frequency – The highest spatial frequency that can be represented in a digital image. The Nyquist frequency is determined by the pixel spacing.

Photodiode – An electronic element which converts light into charge. Indirect-conversion detectors require fabrication of a light-sensitive amorphous silicon photodiode on top of the thin film transistor array. This fabrication step adds significant complexity to the indirect-conversion detector design.

Photoconductor – In direct-conversion detectors, the amorphous selenium layer forms a continuous x-ray-sensitive photoconductor which converts x-ray energy directly to charge. This charge can be directly "read out" by the TFT array, a simple and robust design that does not require complex photodiode fabrication steps needed by indirect-conversion detectors.

Pixel – A "picture element," the smallest area of an image which is represented in a digital image. A digital radiography image consists of a matrix of pixels which is typically several thousand pixels in each direction.

Quantization – While all x-ray detectors respond smoothly and continuously to the incident exposure, digital images require the detector response to be quantized into a fixed number of levels that can be represented digitally. This number is typically 12 to 14 bits or 4096 to 16384 unique levels for flat-panel x-ray detectors.

Scintillator – A material that absorbs x-ray energy and re-emits part of that energy as visible light. Two modern high-efficiency x-ray scintillators are cesium iodide and gadolinium oxysulfide. Cesium iodide is commonly used in x-ray image intensifiers and is highly hygroscopic, which means it readily absorbs water from the air. Cesium iodide must be hermetically sealed to avoid water absorption or it will degrade rapidly. Gadolinium oxysulfide is commonly used in x-ray intensifying screens to expose film. It is a highly stable material, but has significantly more light spread than a layer of cesium iodide with equal x-ray absorption.

Signal-to-Noise Ratio (SNR) – Because noise ultimately limits our ability to see an object (the signal), SNR can be used to describe the detectability of a particular object under well-defined exposure conditions. The SNR in an image is always less than or equal to the SNR of the incident exposure. Detective quantum efficiency (DQE) is a measure of the efficiency with which the SNR of the incident exposure is preserved in an image.

Thin Film Transistor (TFT) – An electronic switch commonly made of amorphous silicon on flat-panel detectors. The TFT allows the charge collected at each pixel to be independently transferred to external electronics, where it is ampified and quantized.

Tiling – A process whereby several flat-panel detectors are joined to obtain one larger detector.

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