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FLAT-PANEL DIGITAL RADIOGRAPHIC DETECTOR TECHNOLOGY

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INTRODUCTION

There are several types of digital detector technologies presently in clinical use for medical radiography. These include computed radiography (CR), amorphous selenium direct radiography (DR) drum systems, charge-coupled device (CCD) based systems of several kinds, selenium-based flat panels, amorphous silicon flat panels using cesium iodide scintillators, and gadolinium-based flat panels. This article provides an overview of the technology used in the more recently developed direct-readout flat-panel radiographic imaging systems.

IMAGE CONVERSION SYSTEM BASICS

All x-ray image detection systems, from screen-film through the latest digital detectors, convert x-ray radiation into an image perceptible to the human eye. Screen-film systems, still in use in many clinics, use exposure of photographic film in contact with x-ray fluorescent screens to convert the x-ray image to an image of varying opacity on the exposed and developed film. The large majority of the conversion in this case is done by the fluorescent screen converting x-ray photons to photons of visible light that then expose the film, although the film is also directly sensitive to the x-rays themselves.

In screen-film systems, the film that records the image is also used, after development, directly to display the image by use of a light box, and then is also used as the archival storage medium for the image as well. Now that computer technology has become widely available and networked, however, the drawbacks of film-based systems can be overcome. Once an image is digitized, it is archived like any other computer file, typically with a picture archiving and communication system (PACS), so it is much more difficult to lose an image, which is possible when filing or archiving a physical piece of film. It is faster and more convenient to send a digital image for viewing at different locations than it is to carry film, and a digital image can more easily take advantage of image-processing techniques, such as edge-enhancing filtration.

In digital electronic image detection systems, the x-ray image is converted into an electrical signal that is then digitized, recorded, and displayed. The first type of system widely used to perform this task for radiographic applications was computed radiography (CR), which is still in use. CR systems use plates coated with photostimulable phosphors, also called storage phosphors, in the place of screen-film cassettes. A typical phosphor used in CR systems is $BaFBr{:}Eu^{2+}$. Conversion is accomplished by the phosphor, which traps energy from the incident x-rays and stores it in the form of electrons trapped in higher energy levels than they normally occupy. CR plates can store images for several hours, although they should be read out as soon as possible because the images slowly degrade as the trapped electrons return to their ground states. The image is read out by scanning the plate with a laser at one wavelength, which causes the phosphor to emit light (giving up its energy) at another wavelength, which is then detected and recorded electronically. The plate is then exposed to a bright light to ensure that the image is completely erased, and it can then be reused. CR systems were widely adopted because they provided a means to digitally acquire radiographic images of sufficient quality for clinical use, produced images approximately similar to film images, and had the advantage that the CR plates could be made to the same sizes as screen-film cassettes, which enabled them to be used in most existing film-based radiography equipment.

Note that the CR plate itself is an analog storage medium, used as a temporary intermediate stage between image acquisition and later electronic image digitization. Other technologies used to produce digital radiographic images are referred to as direct radiography (DR). These systems are then further divided by their means of image conversion into direct conversion and indirect conversion types. Direct conversion refers to systems in which the x-ray photons are directly converted into an electronic signal. Indirect conversion refers to systems in which the x-ray photons are first converted into visible light photons, which are in turn converted into electronic signals.

Once the image information is converted into an electronic signal, regardless of how it is initially converted from x-rays to an electronic signal, it is digitized with analog-to-digital conversion (ADC) circuitry. This converts the electronic signal into a string of ones and zeros that can then be processed and recorded like any other computer data. The details of the ADC circuits vary depending on the characteristics of the signals and related acquisition circuitry of the specific system, but one quantity that is generally specified for all of them is the precision, or bit depth, for each pixel. Pixel values for radiographic images are represented by a single number, which in binary may typically have 8 or 10 bits. These correspond to gray scales with 256 and 1024 gray levels, respectively. The number of gray levels is equal to two to the power of the number of bits in the binary representation of the pixel value. Just for comparison purposes, a pixel of an ordinary computer monitor typically has three eight-bit numbers associated with it, representing 256 possible levels for each of the three color components of red, green, and blue. The radiographic image pixel is represented by a single number because radiographic detectors are unable to distinguish different ranges ("colors") of the x-ray spectrum. (Some x-ray systems nowadays allow a mode of operation called "dual energy," in which two images are taken in succession with different kVp settings or different types of filtration in the beam, which results in image representations of differently weighted x-ray spectra. However, these systems are outside the scope of this article.) For another point of comparison, CT detectors typically digitize their outputs to 20 bits per data value. This level of precision is needed for the reconstruction algorithms used.

Some DR technologies are too bulky to be considered with flat-panel detector technologies, but we will mention some of them briefly here, because they serve to introduce technologies later used in some flat-panel radiographic detectors, and because some are still widely used in niche applications, such as chest imaging, where bulkiness of the detector is less of a concern.

One indirect technology for DR is based on charge-couple devices (CCDs). Each pixel of the CCD detector has a semiconductor capacitor that is light-sensitive. Arrays of these pixels are fabricated on silicon chips similar to computer chips. (CCDs are a common type of image detector also used for applications ranging from some video cameras to astronomical telescopes.) For a radiographic x-ray system, conversion is accomplished by a scintillator, such as cesium iodide doped with thallium (CsI:Tl). One drawback of CCD technology for radiography is that, with present technology, CCDs can be made in only limited sizes, up to a few centimeters. So to make an image detector big enough in area for radiographic use, including chest imaging, either many separate CCD chips must be put together side-by-side, called tiling, or the image from the scintillator must be demagnified (minified) optically, between the scintillator and the CCD. This demagnification can be done with either lenses or fiber optics, but in either case, some light is lost, which hurts the overall efficiency of the system. In the case of tiling, the image information right at the seams where two tiles abut may be lost, causing missing rows or columns in the overall image, which must be corrected for.[1] One way systems do this is to use both optical demagnification and multiple CCDs, but with the lens components designed so that the sections of the image produced from each CCD chip have some overlap instead of gap at the edges. Additionally, optical systems used in radiographic CCD systems sometimes introduce geometric distortion to the image, similar to the distortion that sometimes occurs with image intensifier systems. Finally, the optical systems make the overall CCD detectors that use them too thick and bulky to be considered a flat-panel detector. They still have applications but cannot be used for many of the images for which thinner detectors are used.

One flat-panel technology that uses indirect conversion consists of three layers: a scintillator, which converts the x-rays to visible photons, a layer of photodiodes that convert the visible photons into electrical charges, and a layer of thin-film transistors (TFTs) to read out the charges on the photodiodes. The photodiodes and TFTs are implemented in amorphous silicon (A-Si) technology. Amorphous silicon is a form of silicon that does not contain crystals. Unlike the crystalline silicon used in CCDs, amorphous silicon is deposited on glass substrates, which can be made large enough so that an entire chest image can be captured on a one-piece detector, without the need for either tiling of multiple detector pieces, or image demagnification by optical means.

A typical scintillator used in A-Si detectors is CsI:Tl. In this application, the CsI:Tl crystals are grown in the shape of very thin needles, typically less than 10 µm wide. This is much smaller than the size of the pixel, which depending on the system, may range from 100 to 200 µm. The CsI:Tl crystals are typically several hundred microns long, and are packed like toothpicks, sticking up from the photodiode layer. This is done so that the CsI:Tl crystals act like tiny fiber optic components; when x-ray photons are converted to light photons in the crystal, the light photons are then conducted by the needle-shape of the crystal so they exit from either end, with very few crossing into an adjacent needle. (The optical name for this is total internal reflection, although it is only total for photons traveling at certain ranges of angles to the surfaces of the crystals to begin with.) This keeps the light photons restricted to the pixel they were converted over, and allows thicker layers of scintillator than would otherwise be possible. In practice, an optically reflective layer is put at the ends of the needles away from the photodiode, to reflect light that might have escaped in that direction back towards the detector. This arrangement is similar to the reflective layers used in many conventional screen-film systems. An alternative scintillator used in some A-Si detectors is gadolinium oxysulfide, Gd_2O_2S.

Another flat-panel technology is selenium-based direct conversion. Amorphous-selenium (A-Se) direct conversion detectors also use arrays of thin-film transistors (TFTs) to read out the image. A coating of amorphous selenium is applied over the array of TFTs. An electrical charge is then applied to the selenium prior to exposure. During x-ray exposure, x-ray photons mobilize electrons in the selenium and cause currents to flow proportional to the exposure. These electrical signals flow into the TFTs, and are then digitized. These detectors are physically robust, because the amorphous selenium layer is stronger than the needles of CsI:Tl used in the indirect conversion detector previously described. However, because the fiberoptic effect of the CsI:Tl crystals is absent in selenium-based direct conversion detectors, the range of thickness of the selenium layer that can be used is much more limited. The tradeoff is that if the selenium layer is made thicker to increase conversion efficiency (capture more x-ray photons in the thickness of the Se layer), there will also be more leakage of electrons created over one pixel into the TFTs of adjacent pixels, which affects image quality.

IMPLEMENTATIONS AND CHARACTERISTICS

Basic characteristics of several flat-panel radiographic systems are shown in Table 1, which compares them to other radiography systems.[2] Notice that one feature that all the electronic imaging systems share is that they offer greater dynamic range than do screen-film systems. This means that the linear range of image contrast versus exposure is much broader for electronic radiography systems than for screen-film systems, particularly so for the flat-panel technologies as well as for CR systems. Recall the S-shaped sensitometric curve for radiographic film, which had a narrow linear range in the middle, and wide areas of under-exposure on one side and over-exposure on the other side (Figure 1).[2] The equivalent curve for flat-panel detectors, whether using A-Se, CsI:Tl, or Gd_2O_2S for conversion, has a much broader linear range in the middle, and much smaller ranges of under- or over-exposure. What this means is that there is both greater exposure latitude for these systems, and also a greater amount of information available in each image, in the sense of more gray shades, although it may not be possible to display all the gray shades at the same time on present electronic displays. How to make the most of displaying the image will be discussed below.

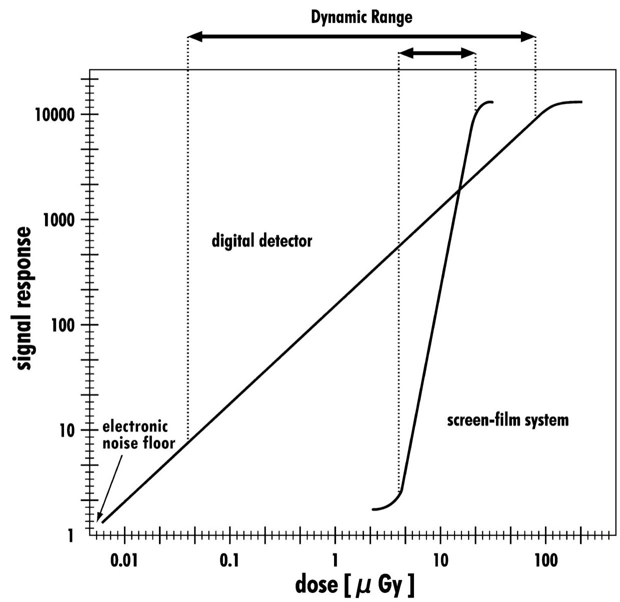

FIGURE 1. This graph illustrates the dynamic range of screen-film combinations and digital detectors. Screen-film systems have only a limited tolerance for radiation exposure, resulting in a steep and tight curve, whereas the curve for digital detectors is less steep and covers a wider range. As a result, an optimal signal response occurs over a wider exposure range with digital detectors than with screen-film combinations.

TABLE 1. Technical Features of Various Digital Radiography Systems[2]

	Screen-Film System	Storage-Phosphor System	Lens-coupled CCD	Slot-Scan CCD	Direct FPD	Indirect FPD	Indirect FPD
Converter	Gd_2O_2S	BaSrFBr:Eu	Gd_2O_2S	CsI:TI	Selenium	Gd_2O_2S	CsI:TI
Readout	Film	Laser	CCD	CCD	Active selenium matrix	Active silicon matrix	Active silicon matrix
Detector size (in)	14 x 17	14 x 17	14 x 17	17 x 17	14 x 17	17 x 17	17 x 17
Pixel size (µm)	N/A	200	167	162	139	160	143
Matrix size	N/A	1760 x 2140	2000 x 2500	2736 x 2736	2560 x 3072	2688 x 2688	3121 x 3121
Nyquist frequency (cycles/mm)	5	2.5	3.0	3.1	3.6	3.1	3.5
Dynamic range	1:30	1:40,000	> 1:40,000	1:10,000	> 1:10,000	> 1:10,000	> 1:10,000

The reduced exposure sensitivity of flat-panel radiography detectors (along with CR receptors) due to their greater dynamic range carries two significant benefits to the patient in terms of dose. One is that the need for retakes due to slight exposure errors is greatly reduced or nearly eliminated by most modern flat-panel radiography systems. While screen-film systems can be so sensitive to exposure that individual patient variation can occasionally cause a poorly exposed image despite the correct technique being set based on available information prior to the exposure, this is a much more rare occurrence with flat-panel radiography systems. Therefore, additional patient dose due to retakes is nearly eliminated.

Secondly, because some flat-panel radiographic systems have detective quantum efficiency (discussed below) superior to screen-film systems, it is sometimes possible to achieve a diagnostically sufficient image with less dose to the patient using a flat-panel system, than a screen-film system would have required.[3] What would be a slightly under-exposed image with a screen-film system is often sufficiently exposed with a flat-panel detector, because the linear range of the flat-panel detector extends farther into the lower-exposure range than does the linear range of the screen-film system. This advantage can result in reduced dose to the patient. This effect is limited, however. As exposure is reduced, noise in the image increases. If exposure is reduced excessively, the image will be so noisy that it will be diagnostically inadequate, and a retake will be necessary.

There is also a similar effect of this extended linear range that the technologist must watch out for. Because the linear range of flat-panel detectors also extends beyond that of screen-film systems in the direction of over-exposure, it is possible to get an acceptable (or better) image even if the exposure is set at what would be slightly too high for a screen-film system. Moreover, because of the inverse relationship between noise and exposure, as the image is over-exposed with a flat-panel detector, noise is reduced, and the image looks better, as long as it is still within the linear range. The technologist must take care not to set the exposure too high, just to make better-looking images that are possible with flat-panel detectors, at the expense of unnecessary extra dose to the patient. The exposure should be set to produce a diagnostically sufficient image for the particular patient, anatomy, and position. The RT should be sure to use exposures within the ranges specified in the device operator's manual, and the clinic's standards of practice, for each case.

Another characteristic to notice from Table 1 is that pixel size for electronic radiographic detectors of all types is slightly larger than the equivalent measure for screen-film systems. Film does not have exact pixels like digital electronic detectors, of course, but it has a characteristic grain size, which effectively produces a similar effect of limiting maximum resolution for a given screen-film system, just as pixel size limits maximum resolution for electronic imaging systems. As pixel size or film grain size goes down, maximum resolution possible in the image goes up. In both the case of film and the case of digital detectors, the maximum resolution can be expressed in terms of the Nyquist frequency, in units of cycles/mm. In the case of digital detectors, the Nyquist frequency is simply the number of pixels per millimeter divided by two. For example, for a detector with a pixel size of 200 µm, there are 5 pixels/mm, so the Nyquist frequency for that detector is 2.5 cycles/mm. The effective Nyquist frequency of any radiographic system can be easily tested by taking an image of a lead resolution wedge test pattern.

However, maximum resolution or, equivalently, Nyquist frequency, does not tell the whole story on image quality. Images of anatomy contain information that is both clinically relevant and evident to human perception (the eye-brain system) at many length scales, referred to as spatial frequencies. A longer length scale of a feature in an image corresponds to a lower spatial frequency, while shorter length scales of features in an image correspond to higher spatial frequencies. Imaging system performance over the range of spatial frequencies is measured by modulation transfer function (MTF). You can think of MTF as a way of expressing how well a system or component represents contrast, from its input to its output, for each spatial frequency. For a digital radiography detector, the input is the x-ray beam after it has gone through the patient, and the output is the digitized image data that is transmitted to the PACS system. The MTF represents how well contrast information, which is originally present in the x-ray beam as it exits the patient, in the form of greater and lesser intensities of x-rays depending on how much attenuation of the beam the patient's anatomy caused, is conveyed by the image output of the detector, across all length scales or spatial frequencies present in the image.

Finally, another measure of image quality is detective quantum efficiency, or DQE. Like MTF, DQE expresses image quality over the whole range of spatial frequencies or length scales present in the image. However, in addition to contrast (signal), DQE also factors in exposure level, noise, and beam quality, which is a combination of voltage and current at the tube, and filtering applied to the beam to adjust its energy spectrum. Although DQE has largely become a popular measure of radiographic system image quality since the introduction of digital image detectors, it is also applicable to measure the image quality of conventional screen-film systems. Before the year 2003, different researchers and manufacturers measured DQE by different methods, so it was hard to effectively compare reports and claims of performance for different systems. In 2003, the method of measuring DQE was standardized by IEC62220-1 of the International Electrotechnical Commission. Measurements reported since then, using that method, are comparable. Figure 2 shows DQE measurements made on four different digital radiography systems: two CR systems, a CsI indirect conversion system, and an A-Se direct conversion system.[2] Screen-film systems have DQEs approximately equivalent to the lower of the CR system curves (lowest of four curves shown).

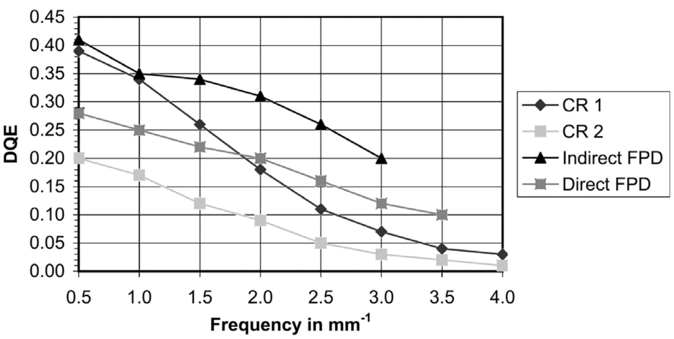

FIGURE 2. This graph illustrates the DQE curves for four digital detectors. CR 1 = needle-structured storage phosphor and line scanner (MD5.0/DX-S; Agfa-Gevaert, Mortsel, Belgium), CR 2 = unstructured storage phosphor and flying-spot scanner (MD40/ADC Compact, Agfa-Gevaert), Indirect FPD = CsI-based flat-panel detector (Pixium 4600; Trixell, Moirans, France), Direct FPD = selenium-based flat-panel detector (DR 9000; Kodak, Rochester, NY).[2]

Present flat-panel radiographic systems are capable of displaying an image within approximately 10 seconds of exposure, which enables the technologist to quickly determine whether the image demonstrates the anatomy properly or if a retake is required with adjustment to the patient positioning, or for any other reason, such as patient motion. This speed is significantly faster than that of CR systems or screen-film systems and contributes to increased patient throughput for a system.

A possible countervailing factor on image acquisition speed should be noted. Some flat-panel radiographic detectors are known to have memory effects. In one test with high-contrast lead phantoms, a CsI indirect conversion flat-panel detector was found to have residual contrast of 1.7% from a previous exposure, when a blank exposure was taken five minutes later.[4] However, further testing of this system, both with anthropomorphic phantoms and with human subjects, did not reveal any cases in which features of a prior image were visible in later images.

MAKING CORRECTIONS TO RAW IMAGE DATA

The raw image data coming from a flat-panel detector is post-processed before being archived and displayed. Some of this post-processing is done for correction, while other post-processing is done for enhancement. The corrections are typically applied automatically by the image acquisition system, while enhancement processing, such as digital filtering, is typically applied either by the technologist making the image, or a radiologist viewing the image later.

There are several types of corrections made to flat-panel images, which generally fall under the headings of bad pixels, and gain and offset corrections. Bad pixels, or dead pixels, are pixels that are either unresponsive entirely, or fall out of bounds for linearity, current leakage, or other parameters measured when the detector is tested and calibrated. Bad pixels are present from the time the detector panel is fabricated, and are presently unavoidable due to the panel fabrication processes. The panel fabrication process consists of starting with a sheet of substrate, typically glass, and adding layers to it, by means of chemical vapor deposition, each layer adding components and elements either to the electrical circuitry, the scintillator, or other layers for the purpose of insulation, sealing, and so on. There are also steps involving selective removal of certain areas of layers laid down previously, by means of chemical etching. In a typical panel fabrication process, there are well over one hundred steps. At each step, any microscopic speck of dirt that somehow gets on a pixel can cause that pixel to come out bad. Great care is taken to fabricate the panels in clean-room environments, but no system of cleaning is perfect. These systems are similar to, and in most cases derived from, the methods used to fabricate computer chips and other electronic components. However, in the case of computer chips, typically a great many small chips are made on each substrate, which are then separated by sawing them apart. The individual chips are then inspected and tested, and those that turn out to have defects are simply discarded. In the case of radiographic flat-panel imagers, the entire panel is used as one imager without being sawed apart, and discarding entire panels with correctable defects is not practical. Instead, a map of each panel is made when it is tested and calibrated. The map records the position of each bad pixel. Some bad pixels are isolated, some are in clusters, and some are grouped in rows or columns. In each type of case, there are bad pixel correction algorithms, which are used to assign data values to each bad pixel after each image is acquired, based on the data values of the good pixels in the immediate neighborhood of each of the bad pixels. These are similar to the algorithms used in the case of tiled CCD detectors, in which values of pixels located on the joints where tiles meet must be determined in a similar manner. Obviously, there is a limit to how many bad pixels can be corrected in this way. Each manufacturer sets quality limits on how many bad pixels can be tolerated by these methods of correction, and panels with excess bad pixels must be discarded entirely. In a test of one flat-panel CsI indirect conversion image system, approximately 0.1% of the pixels were bad pixels, and this imager was found to have "excellent uniformity, repeatability, and linearity, as well as MTF and DQE that are superior to those obtained with a storage phosphor CR system."[4]

Additionally, not all good pixels are equal. Due to microscopic variations from pixel to pixel in the chemical vapor deposition and etching processes described above, each pixel is slightly different. Consider the linear region of the sensitometric curve for the flat-panel detector. In actuality, each pixel has its own individual sensitometric curve. The linear region for this curve for each pixel is approximated by the equation of a line, which has two parameters. From algebra, the equation of a line is $y = mx + b$, where x is the input, y is the output, m is the slope of the line, and b is the y-axis intercept of the line. In the case of the linear region of a sensitometric curve for a pixel, the slope m is called the gain, which is the amount the pixel value changes for a given change in exposure, and the y-intercept b is called the offset, which represents what the pixel value is when the exposure is zero.[5] (For a variety of reasons having to do with the design of the readout electronics, and fabrication of the photodiodes in the case of indirect conversion systems, the pixel value is not always zero for zero exposure.) So each pixel has a slightly different gain and offset, each of which is measured when the panel is tested and calibrated. To correct the pixel values during clinical image acquisition, so that all the pixels appear to be equivalent to each other in performance, maps are constructed, pixel by pixel, for each panel's gain and offset correction factors. These gain and offset corrections are then applied automatically every time an image is taken, so that for each image that is output for archiving and display, it appears that each pixel is behaving sensitometrically identically to every other pixel.

The technologist or radiologist can also apply post-processing to digital radiographic images. Among the most commonly used processing is filtering, which can be used to either enhance fine details such as edges, or suppress excess sharpness of edge contrasts to reveal features that are only perceived at lower spatial frequencies.

VIEWING DIGITAL IMAGES

There are two main ways of viewing digital radiographic images. One method is by printing to film using a printer called a laser camera. This was a common method in prior years, but as computer display technology has improved, it is becoming more rarely used, because by using film, it negates many of the advantages of digital imaging. Nowadays, displaying digital radiographic images on medical-quality computer displays is the predominant method of viewing digital radiographic images. The main factor that should be emphasized when discussing the display of radiographic images on computer displays is that even with the best presently available displays, the number of gray shades that the display can show at one time, no matter how well its brightness and contrast are adjusted, is far fewer than the number of gray shades present in the digital image data. How this is accommodated with present image viewing systems is by adjusting what are called window and level. Window refers to how many gray shades from the image data are being viewed at once. A narrow window setting means that only a few gray shades are being displayed. Usually it is not worth setting the window much smaller than the number of gray shades the display is capable of showing, although there are occasions where this may be desirable. Care must be taken when the window setting is significantly larger than the number of gray shades that the display can show, because then some of the shades in the data are being lumped together, or binned, with adjacent shades, because only so many can be shown by the display. Level refers to where the window is, in relation to the entire range of gray shades.

Due to the large dynamic range of flat-panel digital images, clinically significant information can be present in an image, but it can be at such different levels that it cannot all be displayed effectively at the same time. This is where the level control comes into play. An example is a chest image, which commonly features a wide dynamic range, from highly exposed lung areas, to the areas of dense bone in the spine, which are least exposed. When examining a digital radiographic chest image, if one were concerned with features of the spinal bones, such as when checking for fractures, one would set the level control of the display to show the range, or window, of pixel values corresponding to the less-exposed dense bone regions. However, if one were examining the same image to diagnose a lung condition such as tuberculosis, which is characterized by darker regions among the already dark, highly exposed lung areas, one would set the level control to where the range (window) of pixel values corresponded to the highly-exposed regions. With present display technology, if the window setting were set so wide that both of these different levels were shown within the same window of pixel values, neither the fine bone features nor the fine lung features would likely show up as well as with a narrower window setting at each of the different level settings. Too many gray shades would be binned together into the fewer number of shades that the display can show at once, so the contrast precision available in the image data would be wasted. This is why it is important to select appropriate window and level settings based on the specific bit depth of the image being displayed, and the capabilities of the display on which it is being viewed.

CLINICAL COMPARISONS

Several studies have been done comparing flat-panel digital radiography detectors with other radiographic technologies. Uffmann et al, in a study with phantom test objects for skeletal imaging, compared digital flat-panel detectors to CR storage phosphor systems, with five people reading test images to determine the ability to detect low-contrast details.[6] They found that the digital flat-panel detectors could be used with between 17% and 45% less dose versus the CR systems, while achieving the same contrast detection.

In a study that compared digital flat-panel detectors to screen-film, photofluorography, and storage-phosphor (CR) systems with phantom test objects simulating lung adenocarcinoma lesions, using hard-copy images (that is, printed to film in the case of the digital systems), Ono et al found no significant difference between images coming from digital flat-panel detectors and CR systems versus screen-film and photofluorography systems.[7]

However, a study by Ganten et al also compared digital flat-panel detectors to CR storage-phosphor systems and film-screen systems for chest radiography using 30 human patients, and concluded that digital flat-panel detectors could produce clinically equivalent images to CR and film-screen radiography systems with up to 50% dose reduction to the patient.[8]

In addition, in a study of 100 oncology patients, Fink et al compared hard-copy images produced by a digital flat-panel detector with dose set equivalent to that of a 400-speed film system, to conventional screen-film images taken at 200 speed.[9] Thus, the digital flat-panel system was operated at approximately 50% less dose than the screen-film systems. Their conclusion was that the images from the digital flat-panel detector were equivalent or superior to the screen-film images.

CONCLUSION

Digital flat-panel radiographic image detectors are now widely in clinical use alongside other radiographic technologies. They offer wider exposure latitude compared to conventional screen-film systems, and have demonstrated equivalent or superior image quality while offering the potential to reduce patient dose.

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FLAT-PANEL DIGITAL RADIOGRAPHIC DETECTOR TECHNOLOGY POST TEST

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1. **Which of the following is a TRUE statement regarding screen-film systems?**
   a. Film-screens systems are now obsolete and have been completely replaced by digital detector systems.
   b. The majority of the image is created when x-rays interact directly with the photographic film.
   c. The majority of the image is created by the fluorescent screen.
   d. The film used in these systems is not sensitive to direct x-rays.
2. **CR systems use plates that are coated with**
   a. storage phosphors.
   b. tiny digital detectors.
   c. an oxide semiconductor material.
   d. copper substrate.
3. **What is used to erase a CR plate so that it can be reused?**
   a. A strong electrical charge
   b. A bright light
   c. A laser
   d. Intense heat
4. **Digital technologies that produce radiographic images by first converting the x-ray photons into visible light photons and then converting the light photons into electronic signals are called**
   a. analog storage methods.
   b. computed radiography systems.
   c. direct conversion methods.
   d. indirect conversion methods.
5. **What is it called when many charge-couple devices (CCDs) are put side-by-side to create an area large enough for radiographic use?**
   a. Tiling
   b. Compiling
   c. Integrating
   d. Adjacent arrays
6. **Which of the following correctly describes amorphous silicon?**
   a. Silicon that has been aerosolized
   b. Silicon that does not contain crystals
   c. Mixed inorganic-organic polymers that are use to make various synthetic plastic substances
   d. A radioactive silicon isotope that is produced by argon decay
7. **Compared to screen-film systems, one feature that all electronic images systems share is that they**
   a. are less expensive.
   b. offer greater dynamic range.
   c. require less maintenance.
   d. offer greater exposure sensitivity.
8. **In flat-panel radiographic systems what is the result when exposure (that is, dose) is reduced excessively?**
   a. The image is too dark.
   b. The image is too light.
   c. The image is fine.
   d. The image is noisy.
9. **In flat-panel radiographic systems, what is the result when exposure (that is, dose) is higher than necessary?**
   a. The image is too dark.
   b. The image is too light.
   c. The image is fine.
   d. The image is noisy.
10. **In regards to digital detectors, what is the Nyquist frequency?**
    a. The number of pixels per millimeter divided by two
    b. 22,050 Hz
    c. 78.7 line pairs per millimeter
    d. The number of active pixels per frame
11. **Which of the following correctly describes detective quantum efficiency?**
    a. A measure of image quality
    b. The principle that explains why an x-ray beam is attenuated by matter
    c. The study of the relationship between energy and matter
    d. The number of gray shades from the image data that are being viewed at once
12. **Present flat-panel radiographic systems are capable of displaying an image within approximately _____ seconds of exposure.**
    a. 0.5
    b. 10
    c. 60
    d. 90
13. **Pixels that are either unresponsive or fall outside quality control specifications are called _______ pixels.**
    a. bad
    b. partial
    c. disarranged
    d. inert
14. **Why must flat panels be manufactured in clean-room environments?**
    a. Bacteria that could be harmful to humans are used in the manufacturing process.
    b. Ordinary airborne dust can cause the production machinery to jam up.
    c. Even microscopic particles of dirt have the potential to ruin the panels.
    d. Flat-panel components are radioactive and for the safety of the manufacturing staff they must be handled only in special rooms.

15. In the case of the linear region of a sensitometric curve for a pixel, the slope *m* is called the
  a. rise.
  b. run.
  c. offset.
  d. gain.

16. The amount the pixel value changes for a given change in exposure is called the
  a. aperture-to-medium coupling loss.
  b. transmitter power output.
  c. gain.
  d. fractional pixel offset.

17. In the case of the linear region of a sensitometric curve for a pixel, the y-intercept *b* is called the
  a. rise.
  b. run.
  c. offset.
  d. gain.

18. The pixel value when the exposure is zero is called the
  a. filter.
  b. baseline.
  c. initiation point.
  d. offset.

19. A post-processing digital radiographic technique that can be used to either enhance or suppress the edges of anatomic structures is called
  a. histogram equalization.
  b. filtering.
  c. linear point operations.
  d. image resizing.

20. In regards to the display of radiographic images on computer workstations, window refers to
  a. how many shades of gray are being displayed.
  b. how many images are displayed on the monitor at once.
  c. the type of monitor used.
  d. the number of pixels being displayed.

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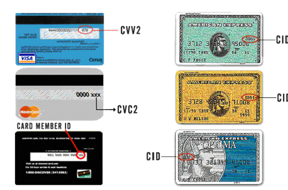

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