Introduction to Digital Imaging - Part I: Digital Detectors

Expires Tuesday, February 28, 2023 Radiology William R. Geisler , M.S. DABR

Objectives

1. Describe the process as to how an image is obtained and processed for the purpose of interpretation utilizing screen-film x-ray imaging.
2. Describe the process as to how an image is obtained and processed for the purpose of interpretation utilizing digital detectors in computed radiography.
3. Discuss the various aspects of photostimulable phosphor detector systems, charge-coupling devices, a-SI detectors, and a-Se detectors.

Article

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Part I: Digital Detectors

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Introduction

Digital radiologic imaging had its beginning with the first computed tomography unit in the early 1970's followed by magnetic resonance imaging about 10 later. Nuclear medicine and ultrasound imaging are also intrinsically digital modalities. However, until the last decade these modalities were printed on film for interpretation by a radiologist. With the recent conversion of general radiographic imaging from film to digital, coupled with the growing use of picture archiving and communication systems (PACS), digitally acquired x-ray images are now being interpreted on computer monitors, referred to as soft-copy interpretation versus hard-copy or film interpretation.

This article will provide an overview of digital versus screen-film x-ray imaging and discuss the various technologies available to acquire a digital image. A future article will discuss further aspects of the digital image (pixel, matrix size and bit depth, storage requirements), image manipulation and display, and PACS systems.

Overview of Screen-Film Systems

The well-known technology of screen-film x-ray imaging will be briefly reviewed. Screen-film imaging is performed by having x-rays that penetrate through the patient (attenuation of x-rays through the body) that then interact with a phosphor material within the cassette, which converts x-rays to visible light (**figure 1**). This conversion of x-ray photon to light photon occurs due to the film being much more sensitive to light radiation than to x-ray radiation, permitting a significant reduction in technique to obtain an acceptable image.

For x-ray film, the relative *speed* of the screen-film combination determines the required radiation exposure needed for an acceptable image. The speed of the screen-film is primarily determined by the thickness of the phosphor material: a thicker

phosphor absorbs a greater number or percentage of x-rays permitting less technique. However, while a fast screen-film (say 1200 speed) will permit a lower technique (and lower radiation exposure to the patient), the visibility of fine detail (resolution) will be lost due to greater light spread in a thicker phosphor material **(figure 2)**. Physical properties of the film, such as the size of the emulsion grains, can also affect the speed but to a lesser extent.

Following exposure of the film to visible light, an image is not yet "visible". An invisible image, called a *latent* image exists, which represents the absorbed light energy that interacted with the film. Obtaining a visible picture requires chemical processing of the image in a film processor (development stage). Following development an image is now visible on the film (and to maintain handling and storage the film undergoes fixing then drying, which then completes the processing cycle).

Another important property of x-ray film is how the film optical density (O.D.) varies depending on the amount of exposure a certain area of the film received. The O.D. determines the shades of gray in the image **(figure 3)**. A plot of the O.D. versus (logarithm) radiation exposure to film is known as the Hurter and Driffield (H&D) curve; this graph is a plot of film O.D. (amount of light transmitted) vs. radiation exposure (common or base 10 logarithm of radiation exposure). The curve indicates the level of grayness a certain region of the film will have. The greater the difference in gray levels (or difference in film O.D.) the greater the contrast (ease of visually differentiating two different regions on the film).

The H&D curve also sets the allowed range of radiation exposures the screen-film requires for an acceptable image. Too low of an exposure will result in an image that is light in gray values and will also exhibit excessive noise (or quantum mottle). This region of the H&D curve is referred to as the toe of the curve. Too high an exposure results in images being too dark and is near the shoulder of the H&D curve. For films exposed near either the toe or shoulder region on the H&D curve, the resulting image contrast will also be poor compared to images with the exposure values in the linear region of the curve **(figure 4)**.

Acquiring a Digital Image

There are a number of ways to obtain a digital x-ray image. These methods are typically broken into what is referred to as indirect capture and direct capture imaging methods **(figure 5)**. Indirect capture involves converting ionizing radiation that penetrates through the body into another form of radiation prior to being converted to a digital signal – either by converting the x-rays to visible light (much like the screen phosphor for x-ray film or the cesium iodide image intensifier for fluoroscopic imaging convert x-ray photons to visible light photons) or having the x-rays "captured" and temporarily stored (as is done with computed radiography or CR based imaging).

With direct capture imaging, x-rays that penetrate the body are converted into a digital signal for display on a computer monitor without temporary storage or conversion of x-ray energy into light energy. Details regarding indirect and direct radiographic digital imaging will now be discussed.

Computed Radiology (CR)

Computed Radiography (CR) was first developed in the 1970s and saw a significant increase in popularity in the late 1980s. CR is actually a marketing term for photostimulable phosphor (PSP) detector systems. These photostimulable phosphor detectors are commonly referred to as either a CR plate or imaging plate (IP). As discussed previously, phosphors are used in screen-film x-ray imaging to convert x-ray photon energies into visible light photon energies due to x-ray film being significantly more sensitive to visible light than to direct x-ray exposure. However, for screen-film imaging, the conversion of x-ray photons into light photons occurs almost immediately, whereas with PSP detectors the x-rays are absorbed by the PSP material and the energy is trapped within the detector.

As an analogy, the image stored on the PSP detector or imaging plate corresponds to the latent image stored on x-ray film. Just as a latent image is not visible until the x-ray film is developed, the stored image on the PSP detector is not visible until a device referred to as a CR reader acquires the stored image. In contrast to the need for chemicals to develop a latent image stored on x-ray film, the CR reader uses a laser-emitting red light and scans the PSP detector. The PSP detector or imaging plate is transported through the CR reader and a laser strikes a rotating mirror (optical scanner) that permits the laser light to scan across the plate during transport. The red light from the laser causes the PSP material to emit blue light (the more x-rays captured in a given region of the imaging plate the greater amount of blue light is emitted).

A photomultiplier tube (PMT) detects the emitted blue light and digitizes this light into a digital signal. The more x-rays absorbed in a particular region of the PSP detector will result in a brighter emission of blue light. The PMT converts the different intensities of blue light into a voltage, which is then digitized and analyzed by a computer. The computer applies various algorithms (computer programs; also referred to as post-processing of the data) to modify the data (depending on the anatomy imaged) and displays an image on a computer monitor **(**figure 6**)**.

Charge-Coupling Device (CCD)

A charge-coupling device (CCD) is a solid-state chip that is sensitive to visible light. The CCD chip was developed in the 1970's for military applications, especially night vision scopes. In the 1980's the CCD was used for fluoroscopy imaging and today it is used in most modern home camcorders and digital cameras. A CCD chip is physically small, typically 2 to 4 cm^2 in size. Due to the size of the CCD chip, a radiographic image is optically reduced in size when a single CCD chip is used to obtain a digital image **(**figure 7**)**. It is also possible to tile the CCD chips together to form a larger size detector, but typically no more than 4 CCD chips are tiled in this fashion.

When visible light strikes the photoelectric cathode of the CCD, electrons are released in proportion to the intensity of the light (which is in proportion to the intensity of the x-rays interacting with the scintillator). The number of collected electrons in a given region of the CCD is measured and converted into a digital signal. This digital signal is then displayed (based on intensity of signal strength) as an x-ray image on a computer display.

As mentioned above, CCD digital imaging is used for digital fluoroscopic imaging equipment and other digital x-ray machines. Due to the popularity of CCD chips for consumer products, CCD technology has both matured and grown significantly. An obvious difference between consumer CCD and radiological imaging CCD is color versus gray sensitivity of the chip. A less obvious difference is the number of shades of gray required for diagnostic application (typically 1024 to 4096 gray levels for radiologic applications) whereas for color application the shades of color are typically 256 (but consumer products have CCD chips that are sensitive to red, blue, and green – the primary additive colors and this provides a total of just under 16.8 million color combinations).

Indirect Capture: a-Si Detector

Amorphous silicon (a-Si) detectors use a cesium iodide (CsI) phosphor to convert x-rays to visible light (CsI is not used for screen-film phosphors due to its hydroscopic nature – CsI easily absorbs water and degrades when exposed to air). A photocathode (or photodiode) emits electrons when struck by visible light. A thin film transistor (TFT) detects the number of electrons emitted by the photocathode and converts this to a signal for digitization to the computer. As with CR, the data is post-processed based on the anatomy imaged prior to displaying it on a monitor.

Figure 8 illustrates the conversion steps from (a) capture of an x-ray to (b) conversion of the x-ray into visible light energy to (c) production of electrons and detection by the TFT. **Figure 9** shows a cross-section sketch of the electronics used to read the data provided by an indirect flat-panel detector, while **figure 10** is a magnified photograph of a detector element used by the Canon Medical Systems flat-panel indirect digital detector.

As seen in figures 9 and 10, the detector is comprised of individual elements known as pixels with each pixel being read to obtain the data for image display. The physical size of an individual pixel determines the resolution capability of a detector, or the smallest object that can be visualized in the final image. A detector will be comprised of thousands to millions of pixels to form a single flat-panel plate. The physical size of a detector pixel is dependent on the imaging application and this will be further discussed below under General Properties.

Direct Capture: a-Se Detector

A direct capture detector uses the same thin film transistor technology with the same data readout, as does the indirect a-Si detector. The difference with a direct capture flat-panel detector is that it lacks a scintillator to convert the x-rays to visible light. Amorphous selenium (a-Se) detects the x-ray energy directly. When an a-Se detector captures an x-ray, an electron is produced, commonly referred to as an "electron-hole" in solid state physics/material science terminology **(**figure 11**)**. A voltage is applied across the detector to drive the electric charge towards the TFT to be measured. From this point on, the process is identical to that of indirect detectors for display of the image. **Figure 12** is a sketch of the electronic components of an a-Se detector.

The primary differences between indirect and direct capture flat-panel detectors are summarized as follows: cost to manufacture; number of x-rays captured by the detector, which is dependent upon x-ray energy (the technical terminology for this is called detected quantum efficiency or DQE); relative resolution or visualization of small objects (with CsI used in indirect flat-panel detectors there is some light spread within the CsI crystals); and ability to construct large area detectors for general radiographic imaging ranging from extremity to chest to scoliosis imaging.

General Properties of CR, CCD, a-Si, and a-Se Technologies

In computed radiography, the response of a-Si and a-Se to x-rays is linear over a large range of radiation exposures **(figure 4)**. There is no toe or shoulder region with a-Si as there is with film exposed to x-rays. However, just as with film, too low of an exposure will produce a suboptimal image because there will be excessive noise in the final image resulting from the small number of contributing x-rays.

The concept of optical density as used with film is replaced with "pixel value" for digital imaging. Pixel value represents the shade of white to black that the computer directs the monitor to display. For digital x-ray imaging there typically will be more than 4000 shades of gray in a given digital image (4096 to be exact). When imaging anatomy on a digital detector, useful clinical information in not found in all 4096 shades of gray, so the computer will select a much smaller range of gray levels to display on the monitor. For example, the unattenuated or raw radiation reaching the image receptor will be detected; however, there is no useful information contained in this region and so this portion of the image is removed from computer analysis. Through the use of "windowing" and "leveling", visualization of different portions of gray levels is available for viewing.

In general, the indirect a-Si detectors have a greater sensitivity to x-ray energy (larger percentage of x-rays captured) as compared to a-Se but can result in less resolution capability due to some light spread within the CsI phosphor. Additionally, the absorption efficiency of a-Se is more sensitive to different x-ray energies than is a-Si, also because of the CsI phosphor used in a-Si detectors. There are manufacturers providing indirect and direct detectors for mammographic (25 – 40 keV x-ray energies) as well as for general x-ray imaging (50 – 150 keV x-ray energies). For interventional fluoroscopic applications most institutions use indirect a-Si detectors.

The physical size of an individual pixel of digital detectors depends on the intended application. General radiographic imaging pixel sizes typically range from 150 – 250 microns in size, while for mammographic imaging the pixel sizes will range from 50 – 100 microns to permit better visibility of microcalcifications. The size of the pixel **(figure 13)** determines the resolution capability or how small an object can be to be measured by the detector (the technical terminology is modulation transfer function or MTF). In general terms, resolution is considered to be one of the weak aspects with regard to digital imaging in that conventional radiography (that is, screen-film) has excellent resolution capability that surpasses any digital imaging modality. However, studies have shown that the interpretative quality of digitally acquired images can be equivalent to film and in some studies, the greater contrast range and ability to modify (post-process) the digital image to enhance different regions of the image, allowed for better interpretation with digitally acquired images compared to that of conventional film-based radiographs. New post-processing routines continue to be developed by manufacturers in order to further enhance this interpretative quality of digital imaging.

The methods and means currently available for clinical use of obtaining a digital image have been reviewed. However, an image immediately displayed on a computer monitor without computer manipulation or post-processing will not render a diagnostically acceptable image. Post-processing of the image is required to produce an acceptable image with adequate quality for interpretation. Further elaboration on post-processing, display, storage, and PACS will be discussed in part II.

Figures:

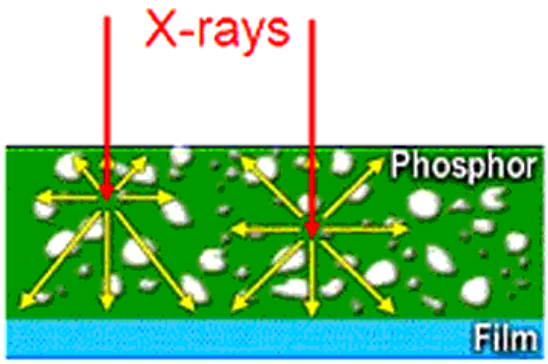

Fig 1: **X-rays (red) absorbed by phosphor and converted to visible light (yellow).**

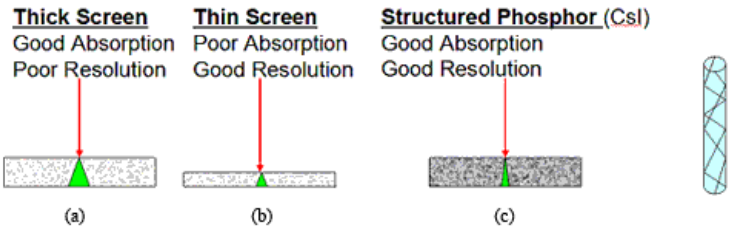

Fig 2: **(a) Thicker phosphor absorbs more x-rays but has reduced resolution as compared to (b). (c) Cesium Iodide (CsI) phosphor are long crystalline needles, maintaining resolution with good absorption efficiency (far right single CsI crystal with light propagation shown).**

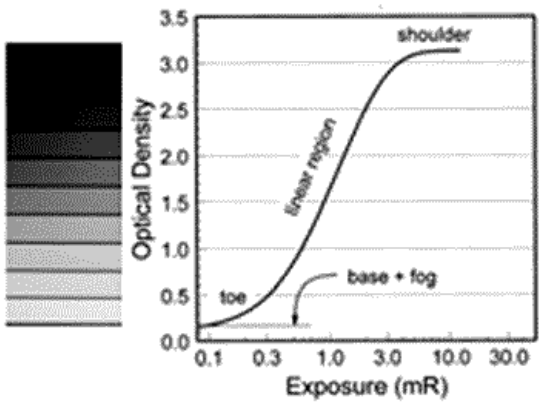

Fig 3: **Gray level (O.D.) of film vs. x-ray exposure (mR). Note semi-logarithmic graph.**

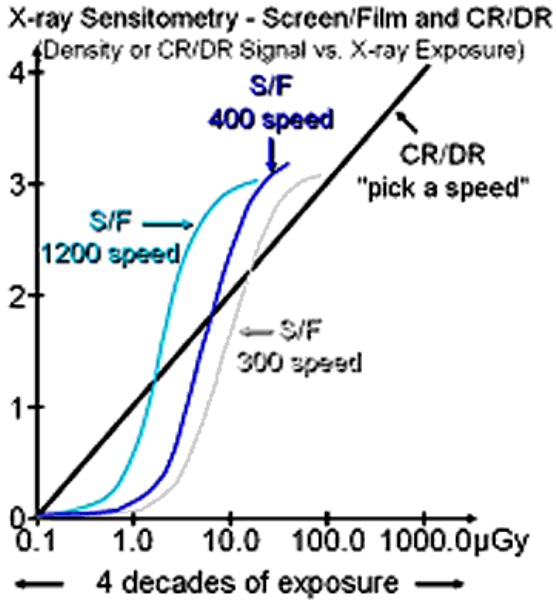

Fig 4: **H&D curves for different screen-film speed systems and for digital detectors.**

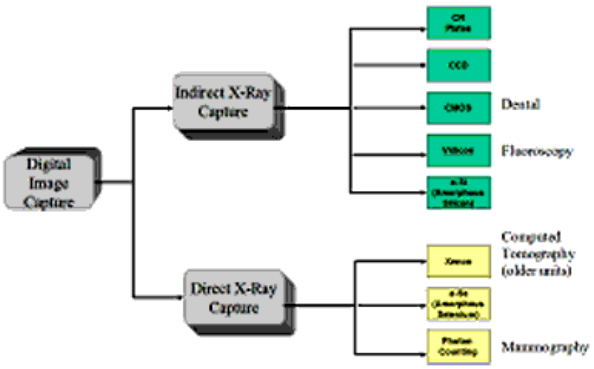

Fig 5: **Illustration of indirect and direct imaging technologies.**

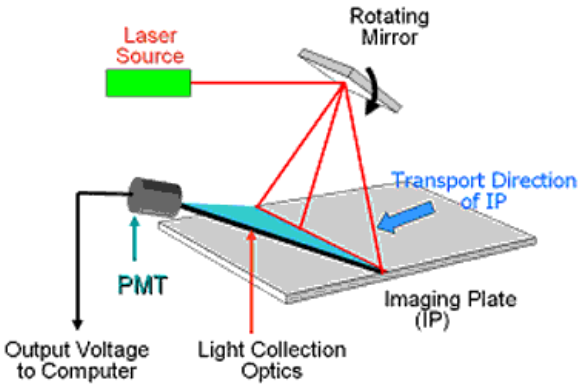

Fig 6: **Schematic of the steps involved with CR imaging.**

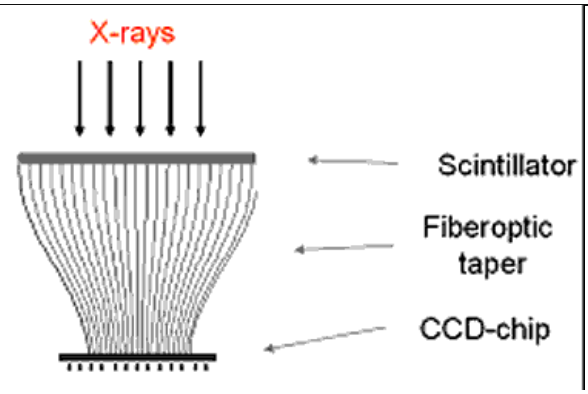

Fig 7: **Schematic of a charge-coupling device acquiring an x-ray image.**

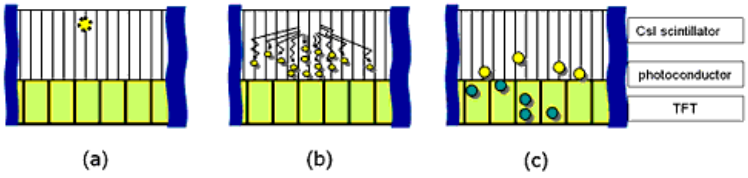

Fig 8: **Illustration of indirect digital imaging: (a) x-ray interacting with CsI phosphor; (b) conversion of x-ray to visible light by scintillator (exaggerated light spread for clarity); (c) conversion of visible light to electrons and capture of electrons by TFT**

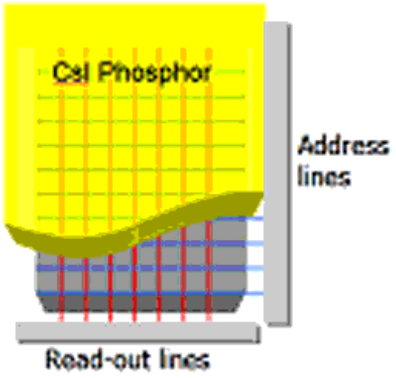

Fig 9: **Cross-section of an indirect flat-panel detector.**

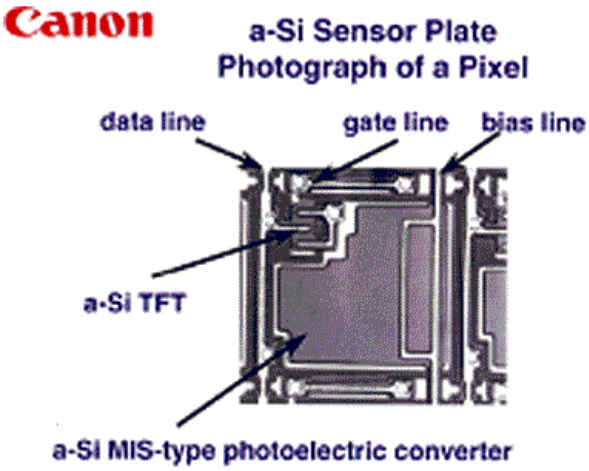

Fig 10: **Photograph of individual a-Si pixel. Courtesy Cannon Medical Systems**

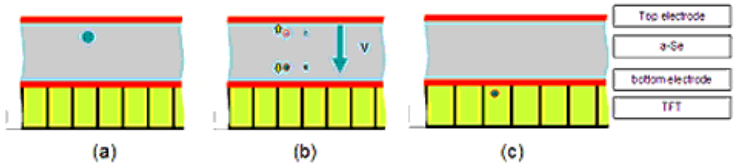

Fig 11: **Direct capture detector: (a) Capture of x-ray by a-Se; (b) conversion of x-ray into electron-hole with applied voltage V; (c) detection of charge by TFT**

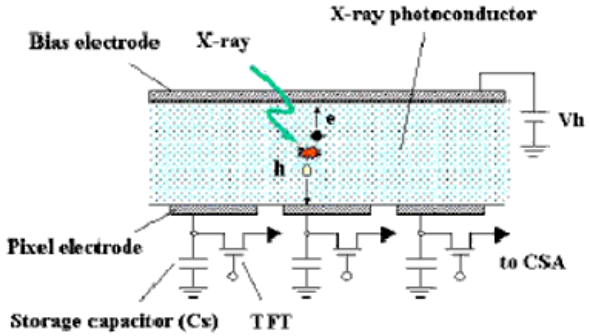

Fig 12: **Diagram of electronic components of a direct capture detector with 3 pixels shown. Vh is the applied voltage and CSA refers to computer system analysis.**

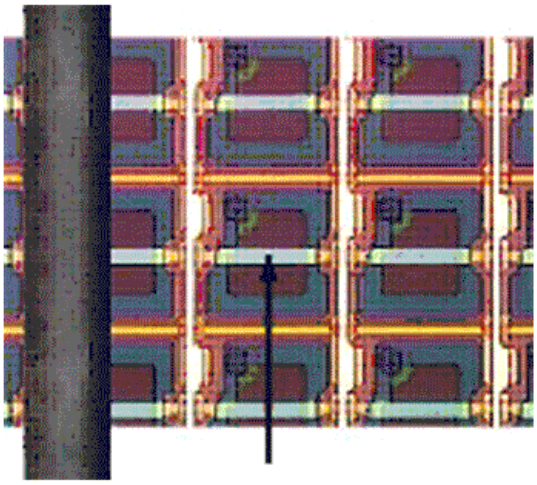

Fig 13: **A human hair as compared to a 140-micron pixel detector.**

1. X-rays (red) absorbed by phosphor and converted to visible light (yellow).
2. (a) Thicker phosphor absorbs more x-rays but has reduced resolution as compared to (b). (c) Cesium Iodide (CsI) phosphor are long crystalline needles, maintaining resolution with good absorption efficiency (far right single CsI crystal with light propagation shown).
3. Gray level (O.D.) of film vs. x-ray exposure (mR). Note semi-logarithmic graph.
4. H&D curves for different screen-film speed systems and for digital detectors.
5. Illustration of indirect and direct imaging technologies.

6. Schematic of the steps involved with CR imaging.
7. Schematic of a charge-coupling device acquiring an x-ray image.
8. Illustration of indirect digital imaging: (a) x-ray interacting with CsI phosphor; (b) conversion of x-ray to visible light by scintillator (exaggerated light spread for clarity); (c) conversion of visible light to electrons and capture of electrons by TFT
9. Cross-section of an indirect flat-panel detector.
10. Photograph of individual a-Si pixel. Courtesy Cannon Medical Systems
11. Direct capture detector: (a) Capture of x-ray by a-Se; (b) conversion of x-ray into electron-hole with applied voltage V; (c) detection of charge by TFT
12. Diagram of electronic components of a direct capture detector with 3 pixels shown. Vh is the applied voltage and CSA refers to computer system analysis.
13. A human hair as compared to a 140-micron pixel detector.

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William R. Geisler is currently the medical physicist at the Marshfield Clinic in Marshfield, Wisconsin. He has extensive experience in the area of medical physics and is directly involved with the Quality Assurance of Computed Radiography. He oversees patient dosimetric calculations and fetal dose calculations, and also determines shielding calculations for general radiography and special room procedures (CT, PET scanning, etc.)

He has served as a speaker at numerous conferences and meetings discussing such topics as full field digital mammography, magnetic resonance physics, and multi-slice computed tomography imaging. He has also conducted numerous inservices regarding imaging physics, radiation safety, and radiation biology.

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