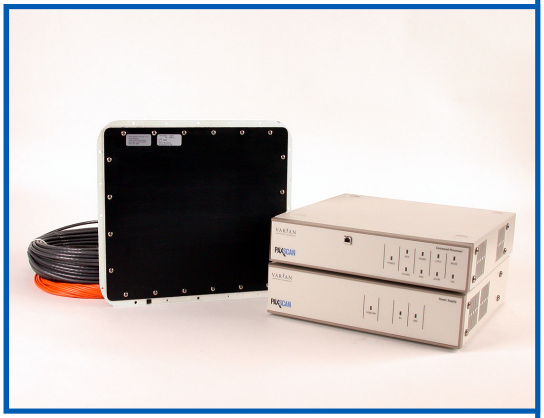
Flat Panel X-ray Imaging

Despite extensive and well-funded research stretching well back into the 70s, the effective design and manufacture of a reliable and affordable digital X-ray radiographic imaging system has proven elusive. Since the mid-80s, radiologists and hospital care-givers have patiently awaited a technology to eliminate film and chemicals in radiography, image plates in computed radiography, and bulky image-intensifiers in fluoroscopy with a smaller, lighter, more portable imager. Finally, late in



the 90s, the first commercially available flat panel digital X-ray detector using amorphous silicon found its way out of Varian's Ginzton Technology Center in Silicon Valley and has begun to change X-ray imaging — forever!

Why Flat Panel?

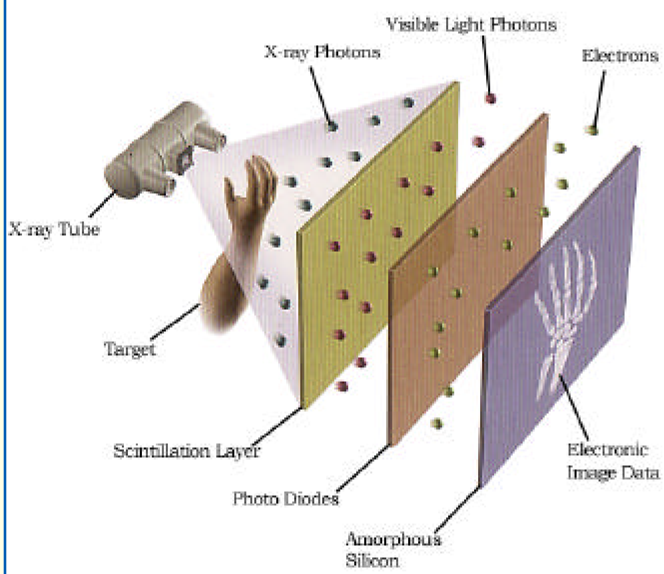
Flat Panel Detectors (FPDs) emerge as next generation digital X-ray technology.

Digital technology has revolutionized our lives. We are collecting, storing, analyzing and using more and more information at a faster and faster pace. X-ray imaging is no exception. The forces behind the digital X-ray revolution are much the same as those driving home and office technologies. Digital devices are smaller and more robust. Once an image is digital, it becomes portable. The image can easily be made available in multiple locations simultaneously. It can be transmitted over long distances in real-time. Digital images make possible computer-assisted diagnosis. Digital images are far simpler to archive and much less costly than their analog counterpart, film. Digital images, video sequences and even volumetric data sets are easily linked to a patient's electronic record. And just as digital technologies have dramatically improved home audio and video fidelity, digital X-ray technology offers significant improvement in image quality and dose utilization.

Many medical modalities, such as CT, PET, SPECT, MRI and ultrasound are inherently digital. However, standard X-ray radiography and fluoroscopy are still primarily based on analog technologies, specifically, screen/film and the image intensifier. Flat panel detectors (FPDs) have emerged as the next generation digital X-ray technology. Flat panel X-ray imagers are based on solid-state integrated circuit (IC) technology, similar in many ways to the imaging chips used in visible wavelength digital photography and video. The primary difference between X-ray imaging and visible imaging is the size of the detector. CCD's and CMOS imagers found in cameras and video recorders are typically on the order of one to two cm in area. Since X-rays are not easily focused, the imager is necessarily on the scale of the object being imaged, which requires an enormous integrated circuit. Fortunately, a large-area IC technology exists in the form of amorphous-silicon thin-film-transistor (TFT) arrays. These TFT arrays are currently in many millions of laptop computers and their application is broadening as these displays increase in size and quality.

A number of detector technologies have been developed based on amorphous silicon TFT arrays, but the most successful and widely used detectors are called "indirect" detectors. These detectors are based on amorphous-silicon TFT/photodiode arrays coupled to X-ray scintillators. The fundamental X-ray conversion chain is shown in Figure 1. The most common scintillators used in flat panel imaging are the same ones used in standard screen/film radiography and fluoroscopy, gadolinium oxysulfide and cesium-iodide (CsI). The success of the indirect FPD technology stems from the fact that both the amorphous silicon and scintillator technologies are well understood and have decades of research behind them.

Figure 1 - Flat panel detector signal chain.



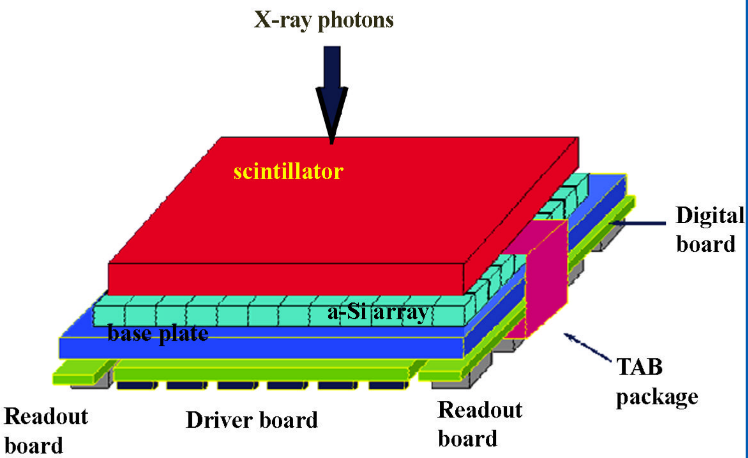
With indirect digital X-ray imaging, an X-ray tube sends a beam of X-ray photons through a target. X-ray photons not absorbed by the target strike a layer of scintillating material that converts them into visible light photons. These photons then strike an array of photodiodes which converts them into electrons that can activate the pixels in a layer of amorphous silicon. The activated pixels generate electronic data that a computer can convert into a high-quality image of the target, which is then displayed on a computer monitor.

With the indirect approach, the flat panel detector consists of a sheet of glass covered with a thin layer of silicon that is in an amorphous, or disordered state. At a microscopic scale, the silicon has been imprinted with millions of transistors arranged in a highly ordered array, like the grid on a sheet of graph paper. Each of these TFTs is attached to a light-absorbing photodiode making up an individual pixel (picture element). Photons striking the photodiode are converted into two carriers of electrical charge, called electron-hole pairs. An electron-hole pair consists of a negatively charged electron and a positively charged hole (a vacant energy space that acts as if it were a positively charged electron).

Since the number of charge carriers produced will vary with the intensity of incoming light photons, an electrical pattern is created that can be swiftly read and interpreted by a computer to produce a digital image.

Although silicon has outstanding electronic properties, it is not a particularly good absorber of X-ray photons. For this reason, X-rays first impinge upon scintillators made from either gadolinium oxysulfide or cesium-iodide. The scintillators absorb the X-rays and convert them into visible light photons that then pass onto the photodiode array.

Figure 2 - Flat panel detector internal architecture.



Because CsI is such an excellent absorber of X-rays, and coverts them to visible light photons at energies that amorphous silicon is best able to convert to charge carriers, the combination of these two materials has the highest-rated Detective Quantum Efficiency DQE) in use today. DQE is the yardstick by which the performance of imagers is measured. A high DQE means images can be acquired with either superior quality or the same quality at a lower dose.

Flat Panel Architecture

The construction of FPDs is similar in many ways to flat panel displays, and uses many of the same technologies. Figure 2 shows the construction of a typical FPD. At the core is an amorphous-silicon TFT/photodiode array. Closely coupled to the array is the X-ray scintillator. Generally, the X-ray conversion screen, rare earth screens such as gadolinium oxysulfide, can be a separate detachable sheet which is mechanically forced into close contact with the array. However, if a CsI screen is used, this is often directly deposited on the array, to give the best optical coupling efficiency. CsI is used in applications like low-dose fluoroscopy, where the photon flux is very low. Figure 3 shows a comparison between the absorption efficiency of CsI and gadolinium oxysulfide. In addition to its much higher absorption efficiency, CsI also produces roughly twice the light output of a gadolinium screen, which results in more than four times the signal at the photodiode for a

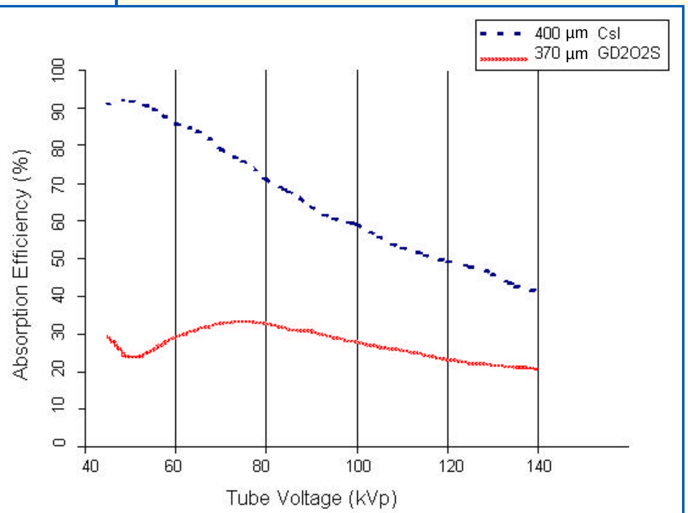
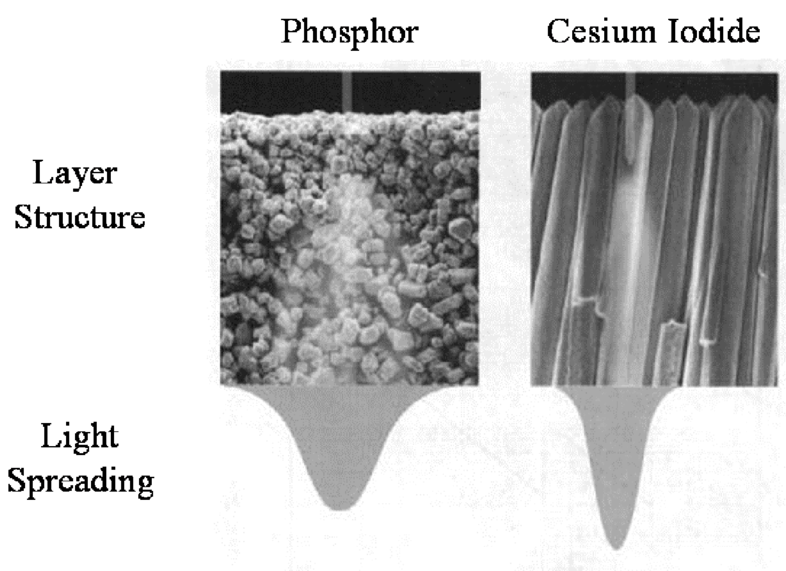


Figure 3 - Absorption efficiency of the primary scintillators used in FPDs. (Image courtesy of Thales Electron Devices.)

given dose. Furthermore, the thickness of the CsI can be greater than that of a rare earth screen because when CsI is deposited on the array it grows in a columnar structure. The columns tend to act as light pipes, reducing the amount of light spreading in the scintillator. So, for example, a 600µm CsI layer can have resolution similar to a 300µm thick rare earth screen. These screens such as gadolinium oxysulfide have the advantage of much lower cost and greater flexibility in that the screen can easily be changed to match the resolution requirements of the application.

Figure 4 - The relative light spreading of columnar CsI versus an amorphous phosphor screen. (Photograph courtesy of Hamamatsu Photonics.)



The light generated by the scintillator is absorbed by the photodiodes in the array, creating electrons which are stored on the capacitance of the photodiode itself. The peak light absorption efficiency of the photodiodes is in the green spectrum, at 550nm wavelength. Both gadolinium oxysulfide and thallium doped cesium iodide, CsI(Tl), produce their peak light output at this frequency. The amorphous- silicon photodiodes are typically the "n-i-p" type. In other words, the layers in the photodiode consist of an electron-rich layer at the bottom, an intrinsic or undoped layer in the middle, and a hole-rich (positively charged) layer at the top. This type of amorphous-silicon photodiode has the advantages of low dark current and a capacitance that is independent of the accumulated signal. The thermally generated dark current intrinsic to the photodiodes is always working to charge the diode. If the photodiodes have large amounts of dark current, much of the diode's signal capacity will be filled up by charge with no signal information. Compared to crystalline silicon photodiodes like those used in CMOS imagers, the dark current in amorphous silicon photodiodes is orders of magnitude less. So it is not unusual for amorphous silicon flat panel arrays to be capable of more than ten second integration times at room temperature. The fact that the diode capacitance is independent of signal helps make the detection system linear.

As is discussed later, the linearity of the detection system is critical to being able to efficiently correct for all the sources of non-uniformity in the array and the electronics.

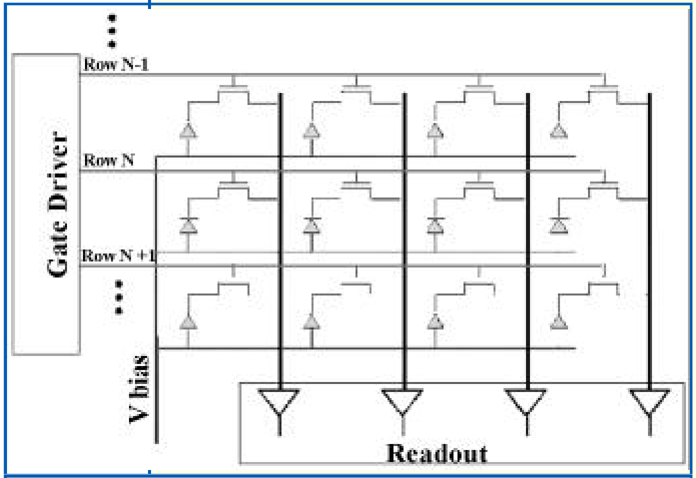
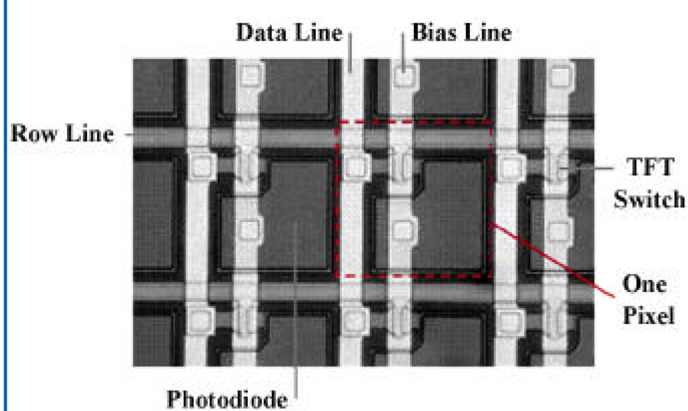


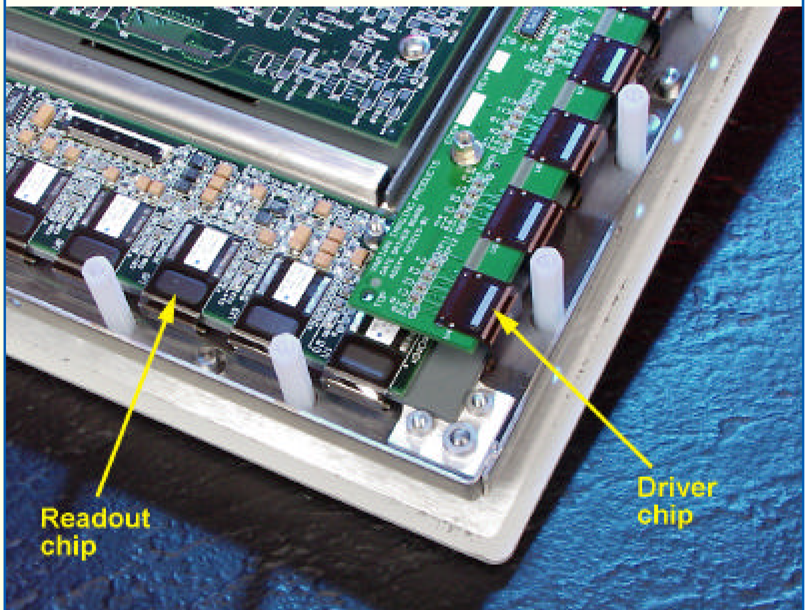
Figure 5 - TFT/Photodiode array schematic and view of a single 127μm pixel.



The TFT/photodiode matrix is normally scanned progressively, one line at a time from top to bottom. The TFTs are essentially switches. When a large positive voltage is applied to one of the gate lines, the switches (TFTs) in the selected row are closed, causing them to conduct electricity. With the TFTs energized, each pixel in the selected row discharges the stored signal electrons onto the dataline. At the end of each dataline is a charge integrating amplifier which converts the charge packet to a voltage. At this point the electronics vary by manufacturer, but generally there is a programmable gain stage and an Analog-to-Digital Converter (ADC), which converts the voltage to a digital number. One important aspect of the scanning is that the FPD is continuously collecting the entire incident signal; none is lost even during the discharge of the pixel. The FPD is an integrating detector and the integration time for each pixel is equal to the frame time.

The electronics can be mounted to the side of the array, out of the beam, as is done in higher energy (MeV) applications to protect against radiation damage. But for diagnostic and interventional procedures, to maintain the best view of the patient, the electronics can also be mounted behind the array and protected by a thin layer of lead. While amorphous silicon has properties sufficient for the detection electronics, it is not suited to the subsequent signal processing. For this reason, every column and row of the array is brought to the edge of the glass, where it is connected to a standard crystalline silicon chip by means of a TAB (tape automated bonding) package. The TAB package bridges the disparity between the connection density on the glass and what a typical circuit board can handle. The glass side of the TAB package may have 128 channels at a 100µm pitch, while the board side connections of the package are on a pitch of 400µm with roughly forty or fewer connections. The chips that need to be directly connected to the array, the readout chip and the driver chip, are mounted in these TAB packages. Figure 6 shows a picture of the row driver and readout chips used in Varian Medical Systems' PaxScan® 4030A, 40x30cm (12"x16") angiography panel.

Figure 6 - Board-side view of TAB packaged row driver and custom ASIC readout chips.

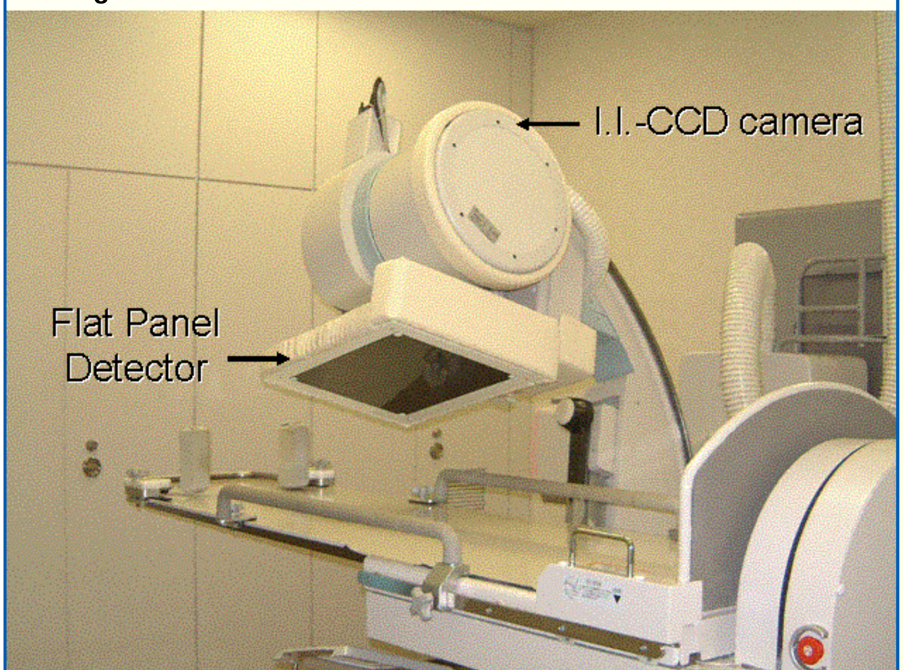


Flat Panel Operational Advantages

The most obvious advantages of flat panel imagers are size and weight. The Image Intensifiers Tubes (IIT) are large and bulky. An FPD with a 12"x16" active area (20" diagonal) takes up less than 25% of the volume of a 12" IIT and less than 15% of that of a 16" IIT. In addition, the FPD takes the place of not only the IIT, but also the attached image recording devices, including the CCD camera, 35mm Cine camera, and the spot film device. The result is vastly improved access to the patient in interventional procedures. In addition to the reduction in size, the weight of the flat panel imager is 60% less than that of the IIT-based imaging chain. Traditionally, the IIT side of the mechanical structure is the heaviest. With flat panels, the heavy side shifts to the X-ray tube, which offers the possibility of a reduction in the bulk and cost of the mechanical superstructure. Mounting a 16" IIT on a mobile C-arm is impractical because of its size and weight, but achieving an even larger active area with a flat panel imager on a mobile C-arm is now straightforward.

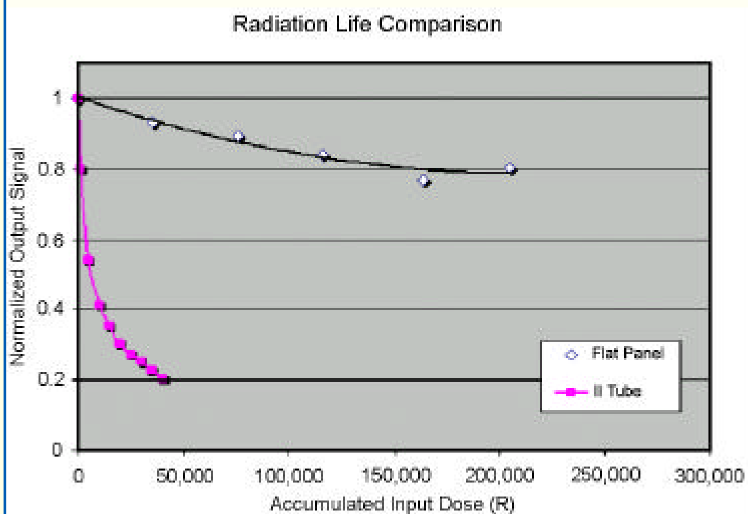
The fact that the recording devices attached to an IIT are not required with FPDs is a result of the multi-mode capability inherent in flat panel technology. From an electrical point of view, the array architecture and readout are very similar to those used for dynamic random access memory (DRAM). Accessing sections or regions-of-interest (ROIs) in the array is only a matter of addressing the proper columns and rows. As with DRAM, the signal is stored as a charge packet, which makes summing the data from more than one pixel a simple matter of combining the charge packets. Reading out a 2x2 neighborhood of pixels as one super-pixel is easily done by combining the signals from neighboring pixels at the front-end charge integrating amplifiers. Pixel binning offers two important opportunities for trade-offs. The first trade-off is sensitivity, since the super pixel will see more X-ray photons and so have higher signal-to-noise ratio (SNR). Very often the maximum digital data conversion rate of the panel is limited to a fixed value. Binning also can reduce the overall matrix size, thus allowing higher frame rates. For example, a 1024x1024 imager capable of 30fps can also be read out as 512x512 super pixels at 60fps or higher. The flat panel imager shown in Figure 7 typically has three modes: full field, full resolution at 7.5fps used for DSA and radiography; full field, half resolution at 30fps used for fluoroscopy and cine; and ¼ field of view, full resolution at 30fps which is used for fluoroscopic zoom mode. Since these modes are defined only by the method of

Figure 7 - This is experimental R&F equipment used by Hitachi in a clinical evaluation of flat panel technology. Here a 12”x16” (active area) FPD is mounted on the side of a 12” Image Intensifier Tube (IIT). With this system, either the FPD or IIT could be rotated into place, facilitating straightforward comparison images.



addressing the array, the panel can switch between modes in less than one second. Flat panels also are more economical than an IIT of comparable size considering that we know IIT image quality deteriorates as a simple consequence of everday X-ray use, and thus IIT's have a relatively short service life. Varian life tests have demonstrated that the service life of a typical flat panel is significantly longer than that of an IIT, assuming the same usage. In the most favorable cases, the flat panel will match the life expectancy of the equipment.

Figure 8 - Signal output comparison vs. X-ray dose between a PaxScan 4030A and an IIT of comparable size.



Flat Panel Image Quality

Beyond the improved form factor and flexibility of flat panels, there are also enhancements to the image quality. In IIT-based imaging there are many stages in the signal conversion chain, each of which is subject to losses and distortion. For example the electron optics inside the IIT is influenced by the earth's magnetic field. And the optics of the output phosphor is subject to veiling glare, which is a kind of a long-range crosstalk. The result is geometric distortion and brightness non-uniformity across the diameter of the image. In comparison, flat panels have a very direct, short signal conversion path, with essentially no optics.

Figure 9 - Comparison between a 12" IIT system and GE's 20cm Revolution flat panel detector. Clearly the image intensifier has both distortion and brightness non-uniformity which is absent from the flat panel detector. (Image courtesy of GE Medical Systems.)

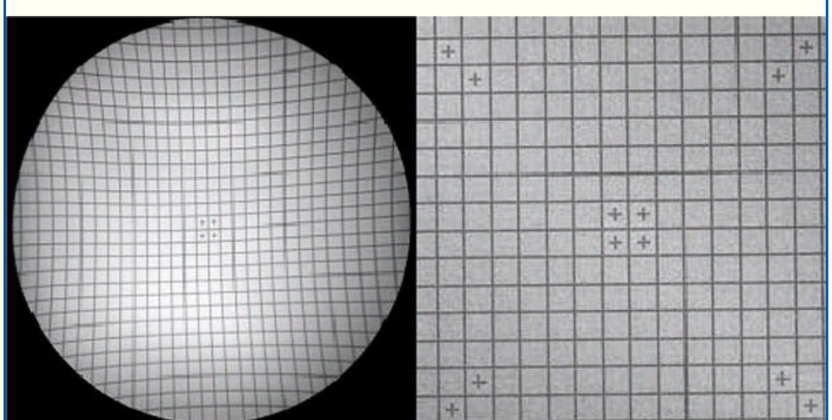
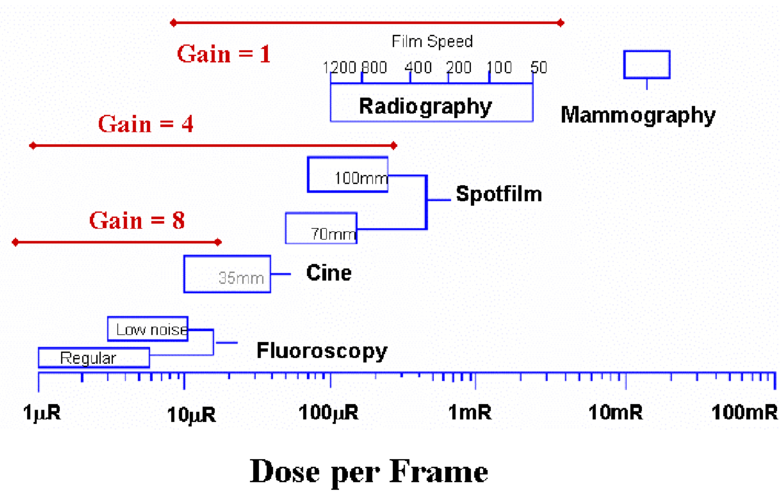


Figure 10 - Entrance dose per frame by application.



The result is a very flat, uniform "film-like" image from edge-to-edge. Similar to the distortion and brightness non-uniformity, the IIT/CCD image chain also suffers from contrast loss at the edges of the image, which is why the IIT performance is always specified at the center.

The ability of flat panel detectors to encompass multiple X-ray modalities is also a function of their very large dynamic range. Figure 10 shows the detector entrance dose per frame by application. An FPD designed for R&F applications can cover the entire range, from low-dose fluoroscopy to radiography. Generally, the Analog-to-Digital Converters (ADCs) inside the FPD have a fixed input range. In order to adapt the signal to that range, there is an analog gain stage prior to digitizing. Higher gain is used for smaller signals, but at the expense of the maximum dose per frame. Because of the very different resolution, energy and dose requirements, mammography requires a specialized flat panel detector. The large exposure latitude in flat panel imaging means that retakes, due to over and under exposure, are virtually eliminated. The large dynamic range also enables computed tomography applications, where the entrance dose per projection can vary from the low dose fluoroscopy level to the equivalent of radiography shots.

Figure 11 - Signal vs. entrance dose for a 194μm, 40x30 cm FPD.

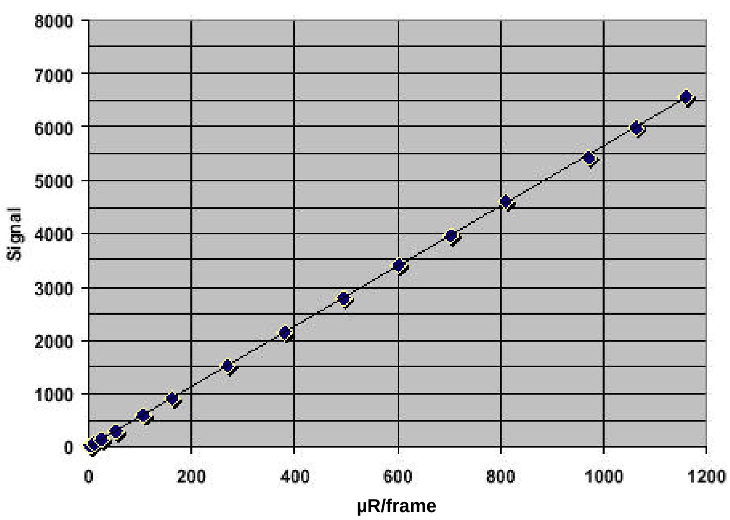
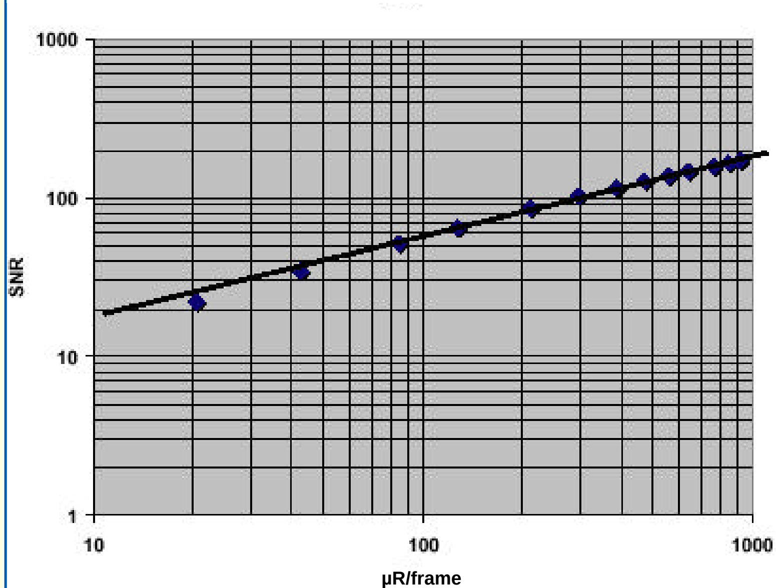


Figure 11 shows the signal response of an angiography flat panel in the full resolution mode, over the dose range of 1μR to 1.2mR. Particularly for the amorphous silicon TFT/photodiode technology, the response to entrance dose of the FPD is extremely linear. The response of the imager deviates from the straight line curve by less than 0.01%. This degree of linearity is necessary to facilitate the real-timeoffset and gain correction that is normally performed for each pixel. Because each pixel in the array has its own gain value and zero signal level, the image would be very non-uniform without normalization. If the imager response is non-linear, the arithmetic for normalizing the pixel response in real-time becomes intractable.

Of equal importance is the Signal-to-Noise (SNR) behavior versus dose. The FPD contributes electronic noise to the image. However, if the electronic noise is low enough, the statistical noise in the X-ray beam is dominant. The noise in the beam follows Poisson statistics. In other words, the noise is equal to the square root of the number of incident X-ray photons.

As shown in Figure 12, the SNR of an FPD has a square root dependence on dose, i.e. is X-ray quantum limited over a very large range. This is an indication that the detector contributes effectively no noise to the image over this dose range.

Figure 12 - SNR vs. entrance dose for a 194μm, 40x30cm FPD.



A more accurate way to look at the sensitivity of an imaging system is to evaluate the Detective Quantum Efficiency (DQE) as a function of spatial frequency. Particularly for digital X-ray technologies, this single measure is a common reference point that accounts for noise, signal-loss mechanisms, sensitivity and resolution. Essentially the DQE is equal to the square of the ratio of the imager's output SNR to its input SNR.

$$DQE\left(\bar{q},u\right)=\frac{SNR_{out}{}^{2}}{SNR_{in}{}^{2}}$$

DQE is typically calculated using the formula:

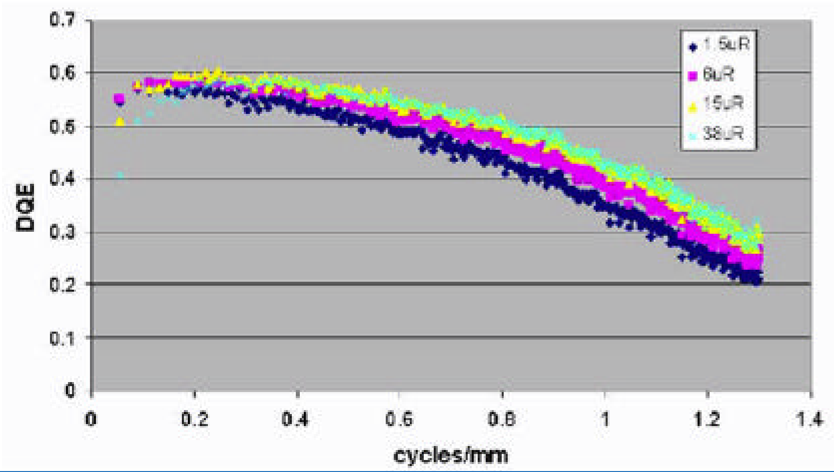
$$DQE\left(\bar{q},u\right)=\frac{\bar{d}^{2}*MTF^{2}\left(u\right)}{\bar{q}*NPS\left(\bar{q},u\right)}$$

where, for a given input dose, **d** is the average output signal produced, MTF is the modulation transfer function (a measure of resolution), **q** is the number of incident X-ray quanta and NPS is the noise power spectrum produced by the imager. Because the X-ray beam statistics are Poisson distributed, the input SNR 2 is in fact equal to the total number of X-ray photons, **q**. The rest of the equation defines the output SNR 2 as a function of signal, resolution and noise.

The DQE as a function of entrance dose (fluoro and cine range) for an angiographic flat panel imager is shown in Figure 13. This is typical of the indirect TFT/photodiode technology with a CsI scintillator. Because of the low-loss, high-absorption detection path, the DQE for indirect CsI-based flat panels is the highest available and is more than double that of computed radiography, film screen and CCD based technologies. Higher DQE translates directly into better imager quality for a given dose.

So with high DQE detection systems, it is possible to get the same image quality as a low DQE system like screen film at a fraction of the dose.

Figure 13 - DQE vs. spatial frequency over the fluoroscopic and cine dose range, for a 40x30cm angiographic flat panel.



Digital X-ray Radiography

The numerous advantages of flat panel detectors over X-ray film, computed radiography image plates, and image-intensifier tubes is clear. This is evident to the thousands of leading radiologists and cardiologists performing routine X-ray examinations, test engineers inspecting for cracks in aircraft structures, and security forces screening for and disarming explosives. Digital X-ray imaging available instantly just like that of modern digital cameras and camrecorders is the future. Please contact your local Varian sales office for the latest on our PaxScan family of flat panel X-ray detectors.

X-Ray Products

Headquartered in Salt Lake City our flat panel team is committed to meeting the diverse and growing needs of Digital Radiography in many markets: Medical, industrial/NDT, security, and scientific. World renown for our X-ray tubes, flat panel X-ray detectors are an important addition to our growing X-Ray business.

MISSION STATEMENT

To be the unquestioned worldwide leader in X-ray products by achieving superior product quality and customer satisfaction.

Ginzton Technology Center

The Ginzton Technology Center acts as Varian Medical Systems' research and development facility for breakthrough technologies and operates a brachytherapy business for the delivery of internal radiation to treat cancer. In addition to brachytherapy, current efforts are focused on emerging flat panel technologies. This business also conducts externally funded research related to medical technology, which leads to long-term partnerships and new business opportunities.

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