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# THE DIGITAL FLAT-PANEL X-RAY DETECTORS

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***Abstract* – In a digital imaging system, the incident x-ray image must be sampled both in the spatial and intensity dimensions. In the spatial dimensions, samples are obtained as averages of the intensity over picture elements or pixels. In the intensity dimension, the signal is digitalized into one of a finite number of levels or bits. Two main types of digital flat-panel detectors are based on the direct conversion, which contains the photoconductor, and on indirect conversion, which contains phosphor. The basics of these detectors are given. Coupling traditional x-ray detection material such as photoconductors and phosphors with a large-area active-matrix readout structure forms the basis of flat panel x-ray imagers. Active matrix technology provides a new, highly efficient, real time method for electronically storing and measuring the product of the x-ray interaction stage whether the product is visible wavelength photons or electrical charges. The direct and indirect detectors, made as the active-matrix flat-panel detectors containing sensing/storage elements, switching elements (diodes or thin film transistors (TFTS)) and image processing module, are described. Strengths and limitations of stimulable phosphors are discussed. The main advantages and disadvantages of mentioned x-ray detectors are also analyzed.**

***Keywords* – X-ray, digital detector, photoconductor, phosphor**

## 1. INTRODUCTION

X-ray film has been most popular in medical imaging. It is cheap, simple to, and readily available in large sheets. Storage phosphors, also known as imaging plates and first commercially developed in Japan in the 1980s, are an alternative to film. In practice, the storage phosphor plate is exposed to x rays and is read out by raster scanning the plate with red laser light. Medical imaging requires that a radiologist be able to see sufficient detail and contrast to make a diagnosis.

A new generation of large-area, flat-panel detectors with integrated, thin-film transistor promises very rapid access to digital images. As digital radiography continues this rapid evolution, it is likely that radiologists will be inundated with information concerning a wide variety of large-area, flat-panel electronic detectors [1-3].

The basis of two digital detector types, direct and indirect detectors, is given. The direct and indirect detectors, made as the active-matrix flat-panel detectors containing sensing/storage elements, switching elements (diodes or thin film transistors (TFTS)) and image processing module, are described. Their main advantages and disadvantages are analyzed. Strengths and limitations of stimulable phosphors are also discussed.

## 2. DIGITAL X-RAY IMAGING

### 2.1. Basis of digital x-ray detectors

In a digital imaging system, the incident x-ray image must be sampled both in the spatial and intensity dimensions. In the spatial dimension, samples are obtained as averages of the intensity over picture elements or pixels. These are usually square, and spaced at equal intervals throughout the plane of the image. A fraction of pixel that is sensitive to the incoming signal is the geometrical fill factor. In the intensity dimension, the signal is digitalized into one of a finite number of levels or bits. The pixel size and the bit number must be appropriate chosen for a given imaging task. Each pixel typically contains a switching element and a sensing/storage element.

Coupling traditional x-ray detection materials such as phosphors or photoconductors with a large-area active-matrix readout structure forms the basis of flat-panel x-ray imagers. Active matrix technology provides a new, highly efficient, real time method for electronically storing and measuring the product of the x-ray interaction stage whether the product is visible wavelength photons or electrical charges. Three steps in image creation are: 1) detection, 2) storage, and 3) measurement stages.

Two types of digital flat-panel x-ray detectors (Fig. 1) exist, depending on detection types:
- Direct detection that incorporate a photoconductor to produce electrical charges on detection of an x-ray,
- Indirect detection that incorporate a phosphor to produce visible photons on detection of an x-ray.

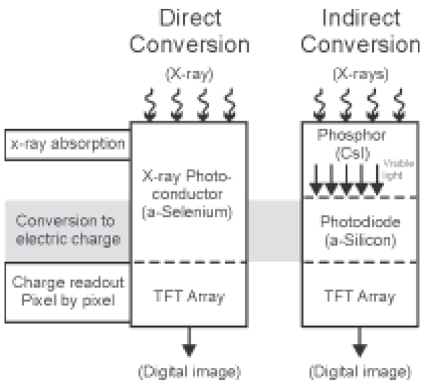


*Fig. 1 – Direct and indirect conversion*

For direct conversion a thick (~0.5 to 1 mm) amorphous Se (a-Se) photoconductive layer (Z = 34) is usually used. The pixels incorporate a conductive electrode to collect charge and a capacitor element to store it. Interacting x-rays produce charge in the photoconductive which is then shared between the inherent capacitance of the photoconductive layer and the pixel-storage capacitance (Fig. 2).

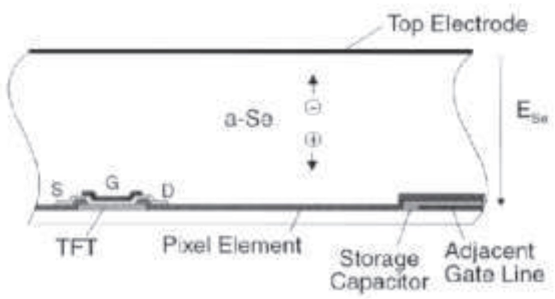


*Fig. 2 – Cross section of a single pixel with a-Se*

In the band structure for a photoconductor shown in Fig. 3a an optical photon with energy Eg can excite an electron from the valence band to the conduction band leaving behind a hole in the valence band (internal photoelectric effects). The energy of light photons is between 1 and 3 eV, but band gap Eg ~ 2 eV for photoconductors (Eg = 2.2 eV for a-Se). Else, Eg ~ 1eV for semiconductors (e.g. 1.1 eV for Si).

For the high energetic x-rays having energy thousands of times higher energy than Eg, the rules are different. The many materials are using in diagnostic imaging have high atomic number Z, and the absorption of diagnostic x-rays is dominant by photoelectric effect. A very energetic electron is liberated, which passes through the material, causes further ionization. Under these circumstances, the amount of energy necessary to create electron-hole pair is not simply Eg, but 3Eg.

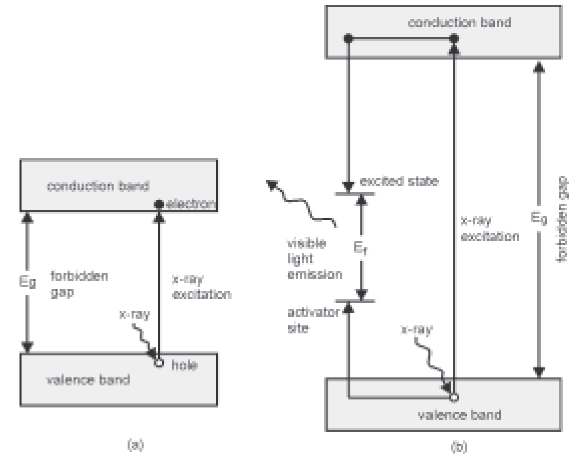


*Fig. 3 – The electronic band structures of a) photoconductors & semiconductors, and b) phosphor*

In indirect conversion, a phosphor layer is placed in intimate contact with an active matrix array (Fig. 4). The intensity of the light emitted from a particular location of the phosphor is a measure of the intensity of the x-ray beam incident on the surface of the detector at that point. Each pixel on the active matrix has a photosensitive element that generates an electrical charge whose magnitude is proportional to the light intensity emitted from the phosphor in the region close to the pixel. This charge is stored in the pixel until the active-matrix array is read out.

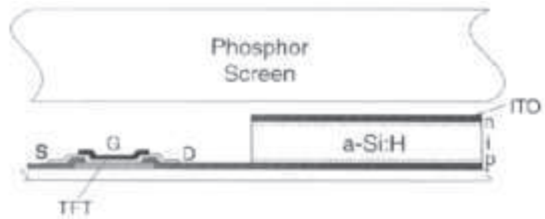


*Fig. 4 – Cross section of a single pixel with phosphor*

The imaging system is completed with peripheral circuitry that amplifies, digitizes, and synchronizes the readout of the image and a computer that manipulates and distributes the final image to the appropriate soft- or hard-copy devices (Fig. 5).

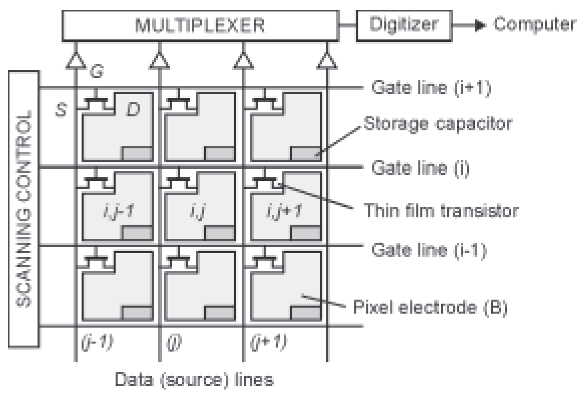


*Fig. 5 – The layout of pixel groups on active-matrix-array*

### 2.1.1. *Photoconductor in direct conversion*

The material used as the x-ray photoconductor is not the pure form of a-Se. Pure a-Se is not thermally stable and tends to crystallize over time. However, the crystallization of a-Se can be prevented be alloying a-Se with about 0.5 % As, denotes as a-Si: 0.5 % As.

a-Se advantages are:
- Could be easily and cheaply made in large areas by a low-temperature process,
- Uniform in imaging properties to a very fines scale (an amorphous material is entirely free from granularity),
- There no free carriers at room temperature.

a-Se disadvantages:
- Very high voltage needed to activate a-Si layer (~10 V/μm); under fault conditions, this voltage could damage the active-matrix array,
- Atomic number Z = 34 is rather low and requires very thick layers for high quantum efficiency at diagnostic energies (~ 100 keV).

Alternatives for photoconductors are: $PbI_2$, PbO, TlBr, and potentially: CdZnTe, CdTe, CdSe and $HgI_2$.

### 2.1.2. *Phosphor in indirect conversion*

A phosphor is common name for a material that glows after exposure to radiation. Many current x-ray imaging detectors employ a phosphor in the initial stage to absorb x rays and produce light. Phosphor work by exciting electrons from the valence band to the conduction band where they are free to move a small distance within the phosphor (Fig. 3b). Some of these electrons will decay back to the valence band without giving the visible photons, but in an efficient phosphor, many of electrons will return to the valence band through a local state created by small amounts of impurities called activators, emitting the light (Fig. 3b). Else, the electrons will be firstly trapped at excited state of activator site (so-called recombination centre), and then be recombined with hole reached from valence band emitting the light.

Thus, phosphor can be relatively efficient converters of the large incident energy of the x-ray into light photons. Because light photons each carry only small (~2-3 eV) energy, many light photons are created from the absorption of a single x-ray. This quantum amplification is the conversion gain of the phosphor.

For instance, CsI:Tl during irradiation produces the light photons with wavelength of 600 nm corresponding to energy of 2 eV. If it is irradiated by 60 keV, 30 000 light photons should be expected. However, energy necessary for a light photon production is not 2 eV, but 18 eV (~3Eg; Eg = 6.2 eV for CsI), and 3300 light photons are produced per a x ray. Else, the band gapes for phosphors are from 5 to 10 eV.

One of the main issues with the phosphor is the balance between spatial resolution and x-ray detection. The thicker phosphor the more x-ray is absorbed, the spatial resolution is worst, since the emitted light can spread further from the point of production before existing the screen. The conflict could be solved using a needle like structure (structured phosphor, similar to optic fibers), as CsI phosphor (83 % of the emitted light will undergo internal reflection).

The besides main advantage of very high resolution comparing to other phosphor types the main disadvantages of CsI are: hydroscopic nature, toxicity, and lack of mechanical robustness.

## 2.2. Active-matrix flat-panel detectors

The first active-matrix flat-panel display was demonstrated by Brody in 1973, and was fabricated using CdSe as the semiconductor. The majority of today's flat-panel arrays are fabricated from hydrogenated amorphous silicon (a-Si:H). The role of hydrogen is to occupy many of the dangling bonds present in raw a-Si. One significant advantage of a-Si:H for medical imaging flat-panel active-matrix devices is that the layers of a-Si:H can be deposited over extremely large areas (exceeding 1x1 $m^2$).

The main parts of active-matrix array pixels are:

1) Sensing/storage elements, and

2) Thin-film switching elements.

1) Sensing and storage elements are photodiodes for the indirect approach and storage elements are capacitors for the direct approach.

2) Thin-film switching elements are:
- two-terminal devices, as diodes, metal-insolator-metal devices (MIMs), metal-semiconductor-isolator devices (MSIs), …
- three-terminal devices, as thin field transistors (TFT).

### 2.2.1. *Sensing/storage elements for indirect detection*

The photodiodes as sensing elements in indirect pixel are designed to detect visible lights. The photodiodes also acts as a capacitor to store the photogenerated charge. The absorption coefficient of a-Si:H (~ 104 to 106 cm-1) for visible lights is an order of magnitude higher than that for crystalline silicon (~ 103 to 105 cm-1) even though the effective optical band gap of a-Si:H (1.7 eV) is larger than that in crystalline silicon (1.1 eV). A 0.5 μm-thick layer of undoped (i.e., interstic) a-Si:H is sufficient to absorb the most lights entering the layer, but often layer thickness of the order of 1 to 2 μm may be used to reduce pixel capacitance (Fig. 6).

Fig. 7 shows the absolute quantum efficiency as a function of photon wavelength in the visible range for an ~1.5 μm thick n-i-p photodiode at -5 V reverse bias. Also in figure are the emission spectra for typical x-ray phosphors. It can bee seen that the absorption of a-Si:H is well matched to the emission from these phosphors.

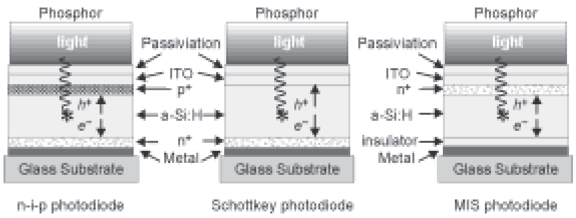


*Fig. 6 Cross section of possible a-Si:H photodiode structures: (a) n-i-p, (b) Schottky, and (c) metal-insulator-semiconductor (MIS)*

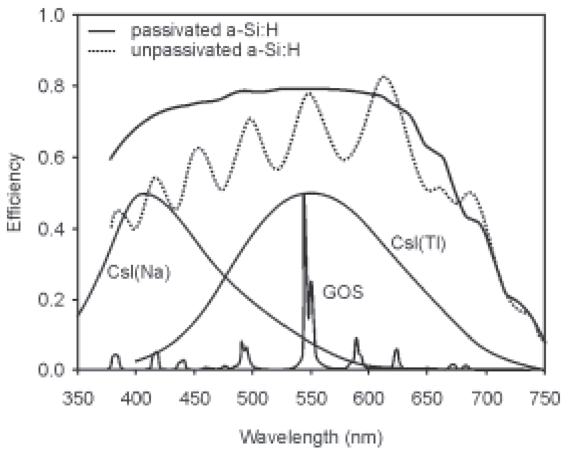


*Fig. 7 – Quantum efficiency of intrinsic a-Si:H n-i-p photodiode and various x-ray phosphors*

### 2.2.2. Sensing/storage elements for direct detection

In the direct detection approach, the photoconductor is sensing element which converts incoming x-ray into charge. A simple charge storage capacitor is storage element that require only dielectric and metal layer. The storage capacitor is electrically connected to the overlying photoconductor at pixel electrode.

A bias voltage must be applied across the thickness of the photoconductor to facilitate the separation and collection of x-ray induced charge. To maintain an internal field of 10 V/μm within a 500-μm-thick layer of a-Se, a bias of 5000 V must be applied. Three methods incorporate a means of draining away the potential on the pixel if it exceeds a predetermined safe design value (Fig. 8):

- including an extra component parallel with the storage capacitor (e.g., a Zener dioda),

- to modify the TFT by incorporating of a second gate connected to the pixel electrode to ensure that its channel current will increase to drain away the excess charge when the pixel potential approaches damaging levels,

- to reverse the a-Se structure to permit a negative bias on the top electrode to ensure that the ordinary TFT will start to conduct when the pixel potential reaches the threshold voltage.

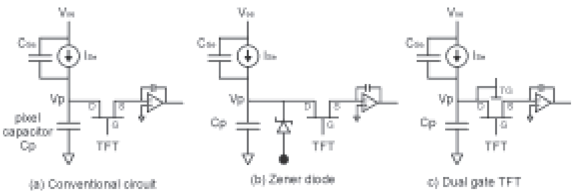


*Fig. 8 – Three methods for high voltage protection of active matrix in direct conversion flat panel imagers*

### 2.2.3. Switching elements: diodes

The other main component of the pixels on an active-matrix array is the switching element, and there are a number of different possibilities for its design. One is using diodes for both the sensing and switching elements, and the diodes are usually of the same type (i.e., either all Schottky or all n-i-p diodes). Fabricating both diodes at the same time reduces the number of mask levels and consequently improves the device yield. Advantage is easy fabrication, and disadvantages are: create a image lag (offset image) and high capacitance.

### 2.2.4. Switching elements: thin film transistor (TFT)

The switching devices with the best properties for the majority of medical imaging applications is the a-Si:H thin film transistor (TFT; see Fig. 9). Silicon nitride (a-$Si_3N_4$:H) is the usual gate dielectric. The positive gate voltage (~+10 to +15 V) is used to switch device on and a negative voltage (~-5 to -10 V) to switch it off. The on to off ratio for the current through this device is extremely high (> 1010), so that they have excellent switching properties. A radiation part is absorbed by TFT, and although it should be radiation hard, the TFT threshold voltages are usually changed.

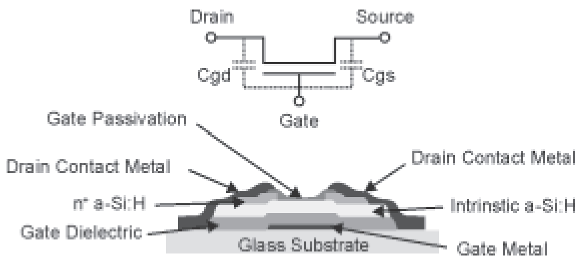


*Fig. 9 – Cross section of TFT*

### 2.2.5. Image processing and manipulation

Because of limited tolerances achievable in controlling the thickness and quality of the different layers on large-area flat panel x-ray imagers, the sensitivity of the pixels to the radiation and their offset signals (from dark current integration and switching transients) vary from pixel to pixel. This includes variations resulting from the thickness and quality of the photoconductor or phosphor layers coupled to the arrays. Tolerance issues in the fabrication of the peripheral amplification and controlling circuitry also add variations in the pixel sensitivity and offset. The most powerful method for removal of these distracting effects is called flat fielding.

Variations in pixel and electronic offsets are corrected using dark-field images acquired with no x-ray exposure. Pixel sensitivity and electronic gain variations are corrected using flood-field images taken with a constant intensity of x-ray exposure across the full area of the detector. Measured drifts of pixel sensitivity are extremely small, even over extended periods of time, so acquisition of flood field data is not needed as frequently as offset correction. A carry over or lag or ghosting may be observed after large exposures, and it may sometimes be necessary corrected. This effect is much more emphasized in direct than in indirect detectors.

## 2.3. The photostimulable (storage) phosphor

One of the most popular and most successful detectors for digital radiography to date have been photo stimulable phosphors, also known as storage phosphors. These phosphors are commonly in the barium fluorohalide family, typically BaFBr:Eu+2, where the atomic energy levels of the europium activator determine the characteristics of light emission. X-ray absorption mechanisms are identical to those of conventional phosphors. They differ in that the useful optical signal is not derived from the light that is emitted in prompt response to the incident radiation, but rather from subsequent emission when electrons and holes are released from traps in the material (Fig. 10).

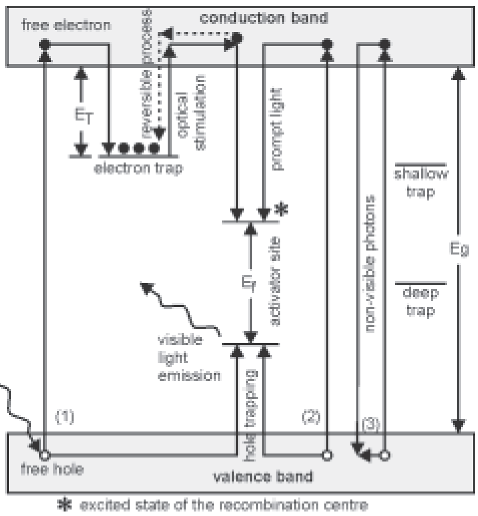


*Fig. 10 – The electronic band structures of photostimulable (storage) phosphor*

The electrons liberated during irradiation are excited to the conductor band, and they either produce light promptly or are stored in traps. The phosphor is intentionally designed to contain traps which store the electrons ("electron traps"). Because the 'prompt' light is not of interest in this application, the efficiency of the storage function can be improved by increasing the probability of electron trapping. On the other hand, when trapped electrons are released by the stimulating light during readout, the probability of their being retrapped instead of producing light, would then be higher, so that the efficiency of readout would be reduced. The optimum balance occurs where the probabilities of an excited electron being retrapped or producing fluorescence (light) are equal. This causes the conversion efficiency to be reduced by a factor of four compared to the same phosphor without traps, i.e. a factor of two from the prompt light given off during x-ray exposure and another factor of two from unwanted retrapping of the electrons during readout. In addition, the decay characteristics of the emission must be sufficiently fast that the image can be read in a conveniently short time while capturing an acceptable fraction of the emitted energy. In practice, depending on the laser intensity, the readout of a stimulable phosphor plate yields only a fraction of the stored signal. This is a disadvantage with respect to sensitivity and readout noise, but it can be helpful by allowing the plate to be 'pre-read', i.e. read out with only a small part of the stored signal, to allow automatic optimization of the sensitivity of the electronic circuitry for the main readout.

In the digital radiography application, the imaging plate is positioned in a light-tight cassette, exposed and then read by raster scanning the plate with a laser to release the luminescence (figure 10). The emitted light is collected and detected with a photomultiplier tube whose output signal is digitized to form the image. Finally, an adequate wavelength separation between the stimulating and emitted light quanta is necessary to avoid contaminating the measured signal.

### *2.3.1. Strengths and limitations of stimulable phosphors*

The photostimulable phosphor is an excellent detector for digital radiography since, when placed in a cassette, it can be used with conventional x-ray machines. Large-area plates are conveniently produced, and images can be acquired quickly.

The plates are reusable, have linear response over a wide range of x-ray intensities, and are erased simply by exposure to a uniform stimulating light source to release any residual traps. One limitation of this type of detector is that because the traps are located throughout the depth of the phosphor material, the laser beam providing the stimulating light must penetrate into the phosphor. Scattering of the light within the phosphor causes release of traps over a greater area of the image than the size of the incident laser beam spot. This results in loss of spatial resolution, which is emphasized if the plate is made thicker to increase x-ray absorption.

The main advantage of this system, comparing to the flat-panel detectors, is reusability. The storage phosphor may be used a lot of times without an information losing. In the flat panel detector the some pixels are damaged by radiation and ghost image will appear after certain time. The main disadvantage is very high price of image reader, and impossibility of real-time imaging (fluoroscopy).

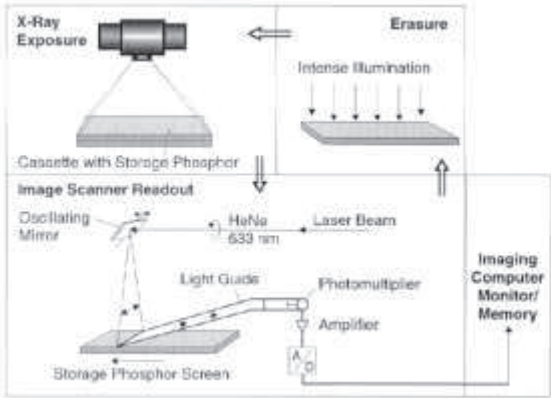


*Fig. 11 – A storage phosphor digital system*

## 2.4. Digital system for fluoroscopy

Fig. 12 shows an x-ray image intensifier (XRII) that is currently only one digital system for fluoroscopy that represent real-time visualization. The resulting real-time images are usually displayed using a video system (conventional or CCD) optically coupled to the x-ray image intensifier. The XRII absorbs the incident x-ray image, amplifies and outputs it as an optical image which is then distributed by lenses to the video camera. X-rays are converted to light in the large input phosphor screen typically of 12.5 cm to 40 cm in diameter. The fluorescence illuminates a photocathode evaporated directly on the phosphor and liberates electrons. The purpose of the photocathode is to convert light photons to electrons efficiently. The electrons are accelerated through a large potential difference (typically 25 kV) and electrostatically focused by the electrodes onto a small (2.5 cm diameter) output phosphor.

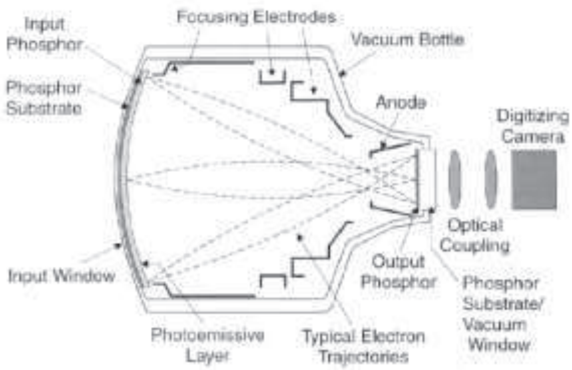


*Fig. 12 – XRII system for digital imaging*

The main disadvantages of digital systems based on the use of XRIIs are loss of image contrast due to x-ray and light scatter within the tube, the geometric distortion of the image due to the curved input phosphor, and an influence of earth's magnetic field.

## 3. CONCLUSION

There are two main types of digital flat-panel x-ray detectors that are based on:

- Direct detection that incorporate a photoconductor to produce electrical charges on detection of an x-ray,
- Indirect detection that incorporate a phosphor to produce visible photons on detection of an x-ray.

For direct conversion a thick (~0.5 to 1 mm) amorphous Se (a-Se) is usually used. The pixels incorporate a conductive electrode to collect charge and a capacitor element to store it.

In indirect conversion, a phosphor layer is placed in intimate contact with an active matrix array.

a-Se advantages are: could be easily and cheaply made in large areas by a low-temperature process, uniform in imaging properties to a very fines scale (an amorphous material is entirely free from granularity), and there no free carriers at room temperature. It disadvantages are very high voltage needed to activate a-Si layer (~10 V/μm), which could damage the active-matrix array, low atomic number (Z = 34), requiring very thick layers for high quantum efficiency. Promising alternatives for a-Se could be: $PbI_2$, PbO, TlBr, and potentially: CdZnTe, CdTe, CdSe and $HgI_2$.

In the case of phosphor, there is the balance between spatial resolution and x-ray detection. The thicker phosphor the more x-ray is absorbed, the spatial resolution is worst, since the emitted light can spread further from the point of production before existing the screen. The conflict could be solved using a needle like structure (similar to optic fibers), as CsI phosphor. The besides main advantage of very high resolution comparing to other phosphor types the main disadvantages of CsI are: hydroscopic nature, toxicity, and lack of mechanical robustness.

The main advantage of photostimulable phosphor system, comparing to the flat-panel detectors, is reusability. The storage phosphor may be used a lot of times without an information losing. In the flat panel detector the some pixels are damaged by radiation and ghost image will appear after certain time. The main disadvantage is very high price of image reader, and impossibility of real-time imaging (fluoroscopy).

## 4. REFERENCES

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