# Digital X-Ray Imaging

**David R. Dance**

**Abstract** The use of X-ray image receptors that produce a digital image is becoming increasingly important. Possible benefits include improved dynamic range and detective quantum efficiency, improved detectability for objects of low intrinsic contrast, and reduced radiation dose. The image can be available quickly. The display is separated from the image capture so that processing and contrast adjustment are possible before the image is viewed. The availability of a digital image means ready input into PACS and opens up the possibility of computer-aided detection and classification of abnormality. Possible drawbacks of digital systems include high cost, limited high contrast resolution and the fact that their clinical value is sometimes not proven in comparison with conventional, analogue techniques. The high contrast resolution attainable with such systems is discussed and the problem of sampling limitations and aliasing considered. The properties and limitations of digital systems using computed radiography, caesium iodide plus CCDs and active matrix arrays with either caesium iodide or selenium detectors are demonstrated. Examples are given of digital systems for mammography and general radiography and their performance is demonstrated in terms of clinical assessment and measurements of the modulation transfer function and detective quantum efficiency.

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## 1 Introduction

The use of digital detectors for medical X-ray imaging is now well-established. The digital capture of projection images has become the norm in many hospitals, where

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screen/film receptors and image intensifiers have been or are being replaced with digital receptors. Digital capture is essential if images are to be stored and accessed using a PACS.

This lecture provides a brief introduction to digital receptors, and gives examples of their performance taken from mammography and general radiography. Further details can be found in the list of references provided. The treatment is limited to consideration of the image receptor; the image display and the merits of soft-copy reporting of images are not considered here.

## 2 Benefits and Drawbacks of Digital Systems

Digital systems have a number of important advantages over conventional analogue imaging systems. Probably the most important are those connected with the limited dynamic range of film. This is illustrated in Figs. 6 and 7 of the companion lecture on mammography, which show a typical characteristic curve, film gamma and detective quantum efficiency (DQE) for a screen/film system. The contrast achievable in the screen/film image is proportional to the slope of the characteristic curve, or film gamma, which is only high over a limited range of optical densities. This means that it is critical to get the exposure right and even then parts of the image will show reduced contrast. Moreover, the detective quantum efficiency (DQE) of the system only has its highest value in the regions where the film gamma is high. In the high and low density regions of the image the DQE falls to low values. In addition, since the highest DQE is typically only 0.3 (at 1 line pair/mm), there is considerable potential for improvement and hence in dose reduction. Digital systems in general have a wide dynamic range and images can be captured over a wide range of exposures. These are important advantages, which, as well as avoiding the need for retakes (important for mobile radiography), means that the exposure can be reduced in some situations where a noisier image can be accepted. The rapid availability of the image can also be helpful. A good example is stereotactic localization, where films must be reviewed before the procedure can be completed.

Another important limitation of screen/film imaging is that the image cannot be manipulated before it is displayed. This of course is not true if the image is captured digitally, which opens up the further possibility of achieving a dose saving by using a higher energy X-ray spectrum and manipulating the image before it is displayed. In addition, the digital image is directly available for computer aided detection and classification of abnormalities.

A particular problem associated with digital systems is that it may be difficult to match the high contrast resolution achievable with screen/film systems because of the pixel sizes which are available and the limitation imposed by the Nyquist sampling theorem (see below). In practice, however, small objects do not present with very high contrast and image noise has an important effect on detectability. In such a situation it is the NEQ (noise equivalent quanta, or signal-to-noise ratio) which is important, which in turn is determined by the exposure used and the DQE. In this respect the digital system *may or may not* have a performance superior to that of the screen/film system.

# 3 Performance SPECIFICATION and Evaluation

The performance requirement for a digital radiology system can be specified in terms of parameters such as NEQ, DQE, MTF, patient dose and dynamic range. For evaluation purposes, these parameters for digital systems are often compared against those for analogue systems. However, some of the parameters can be difficult to measure and an alternative is to compare images of test phantoms and the detectability of details contained within the phantoms. Whilst this can be of great utility, it does not replace the need for clinical evaluation. Such evaluation should be based on the clinical objective of the examination, for example using ROC curves, and may be difficult to make with adequate statistics.

There is not space here for a detailed discussion of performance requirements, and we consider just three aspects, using mammography as an example.

## 3.1 Sampling

The Nyquist sampling theorem tells us that the sampling frequency must be at least twice the highest spatial frequency that is contained within the image. For mammography, we would like to be able to detect microcalcifications of around 100 μm in diameter, leading to a sampling interval of 50 μm, and a pixel size of similar magnitude. However, the satisfying of this constraint does not necessarily mean that such a small object will be visualized as it takes no account of the intrinsic resolution (or the pre-sampling MTF) of the image receptor or the noise. It should also be realised that information in the image above the Nyquist frequency will lead to aliasing of both the imaged information and the noise.[1]

## 3.2 Dynamic Range

For a typical mammographic X-ray spectrum, the attenuation through a 6 cm breast is about a factor 40. The contrast of a 3 mm carcinoma viewed against fatty tissue is about 4%, thus giving a noise level requirement of say 1%. To see 1% noise over a 40:1 transmission range gives a dynamic range requirement of 4,000:1 or 12 bits (for a detailed discussion of dynamic range requirements in digital mammography, see Maidment et al.[2]). For a receptor area of $18 \times 24$ cm$^2$, a pixel depth of 12 bits and a pixel size of 50 μm, the storage requirement for an uncompressed image is 25 Mbyte.
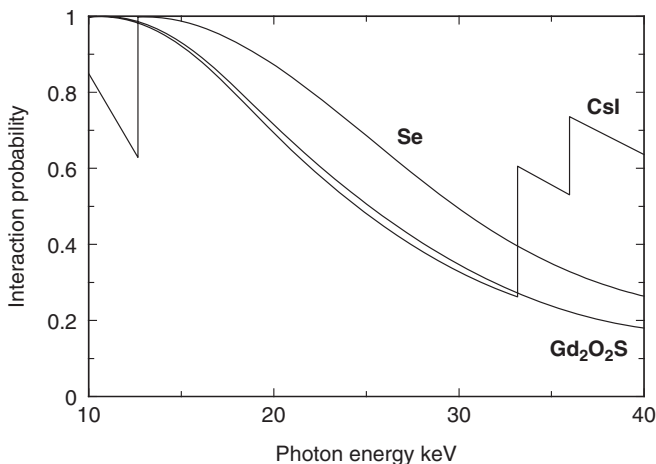
**Fig. 1** Variation of interaction probability with photon energy. Data are given for caesium iodide, gadolinium oxysulphide and selenium receptors 100 μm thick

## 3.3 Quantum Sensitivity and DQE

In order to achieve a high DQE, it is important that the X-ray photons incident on the image receptor have a high probability of interacting. Figure 1 shows a calculation of the interaction probability for three different receptor materials in an extended mammographic energy range. All three receptors are 100 μm thick, but the gadolinium oxysulphide receptor has a packing density of just 50%. It will be seen that in this energy range, the efficiencies of the cesium iodide and gadolinium oxysulphide are similar up to the region of the K-edges for the former. Selenium, however, has a K-edge just above 12 keV, and as a consequence has a high interaction probability for much of this energy range.

The interaction of the X-ray photon, however, is only the start of the story. The energy absorbed from the X-ray has to be converted into an electrical signal, which is then digitized. This whole process may involve a number of steps, each of which may increase the noise in the image and hence decrease the DQE.[3]

## 4 DIGITAL RECEPTORS using Photostimulable Phosphors

Computed radiography (CR) systems have been available since the 1980s and are widely used. They comprise a photostimulable phosphor, (usually barium fluorobromide with an $Eu^{2+}$ dopant deposited on a plastic substrate). When the phosphor is irradiated, energy absorbed from the incident X-rays produces electron-hole pairs. Many of these pairs recombine promptly, giving rise to the emission of fluorescent light. However, some electrons are trapped at colour or F-centres within the

**Fig. 2** Single-side read-out of a CR image plate (Figure based on IPEM (2005). Copyright Institute of Physics and Engineering in Medicine 2005. Reproduced with permission)
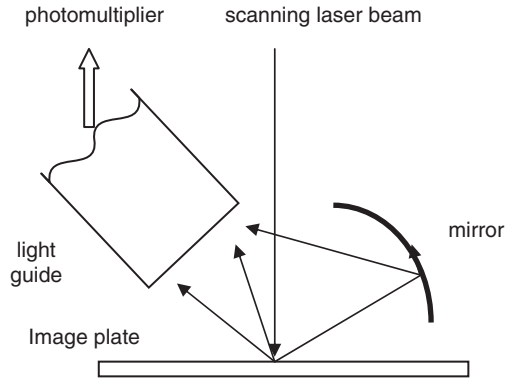


material, caused by the presence of the dopant. Such traps are metastable so that a latent image is built up consisting of electrons trapped at the F-centres. Once the exposure is complete, the plate is stimulated by irradiating with laser light of the right frequency. This frees the electrons from the traps and the energy released leads to the emission of light photons, which can be detected to create an image.

The CR image plate is built into a radiographic cassette, which is exposed in the same way as a screen/film cassette. For image read-out, the image plate is scanned using a fine laser beam which traverses the plate in a in a raster pattern. The light emitted is collected by a light guide and detected by a photomultiplier (Fig. 2).

The resolution of the CR system is limited by the spread of the read-out light within the image plate and the spot size of the laser beam. Some improvement of the resolution can be achieved by reading the plate out from both sides, and systems are available for mammography which use this approach.

The dynamic range of a CR system can be in excess of $10^4$, which is an important advantage compared with screen/film systems, but, depending upon the particular systems chosen and the operating conditions, the resolution and DQE may not be as good. A very full account of CR systems, their limitations and potential is given in Rowlands.[4]

## 5 Direct Digital Systems using Phosphors

The images produced by the CR systems described above are only available after the exposed digital cassette is taken to the digital processor for read-out. For some digital systems, however, the electrical signal generated following the interaction of the X-rays can be read-out in a short period immediately following the exposure, and the image is thus available for viewing straight away. Such systems are sometimes called direct digital systems, and may use a light emitting phosphor or a photoconductor to absorb the incident X-rays.

## 5.1 Direct Digital Systems using Phosphors

Direct digital systems using phosphors rely on the detection of the light fluorescent photons emitted by the phosphor. Possible phosphors include caesium iodide and gadolinium oxysulphide. Cesium iodide will give a better resolution at the same phosphor thickness because its 'cracked' structure of parallel crystals channels the fluorescent photons emitted in the phosphor towards the light photon detector. This reduces the lateral spread which is possible with a fluorescent screen such as gadolinium oxysulphide. On the other hand, the number of light photons produced per X-ray absorbed in gadolinium oxysulphide is greater than that for cesium iodide as they require on average 13 and 19 eV[5] respectively per light photon emitted.

### 5.1.1 Systems using Charge Coupled Devices

One option for detecting the light fluorescent photons emitted by the phosphor is to use a charge-coupled device or CCD. This comprises a series of electrodes or gates on a semi-conductor substrate. An array of metal-on-semiconductor capacitors is formed, which act as storage wells for the charge generated within the CCD by the photoelectric absorption of optical quanta. The 'charge image' is built up in lines of capacitors and is read out by passing the charge from capacitor to capacitor along each line. A high charge transfer efficiency is required.

Because the size of CCDs is limited, it is necessary to employ some means of demagnification as the light photons pass from phosphor to CCD. This can be achieved for example by use of a lens or a fibre optic taper. The light loss associated with coupling using a lens is much greater than that for coupling with a fibre optic taper at the same demagnification. The DQE of systems using the latter approach can therefore be expected to be better. Figure 3 shows the light collection efficiencies for the two approaches. Even with demagnification, an array of CCDs (e.g. a $3 \times 4$ array) may be used to achieve a large enough image matrix. An alternative approach is to use a line or slit CCD array which is scanned across the image whilst the image is read-out in 'time delay integration mode'.[7]

### 5.1.2 Systems using Active Matrix Arrays

Active matrix LCDs constructed from amorphous silicon are widely used in notebook computer displays, but active matrix devices can also be configured as an alternative to the CCD for the detection of light emitted by a phosphor.[5] A layer of cesium iodide can be evaporated directly onto the active matrix. Each pixel is configured as a photodiode, which converts the light fluorescent photons to electrical charge. For each pixel there is an associated region of the device which is configured as a thin-film field-effect transistor (TFT) and is used for image read-out. As the area of the device accommodates both the photodiode and read-out control circuitry, there is some loss of efficiency, usually expressed as the 'fill-factor'. For
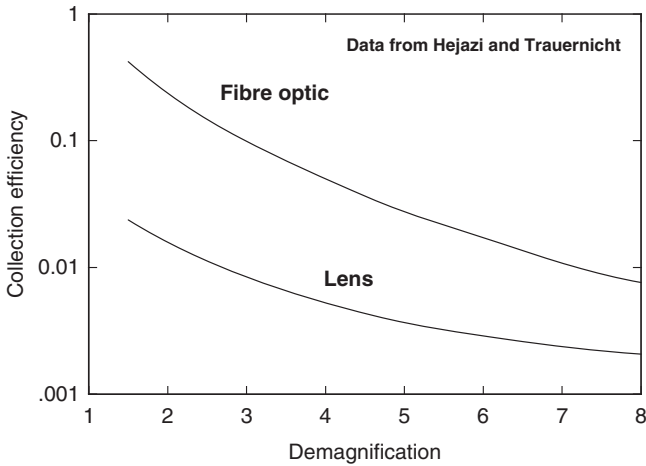
**Fig. 3** Light collection efficiency for fibreoptic and lens systems for coupling to the CCD (Data taken from Hejazi and Trauernicht[6])

example, a particular digital system using an amorphous silicon active matrix read-out has a sampling interval of $100\,\mu m$, but the pixel size is only $87\,\mu m$, leading to a 'fill factor' of 0.75, and a reduction of quantum sensitivity and DQE by this factor. This problem becomes more important as the size of the pixel decreases.

## 5.2 Direct Digital Systems using Selenium Photoconductors

It can be deduced from Fig. 1 that amorphous selenium can be deposited in thick enough layers to have a reasonably high X-ray absorption efficiency for moderate photon energies. This and its property of photoconduction makes it attractive as part of an imaging device that can directly produce electrical charge following the interaction of an X-ray photon (and without the need to produce fluorescent light photons). The efficiency with which charge is produced depends upon the electric field strength in the selenium, but a typical value is 50 eV per electron/hole pair. Because the movement of the electrical charge within the selenium is along the direction of the electric field, there is very little lateral spread of image information as the charge is moved from its production point to its measurement point. There is thus potential for excellent spatial resolution.

For read-out purposes, the amorphous selenium can be evaporated onto an silicon active matrix in which the pixel elements are electrodes and the read-out can again be controlled via TFTs.

## 6 Performance of Digital Systems

As noted in section 3, the performance of digital and conventional X-ray imaging systems may be compared in many ways. We give two examples: a clinical comparison using cancer detection rates and a physical comparison using DQE and MTF.

The use of screen/film systems for screening mammography is well established, but following the introduction of digital mammographic systems, there has been strong need to compare the performance of both modalities in the screening situation. The low incidence of breast cancer necessitated a very large clinical trial. Pisano et al.[8] looked at breast cancer detection for 42,760 paired screening examinations, each woman receiving both digital mammographic (DM) and screen-film (SFM) exposures. Cancer detection was measured in terms of the area under the ROC curve. Results are listed below for three groupings: all women; women under 50 years; and women with radiographically dense breasts.

| | | |
|---|---|---|
| All: | DM $0.78 \pm 0.02$; | SFM $0.74 \pm 0.02$ |
| <50years: | DM $0.84 \pm 0.02$; | SFM $0.69 \pm 0.02$ |
| Dense: | DM $0.78 \pm 0.04$; | SFM $0.68 \pm 0.03$ |

The differences observed between DM and SFM for women aged less then 50 years and for dense breasts are both statistically significant at the 95% confidence interval.

Samei and Flynn[9,10] have compared different digital receptors in terms of MTF and DQE. Some of their results are shown in Figs. 4 and 5 for: DR1000, a system using a $500\,\mu m$ thick selenium photoconductor and a pixel size of $139\,\mu m$; X/Qi and
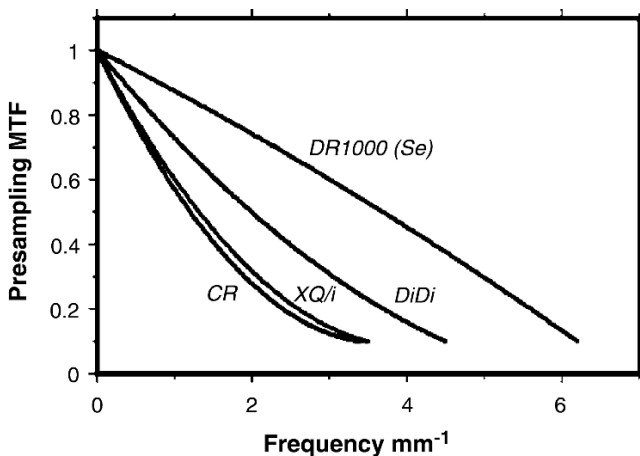


**Fig. 4** Digital (pre-sampling) MTF for particular CR, caesium iodide flat panel (XQ/i and DiDi), and selenium flat panel systems (Data taken from Samei and Flynn[9,10])
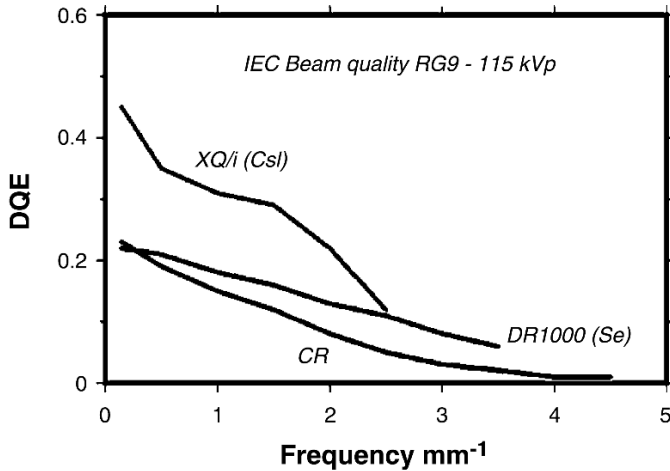
**Fig. 5** DQE for particular CR, caesium iodide flat panel (XQ/i), and selenium flat panel systems (Data taken from Samei and Flynn [10])

DiDi, two systems using a CsI phosphor and a 200 μm pixel size; and a CR system with a pixel size of 100 μm. The MTF shown in Fig. 4 for one of the CsI-based systems has been enhanced artificially by the manufacturer, but in spite of this the selenium-based system has the best MTF.

This advantage, however, does not carry over to the DQE when measured at IEC beam quality RG9 at 115 kVp, where the XQ/i CsI system performs best up to the Nyquist frequency of 2.5 cycles per mm. This result is to be expected from the X-ray absorption properties shown in Fig. 1. For this particular example CR has both the worst MTF and DQE.

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