Chapter 7: Image Receptors

Slide set of 220 slides based on the chapter authored by John A. Rowlands and Ulrich Neitzel of the IAEA publication (ISBN 978-92-0-131010-1):

Diagnostic Radiology Physics: A Handbook for Teachers and Students

Objective:

To familiarize the student with image receptors used in X-ray imaging systems.

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7.1 INTRODUCTION

X ray images are formed as **Shadows** of the interior of the body

Since it is not yet practical to focus X rays, an X ray receptor has to be **Larger** than the body part to be imaged

Thus the **First** challenge in making an X ray receptor is the need to image a large area

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7.1 INTRODUCTION

A **Second** challenge is to make a system which has image quality as good as allowed by the physics

i.e. permits the detection of objects whose size and contrast is limited only by the **Quantum Statistics**

This means absorbing most of the X ray quanta and using these in an **Efficient**, i.e. a quantum noise limited, manner

while simultaneously providing adequate spatial resolution

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7.1 INTRODUCTION

The **Capture** of an X ray image may conceptually be divided into three stages (with a possible fourth stage):

- The **Interaction** of the X ray with a suitable detection medium to generate a measurable response
- The temporary **Storage** of this response with a recording device
- The **Measurement** of this stored response
- Setting the system ready to start **again**

Example 1

The stages for a **Screen-Film** system are:

1. The **Interaction** of an X ray in a phosphor material followed by generation of visible light photons
2. The creation of a **Latent Image** in the photographic film by these photons
3. The **Development** of a fixed photographic image
4. For re-usable systems (i.e. those not requiring consumables such as film) is the **Erasure** of all previous images within the detection system in order to prepare for a fresh image

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Example 2

For a **Digital Direct** conversion flat panel imaging system:

1. The **Absorption** of an X ray followed by the release of multiple secondary electrons in a photoconductor
2. The drifting of the electrons and holes to individual electrodes where they are **Stored**
3. The **Readout** phase - the charges are transferred to amplifiers where they are digitized line by line
4. This is achieved in step 3 - the readout simultaneously and without further effort performs the essential **Erasure**

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7.1 INTRODUCTION

Breaking up the **stages** in this manner is helpful to the understanding of the physics of image acquisition

which itself is key to the:

- **Optimization** of the receptor design and the
- **Understanding** of fundamental limitations on image quality

It is also key to developing an understanding of the **complementary** strengths of the various approaches used in the past, the present and in the future

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7.2 GENERAL PROPERTIES OF RECEPTORS

Before describing in more detail the properties of the different types of image receptor used for projection radiography

it is necessary to consider the various:

- Physical Properties and
- Quantities

which are used to **specify** their performance

7.2 GENERAL PROPERTIES OF RECEPTORS
7.2.1 Receptor Sensitivity

The initial image **acquisition** operation is identical in all X ray receptors

To produce a signal, the X ray quanta must **interact** with the receptor material

The probability of interaction, or **Quantum Detection Efficiency** for an X ray of energy **E** is given by:

$$A_Q = 1 - \exp [- \mu(E, Z)\, T]$$

where μ is the linear attenuation coefficient of the receptor material, Z is the material's atomic number, and T its thickness

7.2 GENERAL PROPERTIES OF RECEPTORS

7.2.1 Receptor Sensitivity

Because virtually all X ray sources for radiography emit X rays over a spectrum of energies, the **Quantum Detection Efficiency** must either be specified as a function of energy or as an effective value over the spectrum of X rays **incident** on the receptor

A_Q will in general be highest at low E decreasing with increasing E

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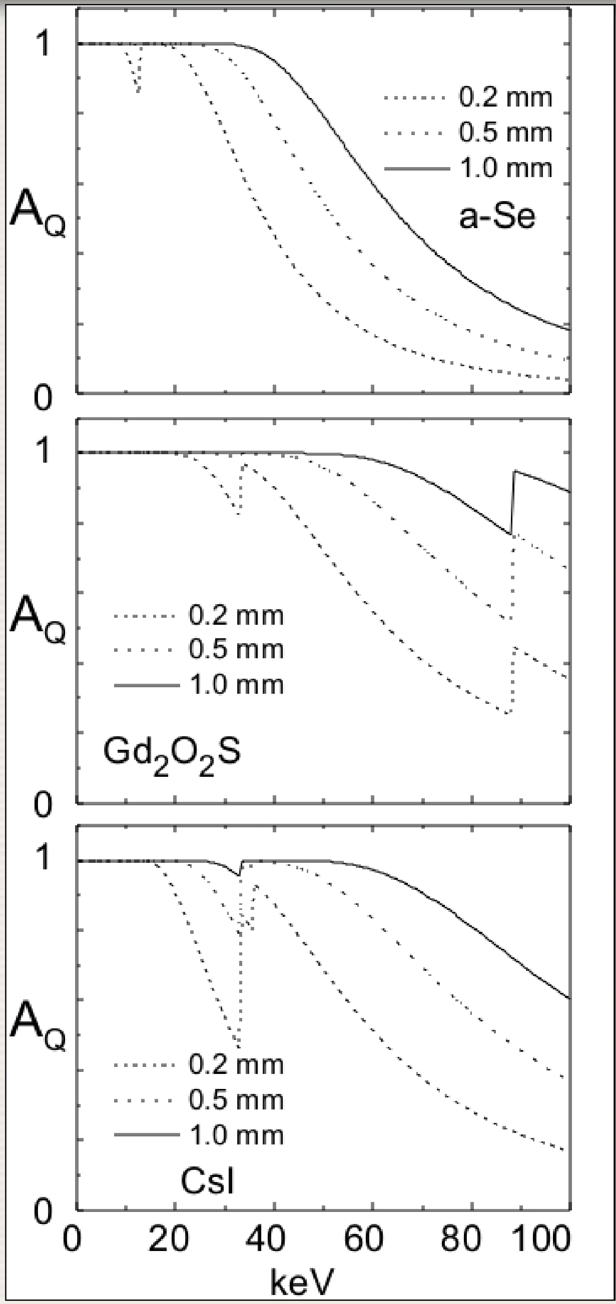
7.2 GENERAL PROPERTIES OF RECEPTORS

7.2.1 Receptor Sensitivity

A_Q for representative examples of:

- an X ray photoconduct or a-Se
- a screen phosphor Gd_2O_2S
- the scintillator CsI



The curves are for the primary interaction using the photoelectric coefficient **only**

The thicknesses are for assumed 100% packing fraction of the material which:

- is realistic for a-Se
- should be increased by ~2x for a powder screen such as Gd_2O_2S, and
- by ~1.1-1.2 x for an evaporated structured CsI layer

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At diagnostic X ray energies, the main interaction process is the **Photoelectric Effect** because of the relatively high **Z** of most receptor materials

If the material has a K atomic absorption edge $\mathbf{E_K}$ in the energy region of interest, then

$$\mathbf{A_Q} \text{ increases dramatically at } E_K$$

causing a local minimum in $\mathbf{A_Q}$ for $E < E_K$

7.2.1 Receptor Sensitivity

The photoelectric interaction of an X ray quantum with the receptor generates a high-speed **Photoelectron**

During the subsequent loss of kinetic energy of the electron in the receptor, **Excitation** and **Ionization** occur, producing the secondary signal (optical quanta or electronic charge)

The sensitivity of any imaging system therefore depends both on $\mathbf{A_Q}$ and the **primary Conversion Efficiency**

the **efficiency** of converting the energy of the interacting X ray to a more easily measurable form such as optical quanta or electrical charge

7.2 GENERAL PROPERTIES OF RECEPTORS

7.2.1 Receptor Sensitivity

Conversion Efficiency can be re-expressed as the **Conversion Factor**

i.e. in terms of the number of secondary particles (light photons in a phosphor or electron–hole pairs, **EHPs**, in a photoconductor) released per X ray

For a surprising number of materials and systems this is **~1000** quanta or EHPs per 50 keV X ray

The Conversion Factor is closely related to the intrinsic **Band Structure** of the solid from which the receptor is made

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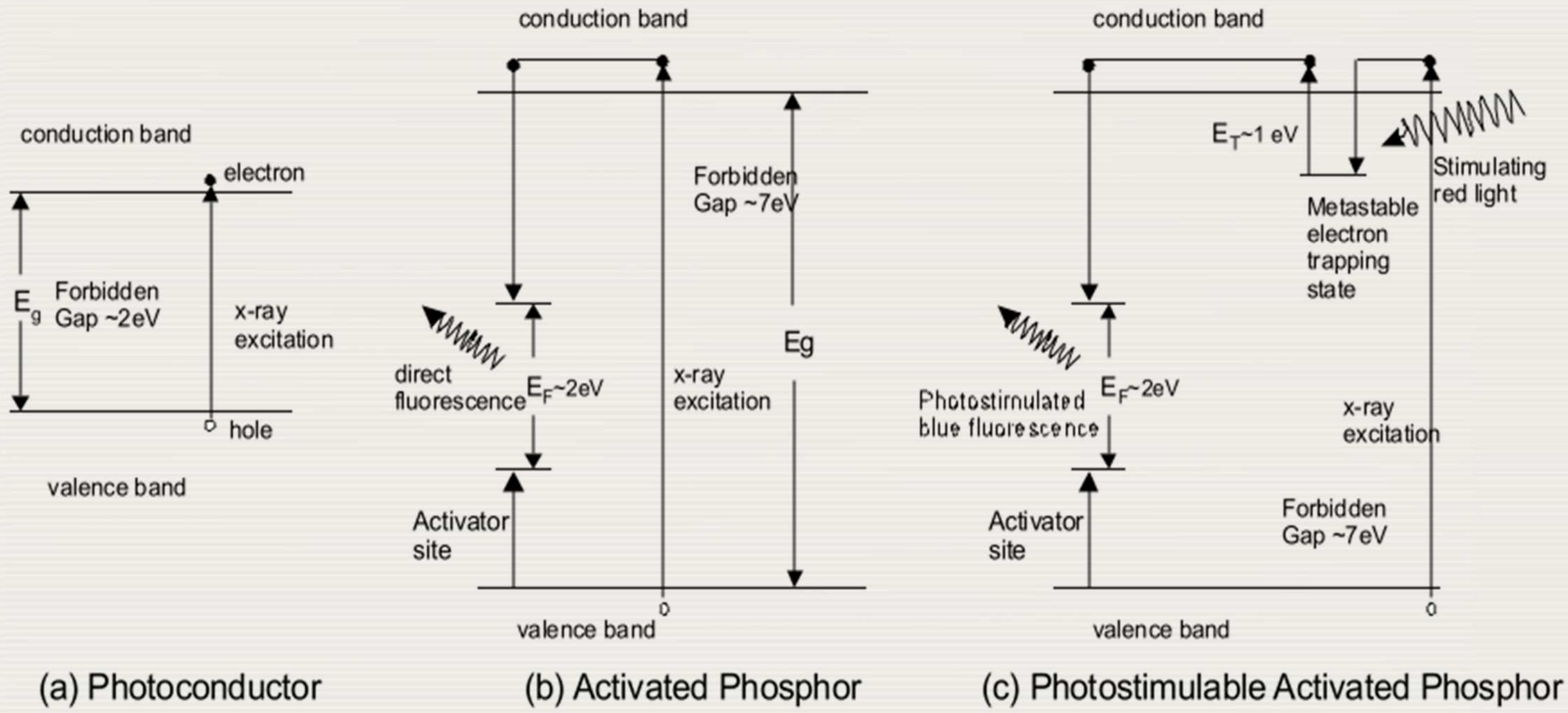
7.2 GENERAL PROPERTIES OF RECEPTORS
7.2.1 Receptor Sensitivity

Band Structure



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In all receptor materials the **Valence Band** is almost fully populated with electrons and the **Conduction Band** is practically empty

The **Forbidden Energy Gap**, E_g, governs the energy scale necessary to release a mobile EHP

i.e., to promote an electron from the valence band to the conduction band

Although $\mathbf{E_g}$ is the minimum permitted by the principle of conservation of energy, this can be accomplished only for photons of energy E_g

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For **Charged Particles** releasing energy

e.g. through the slowing down of high-energy electrons created by an initial X ray interaction

conservation of both **Energy** and **Crystal Momentum** and the presence of competing energy loss processes necessitate ~3E_g to release an EHP

For a photoconductor the maximum number of EHP is 50,000/(2 eV x 3)

~8,000 EHP

This is possible for good **Photoconductors**

but for the only practical photoconductor used in commercial systems at this time (**a-Se**) there are other losses

primarily to **Geminate Recombination**

i.e. the created EHPs recombine before separation by the applied electric field

which limit it to **~1,000-3,000 EHP** depending on the applied electric field

In **Phosphors**, the band gap is usually much higher (~8 eV) so the intrinsic conversion factor is typically lower

only **~2,000 EHPs** are released

50,000/8 eV x 3

which however in an **Activated Phosphor** results in emission of only slightly less (1,800) light photons

7.2.1 Receptor Sensitivity

Additional optical losses due to:

- **Light Absorption** in the phosphor layer (sometimes with an intentionally included **dye**) and/or **non-reflective backing**
- Dead space between the photoreceptors (**Fill Factor**)
- Non-ideal **Quantum Efficiency** of the photoreceptors

further reduces the light per X ray

This typically results in **~1,000 EHPs** collected in the photoreceptor for our prototypical 50 keV X ray

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All images generated by quanta are **Statistical** in nature

i.e., although the image pattern can be predicted from the attenuation properties of the patient, it will fluctuate **randomly** about the mean predicted value

The fluctuation of the X ray intensity follows **Poisson statistics**

so that the **variance**, σ^2, about the mean number of X ray quanta, N_0, falling on a receptor element of a given area, is equal to N_0

Interaction with the receptor can be represented as a **Binomial** process with **probability** of success, A_Q, and the distribution of interacting quanta is still Poisson with variance:

$$\sigma^2 = N_0 A_Q$$

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7.2.2 Receptor Noise

If the detection stage is followed by a process that provides a mean gain **g** then the distribution will **not** be Poisson even if **g** is Poisson distributed

It is also possible that **other** independent sources of noise will contribute at different stages of the imaging system

Their effect on the variance will be **Additive**

A complete linear analysis of signal and noise propagation in a receptor system must also take into account the **Spatial Frequency Dependence** of both signal and noise

It is important that the number of secondary quanta or electrons at each stage of the image production be considerably greater than N_0, to avoid having the receptor noise dominated by a **Secondary Quantum Sink**

Consideration of the propagation of noise is greatly facilitated by the consideration of a

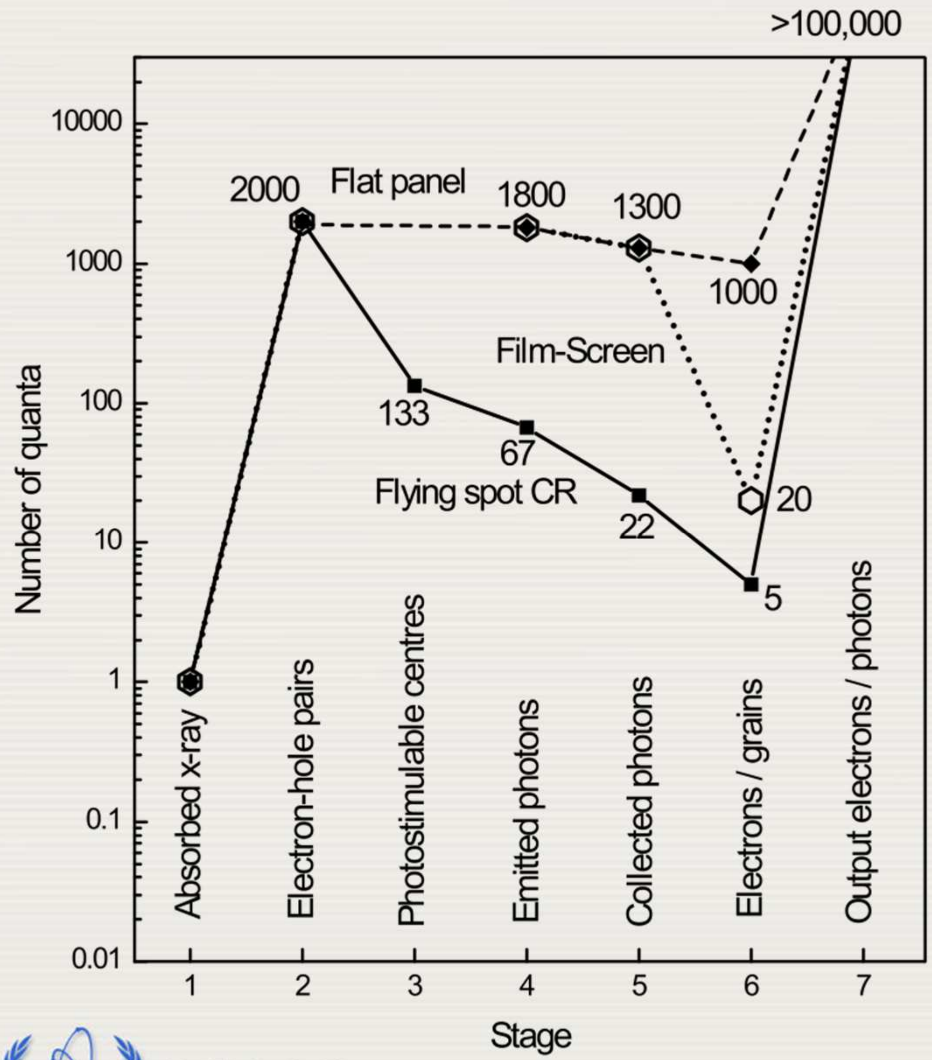
Quantum Accounting Diagram

The vertical axis represents the average number of quanta or individual particles (electrons or film grains) representing the initial absorbed X ray (assumed to be of 50 keV) at each stage in the imaging system

7.2 GENERAL PROPERTIES OF RECEPTORS

7.2.2 Receptor Noise

Quantum Accounting Diagrams for screen-film, CR and flat panel DR



The critical point for each modality is where the minimum number of quanta or EHPs represents a single X ray

For flat panel systems this is ~1,000 while for screen-film it is 20 and only 5 for CR

This is the **Weakest Link**

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The concept is that the **Noise** from each stage of the imaging system is related to the number of secondary quanta or electrons at each stage

So ideally there should for all stages be **many more** (if possible exceeding 1,000) such secondary quanta or particles representing each primary quantum (i.e. X ray)

The point at which this number is lowest is the

Secondary Quantum Sink

7.2.2 Receptor Noise

With the examples given (all of which are used commercially) have at least **5** secondaries per primary

but it is very easy to find systems in which this is **not** the case, such as:

- Non-intensified fluoroscopy or
- Some optically (lens) coupled radiographic X ray systems

7.2 GENERAL PROPERTIES OF RECEPTORS

7.2.2 Receptor Noise

The noise in X ray images is related to the **Number of X Rays per Pixel** in the image and hence to the X ray exposure to the receptor

However, the relative noise can be increased by lack of absorption of the X rays, as well as by fluctuations in the response of the receptor to those X rays which are absorbed

There are also unavoidable fluctuations in the signal produced in the detection medium even when X rays of identical energy interact and produce a response

These are caused by the statistical nature of the **competing** mechanisms that occur as the X ray deposits energy in the medium

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Together they give rise to a category of noise known as gain-fluctuation or **Swank Noise**

The gain-fluctuation noise can be determined experimentally using the pulse height spectrum, or **PHS**

From this the **Swank Factor**, A_S, is obtained as a combination of the zeroth, first and second moments (M_N) of the PHS using the formula:

$$A_s = M_1^2 / (M_0 M_2)$$

The **ideal** PHS, obtained when all absorbed X rays give rise to equal amounts of signal

results in a **Delta Function**

and

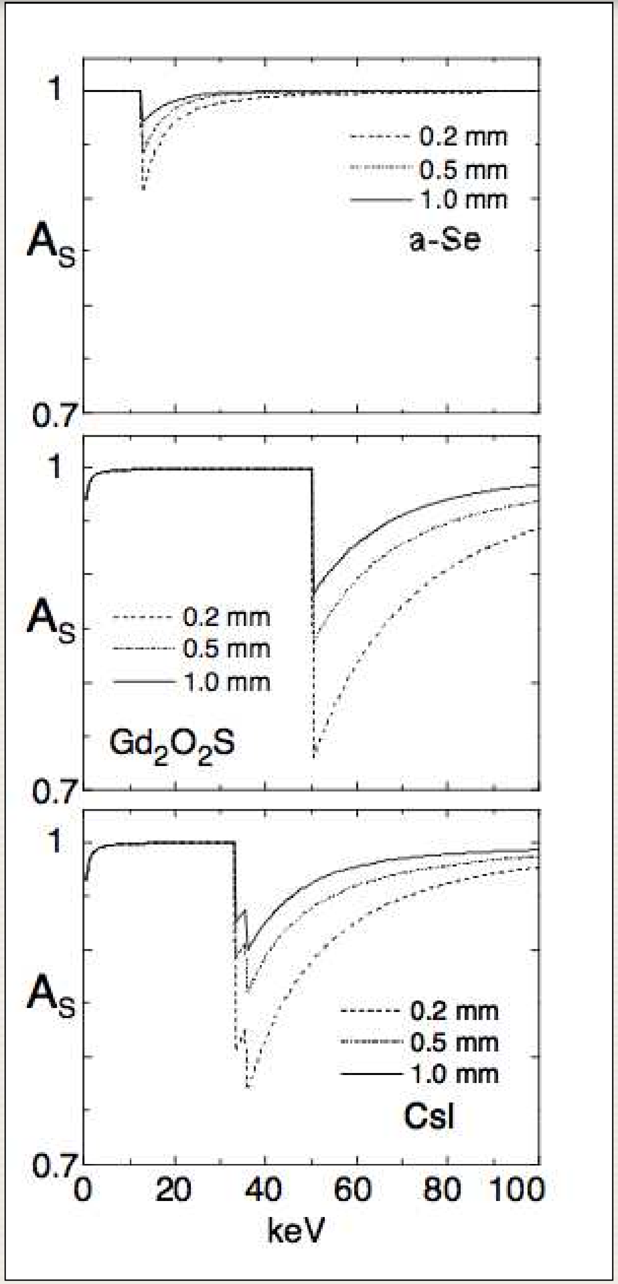
a Swank factor of **unity**

However in practice there are a number of effects which may **broaden** this spectrum, resulting in a Swank factor of less than unity

Swank Factor

A_S for representative examples of:

- an X ray photoconduct or a-Se
- a screen phosphor Gd_2O_2S
- the scintillator CsI



These are calculated values and are for the Photoelectric Effect **only**

which effectively means that only **K-escape** is accounted for

K-Escape is the emission of a K-fluorescent X ray following a photoelectric interaction, which then **escapes** from the receptor without depositing further energy

As the energy of the K-fluorescent X ray is below the K-edge, it has a smaller interaction **probability** than the original incident X ray photon

The range of values of A_s is from **0.7-1**

Further losses due to optical effects will be seen in screens, and other losses in photoconductors due to trapping of charge

Swank demonstrated that for many situations the combination of factors can be performed simply by **multiplying** the component factors

For example the K-escape Swank factor can be multiplied by the optical Swank factor to obtain the overall Swank factor

Theoretically, for an exponential PHS which can occur for screens with very high optical absorption the optical value can be as poor as **0.5**, resulting in a range 0.5-1 for the optical effects

Overall therefore the possible range of receptor Swank factors for screens is **0.35-1**

The noise due to both **Quantum Absorption** and **Gain Fluctuations** can be combined to create the zero spatial frequency **Detective Quantum Efficiency** that is given by:

$$DQE(0) = A_Q A_S$$

Due to its importance it is worth reiterating that, the DQE(0) is the effective quantum efficiency obtained when we **compare** the noise in the measured image to what it would be if it was an **ideal** (perfect) absorber and there was **no** gain fluctuation noise

7.2.3 Grayscale Response & Dynamic Range

The **Grayscale Response** used for an imaging system has to do with

- the physics of **X Ray Detection**
- the **Imaging Task** to be performed and
- the response of the human **Eye-Brain System** to optical images (the most difficult part)

In practice many of the decisions made by system designers are **empirical** rather than fully from theoretical analysis

However, some **Rules of Thumb** can be helpful

Regarding human vision, there is a spatial frequency **range** in which the human eye is most acute

This is an intermediate frequency range, neither too low nor too high

Regarding intensity, it is, as for all human senses, essentially **logarithmic** in its response, i.e., it is fairly good at seeing fractional differences, provided these are directly juxtaposed

Otherwise the human eye is quite poor at quantitative evaluations of intensity

In order to separate the **Subjective** eye-brain response from the more quantitative issues it is usual to:

- **Leave Out** (in the case of inherently non-linear systems like film) or to
- **Correct**

for the optical display part of the system which has a non-linear response (e.g. CRT monitors or LCD flat panel displays)

Only then can most systems be, for practical purposes, modelled as being **Linear**

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The grayscale response is usually expressed as the

Characteristic Curve

a plot of the response of the system to a stimulus

For example

in Fluoroscopy this would be the optical intensity at the video monitor plotted as a function of the irradiation of the sensor at the corresponding point in the image

The range of intensities that can be represented by an imaging system is called the **Dynamic Range** and it depends on the pixel size that is used in a manner that depends on the MTF

However, for any pixel size the dynamic range for an X ray imaging task can be broken into **two** components:

- Describes the ratio between the X ray attenuation of the most **radiolucent** and the **most radio-opaque** paths through the patient appearing on the same image
- The required **Precision** of the X ray signal measured in the part of the image representing the most radio-opaque anatomy

If, **for example**

there is a factor of **10** in attenuation across the image field

and

it is desired to have **10%** precision in measuring the signal in the most attenuating region

then the **Dynamic Range** requirement for the receptor would be

100

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The Dynamic Range that can be **achieved** by a practical linear imaging system can be defined in terms of the response at the output referred back to the input in terms of the X ray exposure:

$$\textbf{Dynamic Range} = X_{max} / X_{noise}$$

where X_{max} is the X ray exposure providing the maximum signal that the receptor can respond to before **saturation**

i.e. that point where the receptor output ceases to respond sensibly to further input

X_{noise} is the rms receptor noise in the **dark**

i.e. no X ray exposure

X rays are attenuated **exponentially**:

- thus an extra tenth-value layer thickness of tissue will attenuate the beam by **10**
- while a lack of the same tenth-value thickness will increase the X ray exposure by **10**

Thus when a mean exposure value $\mathbf{X_{mean}}$ for the system is established by irradiating a uniform phantom, we are interested in multiplicative factors above and below this mean value

i.e. $\mathbf{X_{mean}}$ is a **Geometric** rather than **Arithmetic** mean

For a DR of **100**:

the correct range is that given by the

Geometric Mean of

$10X_{mean}$ and $0.1X_{mean}$

not that given by the

Arithmetic Mean which would be

$\sim 2X_{mean}$ and $0.02X_{mean}$

7.2.4 Receptor Blur

Spatial Resolution in radiography is determined both by the receptor characteristics and by factors unrelated to the receptor

The latter includes **Unsharpness** (blurring) arising from geometrical factors such as:

(1) **Penumbra** (partial X ray shadow) due to the effective size of the X ray source and the magnification between the anatomical structure of interest and the plane of the image receptor and

(2) **Motion Blurring** due to relative motion of the patient with respect to the receptor and X ray focal spot

In the overall design of an imaging system, it is important that these other physical sources of unsharpness be considered when the

- **Aperture Size** and
- **Sampling Interval**

are chosen

If, **for example**, the MTF is limited by unsharpness due to the focal spot, it would be of little value to attempt to improve the system by designing the receptor with much smaller receptor elements

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7.2 GENERAL PROPERTIES OF RECEPTORS

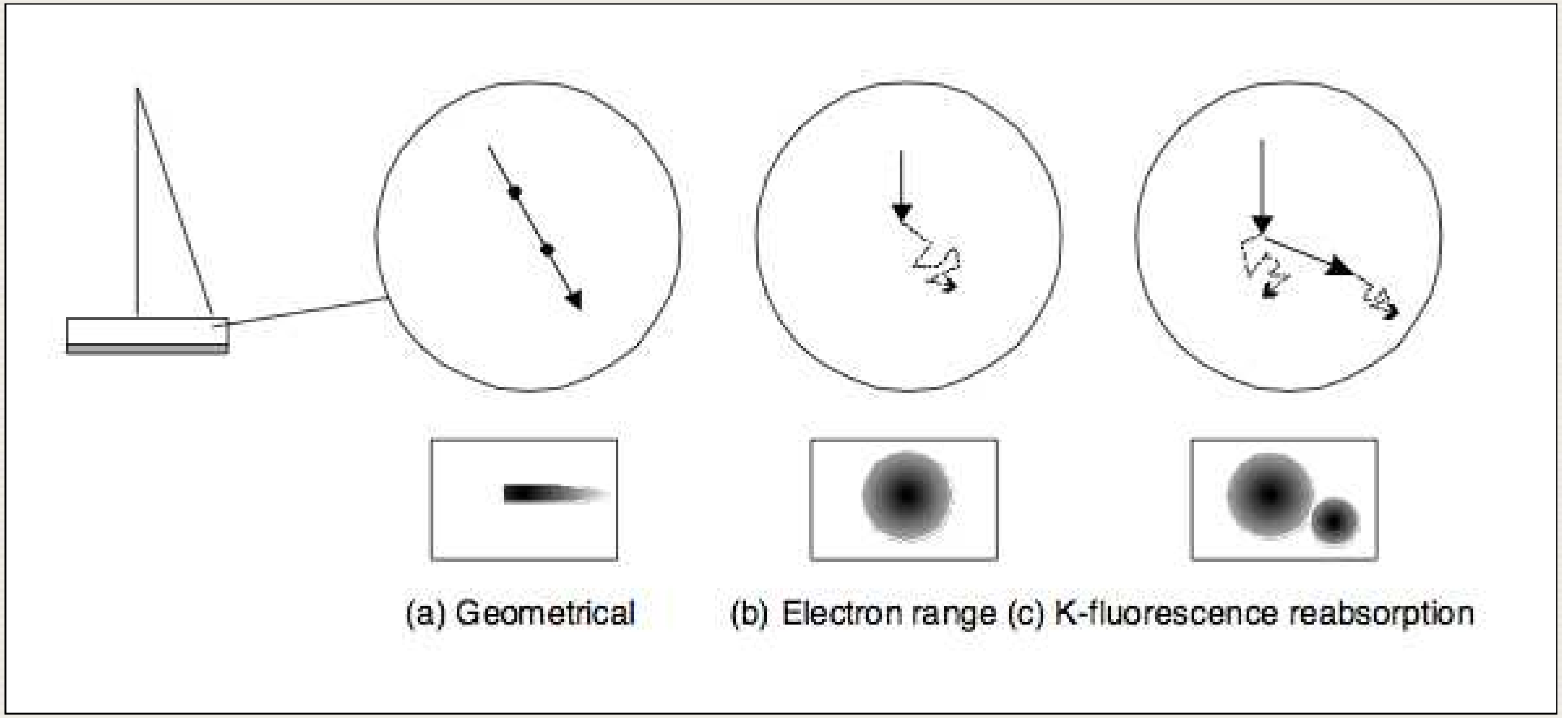
7.2.4 Receptor Blur

The major issues related to receptor blur include the fundamental issues arising within any material which are:

(a) **Geometrical Blurring**

(b) the **Range** of the primary electron

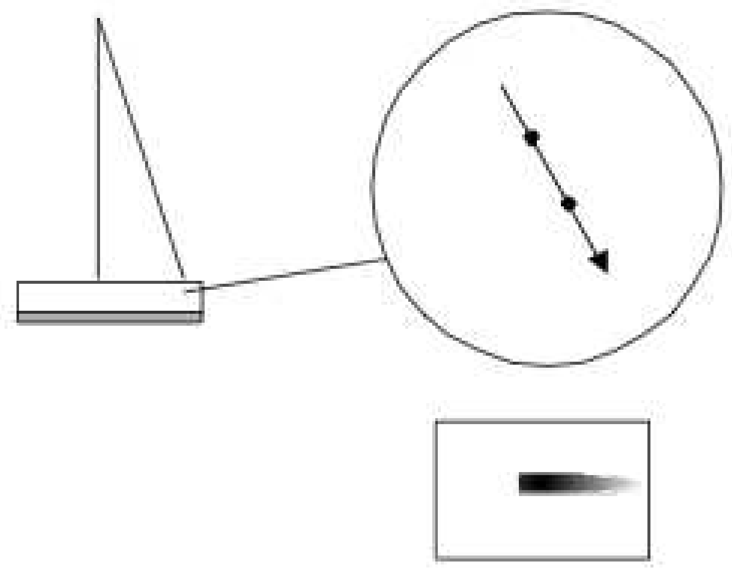
(c) the **Re-Absorption** of K fluorescence



Geometrical Blurring

due to oblique incidence of X rays which is especially marked far from the **Central Ray**

i.e. the X ray which strikes the receptor normally



(a) Geometrical

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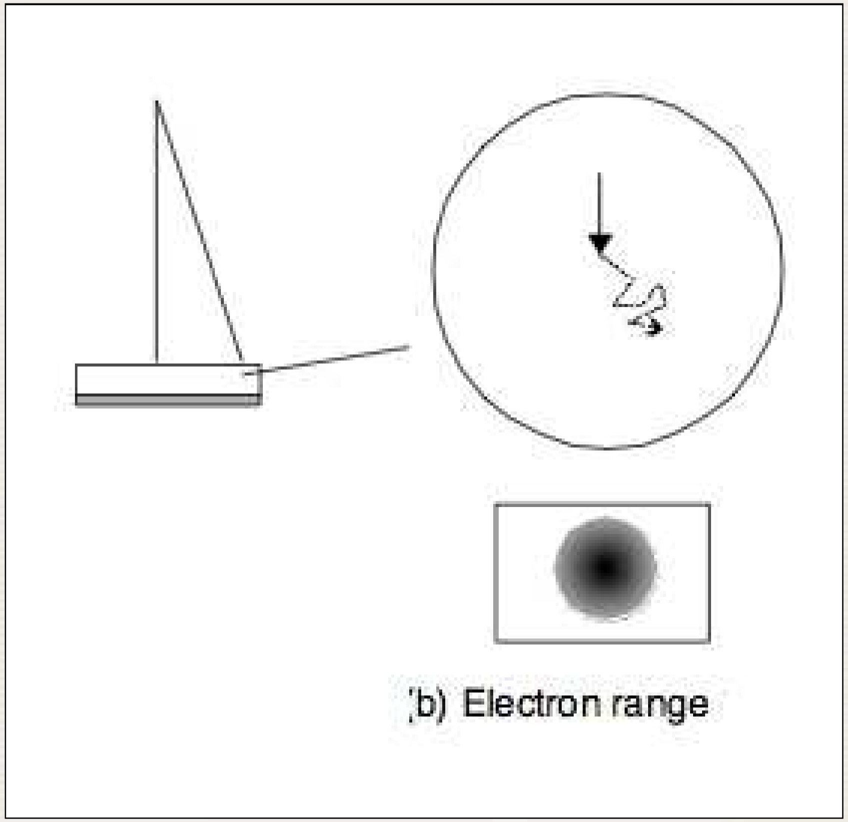
Range of the Primary Electron



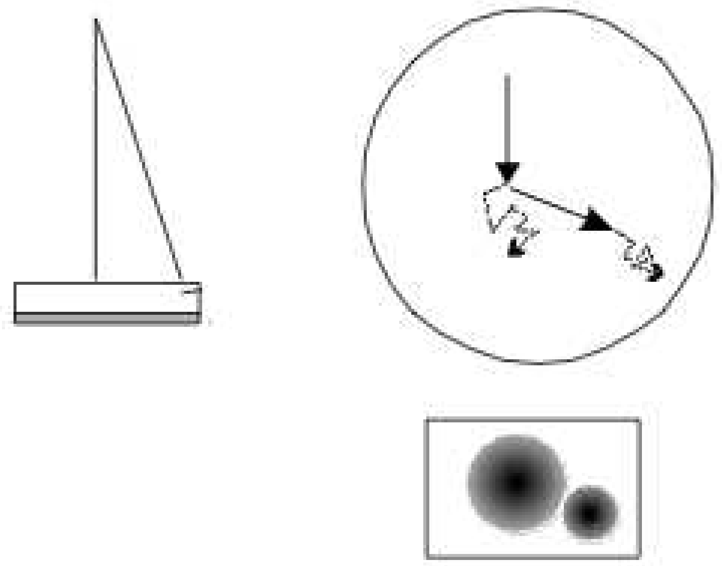
the primary electron usually gives up its energy in small amounts, ~100-200 eV at a time

but this is sufficient to scatter the electron at any angle, thus the path of the primary is usually a **random walk** and it does not go so far from its initial point of interaction

with a separation of **~1 µm** for 10 keV electrons and **~50-100 µm** for 100 keV depending on the medium in which it is absorbed

Re-Absorption of K Fluorescence X Rays

some distance from the primary photoelectric interaction, something which is likely because of the general rule that a material is relatively transparent to its own K-fluorescence due to the minimum in attenuation below the K-edge



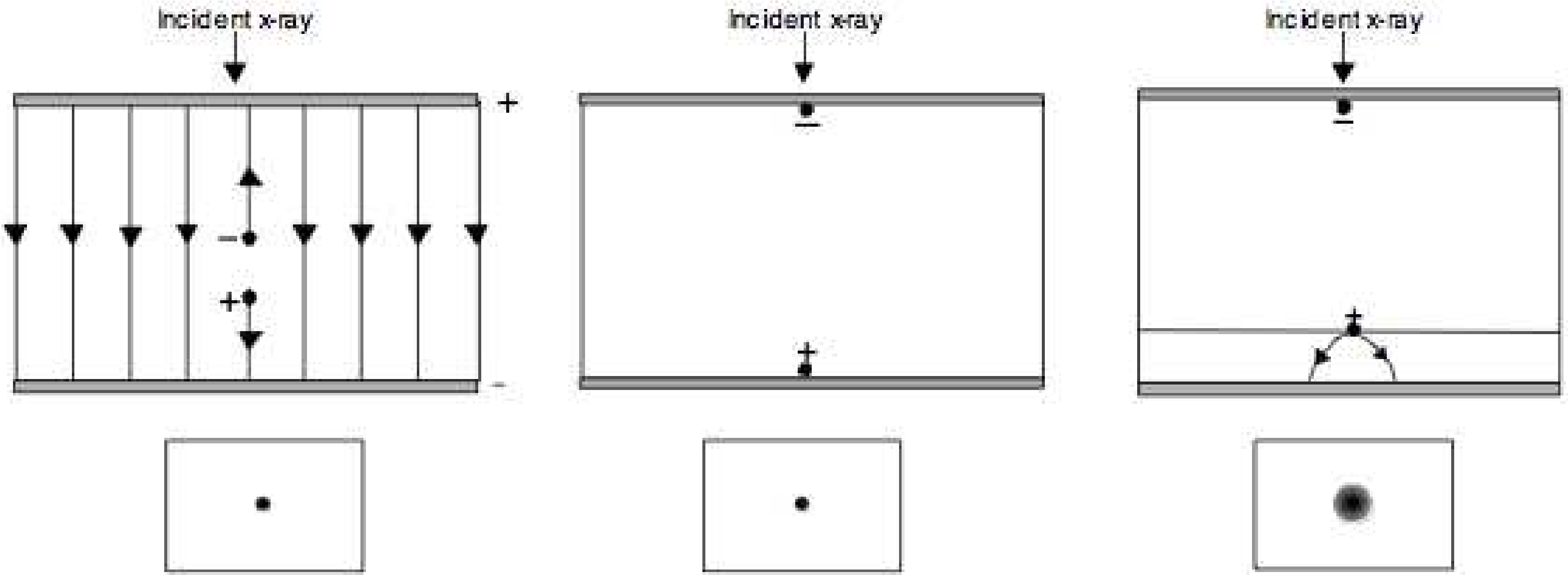
(c) K-fluorescence reabsorption

In addition there are material dependent effects specific to direct conversion and indirect conversion receptors shown in the following slides

Also to be considered in **Digital Systems** are the effects of:

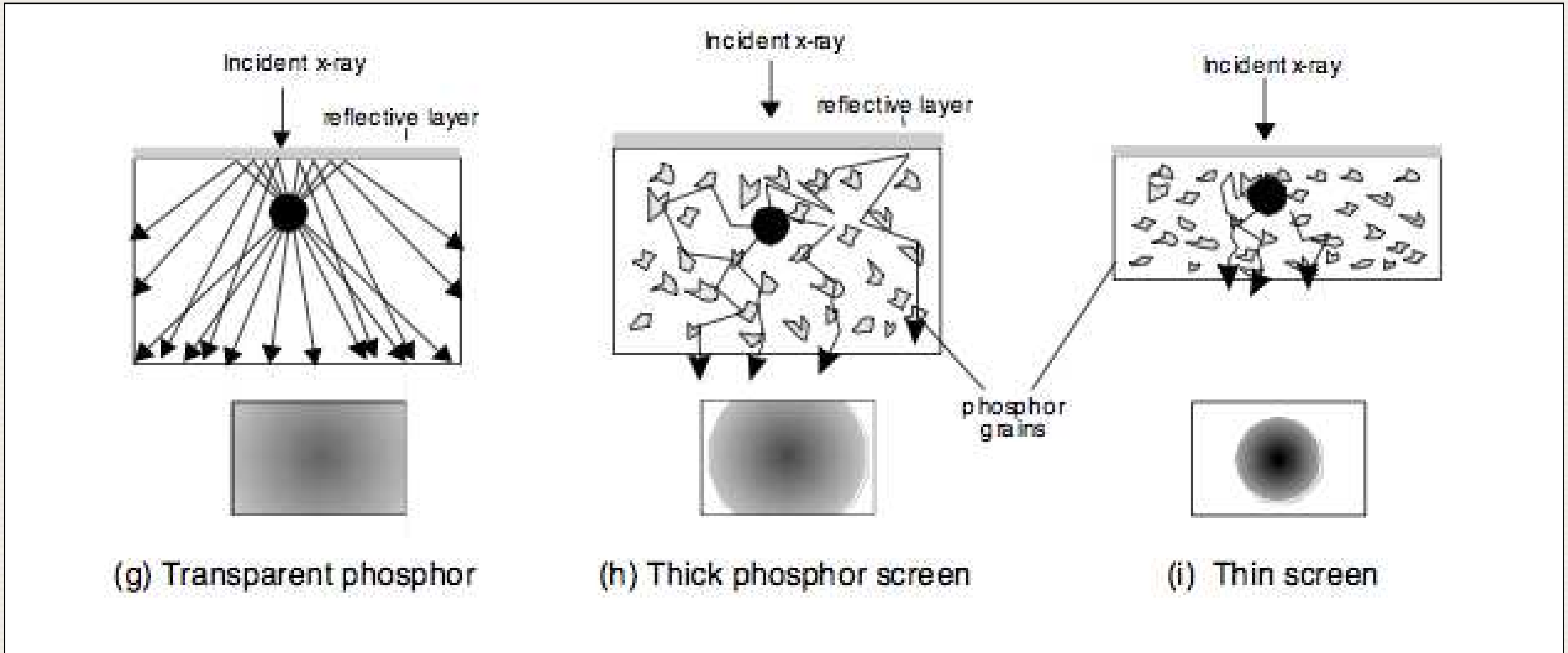
- **Del Aperture** and
- **Sampling Interval**

Mechanisms of resolution loss in **Direct** conversion layers (Photoconductors)

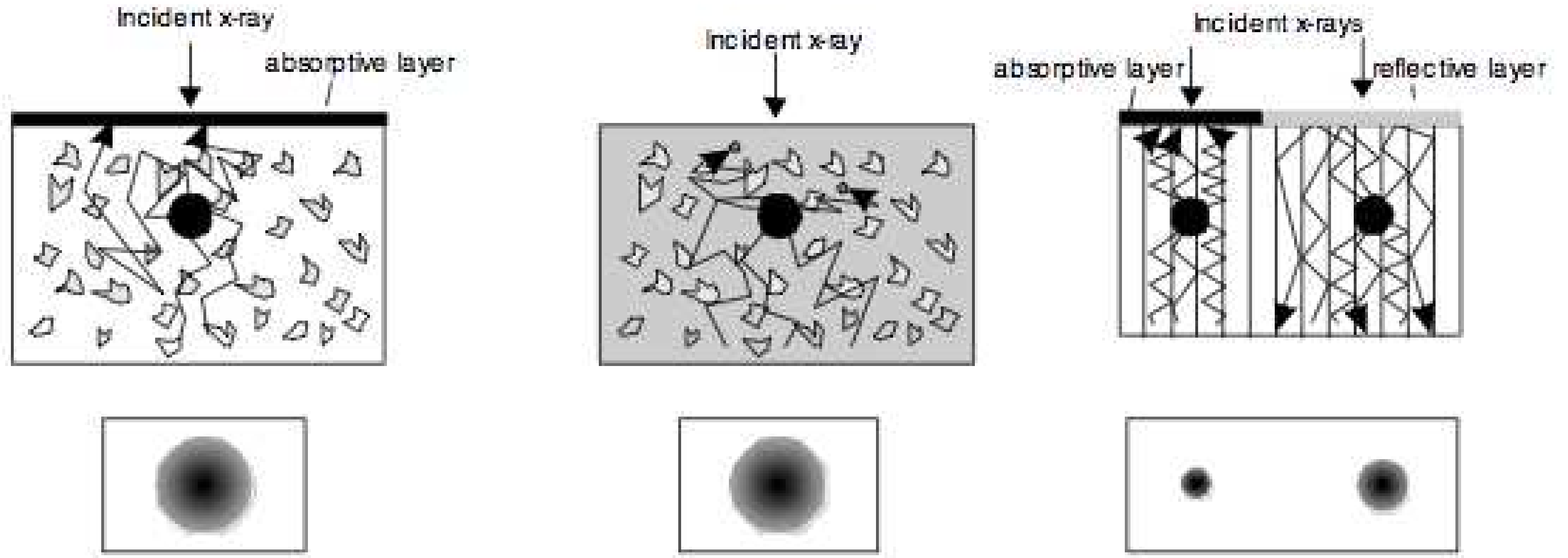


Photocondutor: (d) Charge generation (e) Charge collection (f) Trapping layer

Mechanisms of resolution loss in **Indirect** conversion types - Powder Phosphor Screens



Mechanisms of resolution loss in indirect conversion types - Powder Phosphor Screens and Structured Phosphors (usually called **Scintillators**)



(j) Absorptive backing on screen (k) Dye impregnated screen (l) Structured phosphor/scintillator

7.2.5 Fixed pattern noise

It is important that the radiographic imaging system provide uniformity, i.e. the sensitivity is constant over the entire image

If this is not the case, patterns that might disrupt the interpretation of the image will result

This is **Fixed Pattern Noise**

In an analogue imaging system, great pains must be taken in the design and manufacture of receptors to ensure that they provide a uniform response

In **Digital** systems post processing can be often be used to alleviate manufacturing limitations in uniformity

7.2.5 Fixed pattern noise

In a linear imaging system the fixed pattern noise, which can be expressed as a pixel-to-pixel variation in **Gain** and **Offset** (often due to a **dark current** in the related sensor), can in principle be corrected and their effect completely eliminated

The procedure is to correct patient images using:

- **Dark Field** (un-irradiated) and
- **Bright Field** (uniformly irradiated)

images

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7.3 FILM AND SCREEN-FILM SYSTEMS

In using a **Screen-Film** receptor to take an X ray image, a Radiographer must:

- **Load** a film into a cassette
- **Carry** the cassette to the examination room
- **Insert** the cassette into the X ray table
- **Position** the patient
- **Make** the X ray exposure
- **Carry** the cassette back to the processor to develop the film
- **Wait** for the film to be developed, and finally
- **Check** the processed film for any obvious problems to ensure that the film is suitable for making a medical diagnosis before returning to the X ray room

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The Screen-Film Combination & Cassette

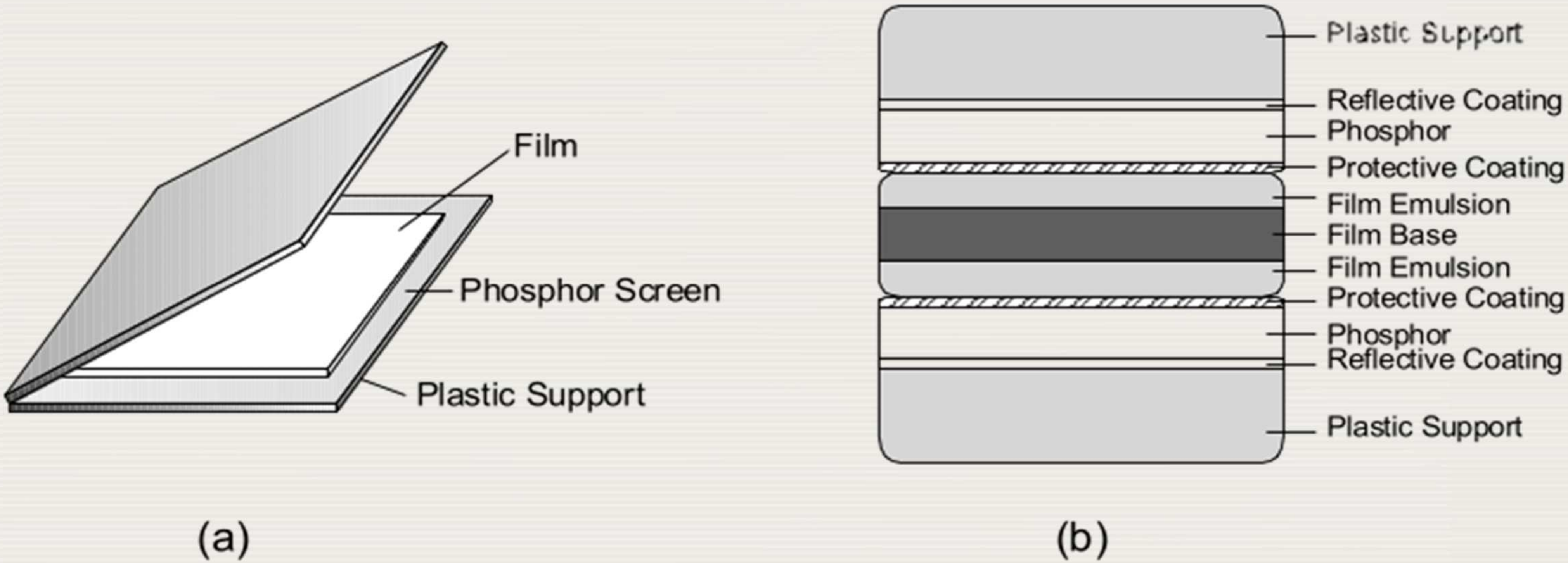
The screen-film **Combination** consists of phosphor screen(s) and film designed to work together enclosed within a **Cassette**

The cassette can be opened (in a dark environment) to allow the film to be inserted

When the cassette is closed, the film is kept in **Close Contact** with the screen

or more commonly a **pair** of screens, facing toward the film

The Screen-Film Combination & Cassette



Screen-Film receptor: (a) **Opened** cassette showing placement of film and position of screens, and (b) **Cross-sectional** view through a **dual screen** system used in general purpose radiography with the film sandwiched **between two screens**

The Screen-Film Combination & Cassette

Incident X rays **first** pass through the front of the cassette before reaching the screens

When they interact in the screen some of the energy deposited is converted to **Light**

which can travel from the interior to the **screen surface**

where it enters the optically sensitive part of the film called the **Emulsion** and

transfers its information into a **Latent Image** in the film

The Screen-Film Combination & Cassette

The film is then

- **Removed** from the cassette and
- **Developed** so that the latent image is converted

to a

Permanent Image

in the form of:

Silver deposited in the **Emulsion** layer of the film

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The Screen-Film Combination & Cassette

In most cases **two** screens face the film which has two emulsions:

one on either side of the film base

with an **Anti-Halation** layer placed between the two emulsions

During X ray exposure, the anti-halation layer is **opaque** and prevents light crossing over from one emulsion to the other, thus reducing **Cross-Talk**

and hence **Blurring**

The Screen-Film Combination & Cassette

The opaque anti-halation layer is removed during **Film Development**

rendering the film **Transparent** for subsequent viewing

For the highest resolution (e.g. Mammography) a **single screen** in the back of the cassette may be used in contact with a single emulsion film

The Screen-Film Combination & Cassette

Film emulsions can be used as **Direct** receptors for X ray images

The earliest X ray images were taken with film alone
In fact film was used in this way for **Mammography** up to the 1960s

The sole remaining clinical application for film without screens is in dental radiography using **Intraoral** films

However, the X ray absorption efficiency for such films is relatively poor (~1-5%)

Thus currently all diagnostic X ray film images are obtained using screen(s) in **conjunction** with the film

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Screen Structure

Phosphor grains are combined with a **Polymer Binder** and deposited on a substrate or backing

The ratio of binder volume to phosphor volume in the mixture controls the fractional volume of the phosphor layer finally occupied by air pockets or **Voids**

Typically the binder is nitrocellulose, polyester, acrylic or polyurethane and the plastic support or **Backing Material** is also a polymer e.g. polyethylene terephthalate 200-400 μm thick

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Screen Structure

The use of a black or white backing permits adjustments of the **Reflectivity** and **Absorptivity** at the phosphor interface

In most screens the typical phosphor **Grain Size** is 3-8 μm

Usually a very thin transparent **Protective Layer** is subsequently applied to complete the screen structure

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Phosphor

Screen-film systems employ a phosphor in the initial stage to **absorb** X rays and produce light

Phosphors work by exciting electrons from the valence band to the conduction band so creating **Electron Hole Pairs** (EHPs) which are free to move within the phosphor

Some of these will **recombine** without giving off any radiant energy

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Phosphor

However in an **Activated** phosphor, most (>90%) EHPs will recombine at an activation centre (created by atomic impurities called **Activators**) and in the process emit light

Since light photons each carry only a small amount of energy (~2-3eV), **many** light photons can be created from the absorption of a **single** X ray

The specific **Colour** of the light emitted is related to the optical transitions in the activator

By changing the activator, the light colour can be changed **changed**

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Phosphor

This **Quantum Amplification** is the **Conversion Gain** of the phosphor

The original screens used until the 1970s were **Calcium Tungstate** ($CaWO_4$) which is naturally activated and hence not particularly efficient and emits light in the deep blue and UV radiation

More recently **Rare Earth** phosphors with explicit centres for the emission of light at the activator site have resulted in the most commonly used material **Gadolinium Oxysulphide**

Gd_2O_2S:Tb with Tb in dilute amounts 0.1-1% as an activator

Phosphor

For Gd_2O_2S, an X ray photon energy of 50 keV is equivalent to that of ~20,000 **Green** light quanta (E = 2.4 eV) although due to losses typically only 1,800 are produced in practice

The **Green** emission from the rare earth phosphors has also required a change from conventional film in **two** regards:

1. Ordinary film is sensitive only in the blue, it requires additional sensitization to be green sensitive and then is called **Orthochromatic**
2. Green light is far more penetrating than blue light and so requires an anti-halation layer to prevent **Crossover** of images between emulsions

Thickness

The choice of the **Thickness** of a radiographic screen has to balance:

- the increase in $\mathbf{A_Q}$ with thickness - which favours a **thick** screen, and
- the efficient **escape** of light and usually more importantly blurring due to spreading of light - which favours a **thin** screen

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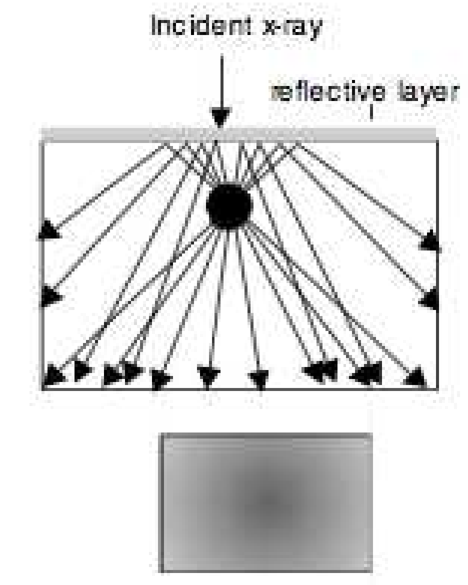
Thickness

In order to create a sharp X ray image, a transparent phosphor screen would be ineffective since light could move large distances within the phosphor and cause excessive blurring



(g) Transparent phosphor

Thickness

Instead X ray screens are made highly scattering or **Turbid**

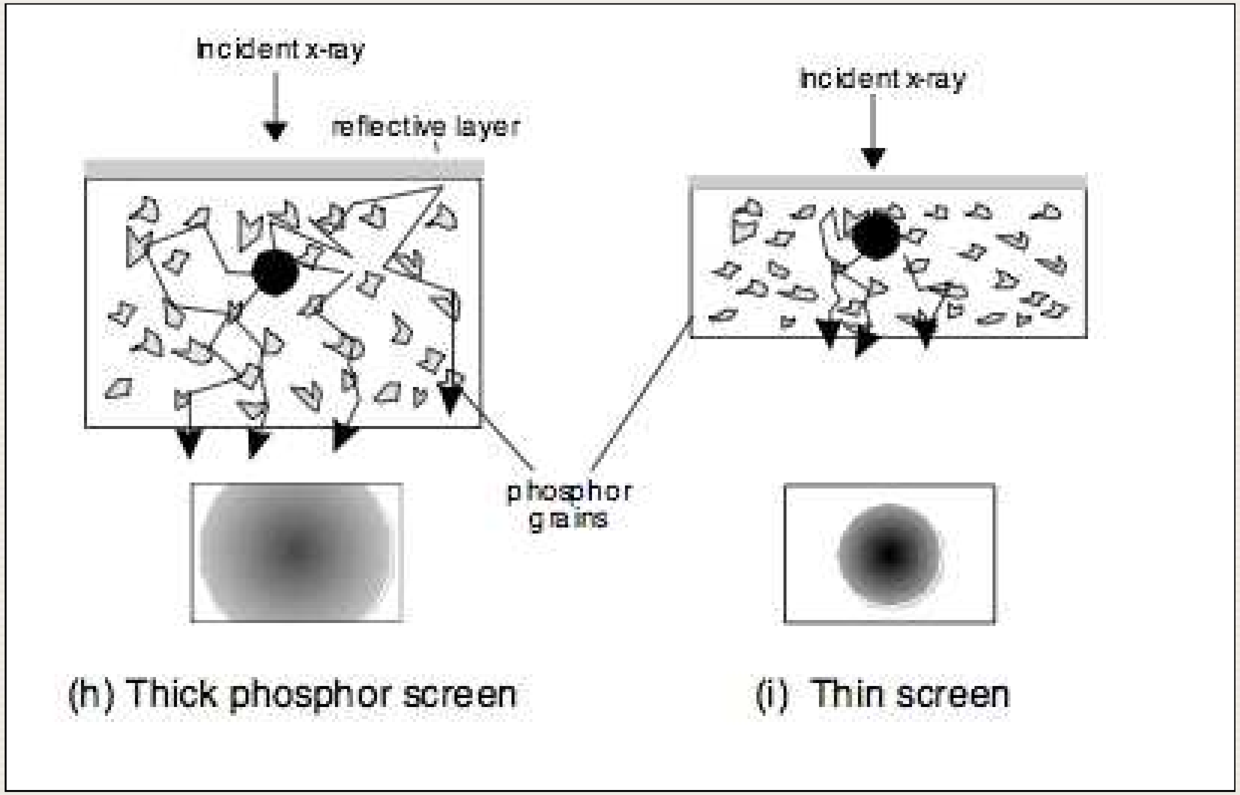
This is accomplished by using high refractive index phosphor grains **embedded** in a low reflective index binder

Once a light photon exits a grain it tends to **reflect** off the neighbouring grain surfaces rather than passing through them

Thus the lateral spread of the light is confined by **Diffusion** (multiple scattering) which helps to maintain the spatial resolution of the phosphor layer

Thickness

Effect of phosphor thickness on spatial resolution of a **Turbid** phosphor screen:



Thickness

Because of the presence of the binder material the amount of phosphor present in a screen is usually quoted in terms of the **Screen Loading** or areal density, i.e.

the mass of phosphor per unit area of the screen

Typical values for Gd_2O_2S phosphor screens or screen pairs are from **20-200 mg/cm^2** depending on the application

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Optical Design

Since the X rays **also** pass through the film:

Some darkening will be developed due to **Direct** interactions of X rays with the film emulsion

Usually this is so **small** that it can be ignored

Optical Design

The **Optical Design** of a phosphor screen critically affects its imaging performance

Factors such as:

- Phosphor grain size
- Size distribution
- Bulk absorption
- Surface reflectivity
- Intentionally entrained tiny **bubbles** to increase scattering

can have significant effects on the image quality

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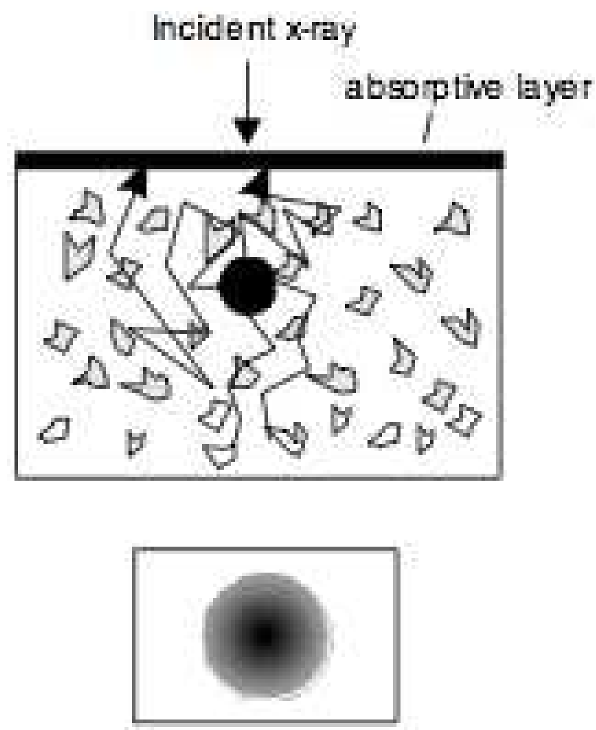
Optical Design

An **Absorptive** backing helps reduce blurring by preferentially absorbing the long path length photons, but at the cost of reduced overall **Conversion Efficiency**

With a **Reflective** backing, most of the light escapes the front of the screen and is available to be recorded



(j) Absorptive backing on screen

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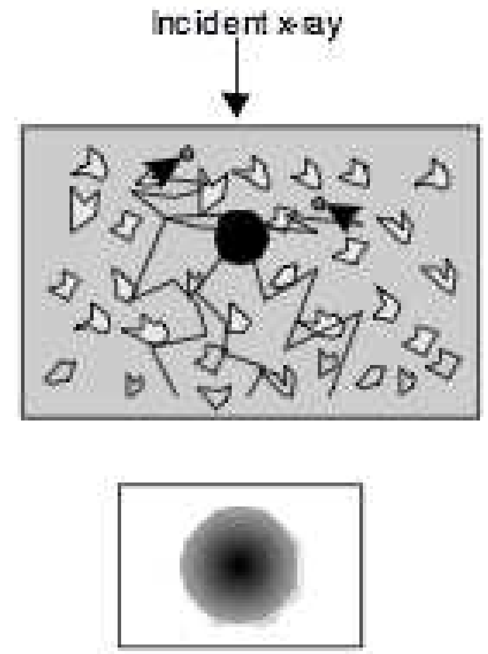
Optical Design

Light absorbing **dye** can be added to the screen to enhance the resolution - this is **similar** to but not identical in effect to an absorptive backing



(k) Dye impregnated screen

Photographic Film is a unique material that is sensitive to a very few quanta of light

At normal ambient temperature, it can:

- Record a **Latent** optical image from a fraction of a second exposure
- **Maintain** this latent image for months, and
- Eventually be **developed** without significant loss of information

It is also used as a **Display** and **Archiving** medium

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Film Structure

The photographic process uses a thin layer (called the **Emulsion**) of silver halide crystals called **Grains** suspended in **Gelatine** and supported on a transparent film base

The grains are primarily **Silver Bromide** (~95%) with the balance being silver iodide and sometimes trace amounts of silver chloride

The grains are of variable size (**~μm**) and shape (**cubic** or **tabular** i.e. flat) depending on the application

Film Structure

The **Film Base** thickness is standardized to ~180 µm to allow smooth transport through **Automatic Film Processors**

The emulsion is typically 3-5 µm thick and can be on one (**Single Sided**) or both (**Double Sided**) sides of the base

During the manufacturing process the grains are sensitized by introducing **Sensitivity Specks** onto the grains by reaction with sulphur compounds

The Photographic Process

The **key feature** which gives unexposed film its long shelf life is that more than one light photon must impinge on an individual silver halide crystal grain in order to create a stable latent image

A single light photon creates an electron that is trapped for a short time (about a second) in a unique point on the grain called the **Sensitivity Speck**

If no other photons are absorbed by this grain, then the electron will escape from the grain

The Photographic Process

However, if a few more electrons are released in the same grain within this time:

the electrons stabilize each other at the sensitivity speck and a **Latent Image** is established

The **multi-electron** process is key to understanding not only the long shelf life and the non-linear response of film to light but other behaviours to be described such as **Reciprocity Failure**

Development of the Latent Image

After exposure of the film to light, the **Latent Image** is formed at the sensitivity specks on the individual film grains

Film Processing converts the latent image to a viewable permanent image

Film processing can be split into **three** phases:

- Development
- Fixing
- Wash

These processes are facilitated by the **suspension** of the grains in a thin layer of water-permeable gelatine supported on a flexible substrate

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Development of the Latent Image

Chemicals are transported to the **crystals** without disturbing their positions when the film is dipped into chemical solutions

The **Development Process** turns a sensitized transparent crystal of silver halide grain into a **Speck** of metallic silver that absorbs light and therefore is opaque

Since these are very small (**<1 μm**), light absorption dominates over reflection due to multiple scattering and they appear black in the same way as any finely powdered metal

Development of the Latent Image

The **Gain** of the system in terms of

the **number** of silver halide molecules converted into metallic silver per absorbed light photon

is a staggeringly large number

$> 10^8$

which is the basis of its uniquely high **Sensitivity**

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Fixing of the Image

After the Latent Image has been **Developed**, the unexposed and therefore undeveloped transparent silver halide crystals remain within the gelatine layer

Thus the emulsion is **still sensitive** to light and if further exposed, would sensitize the grains which could self-develop and change the image

During the **Fixing** stage, these undeveloped silver halide crystals are dissolved and removed chemically

thereby **Fixing** the image

Wash - Making it Archival

Next after fixing is the **Water Wash** where the processing chemicals and any remaining dissolved silver halide are removed

leaving only the insoluble silver grains **embedded** in pure gelatine

Drying removes the excess water solvent from the gelatine and results in a completely permanent archival material known as

Photographic Film

Automated Processor Design

While the **Development Process** can be performed **manually** with trays filled with chemical solutions for a few images per day, or deep tanks for more, these are very labour intensive processes which are difficult to control

Automated Processors are therefore used which are:

- Much **faster** in operation (typically 90 s from introducing the exposed film to receiving the dried image)
- Suitable for maintaining consistent image quality by not only keeping the **Speed Point** consistent but the complete **Characteristic Curve**

Automated Processor Design

The **basic concept** is to use deep processing tanks kept at a constant temperature and rollers immersed in the tanks to transport the film and ensure that it is subjected to **consistent** processing conditions

A key additional process is the **Drying** step which means that the film can be used as soon as it emerges from the processor

In practice, the simplest practical arrangement is for the processor to be built into the wall of the **Dark Room** with film removed from the cassette by a Darkroom Technician with a final image deposited in a tray outside the darkroom

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Automated Processor Design

The cassette is brought into the darkroom by an interlocked pass through or **Light-Lock** which:

- Permits the Darkroom Technician to keep his eyes dark adapted and
- Reduces risk of accidental exposure

In more advanced departments the darkroom can be completely eliminated and the film **automatically** removed from the cassette and reloaded by an automated system

Automated Processor Design

Artefacts related to the automatic film processors include:

- picking up of unwanted debris resulting in **Dust Specks** on the developed film, and
- periodic variations in intensity of the film (**Roller Marks**), which arise primarily from the variation of speed in the tanks due to the difficulty of manufacturing and maintaining the rollers to be exactly round and smooth

In some dry environments **Electrostatic Discharges** can cause characteristic artefacts

Optical Density

A film radiograph is an image of the incident pattern of X radiation in which

the distribution of the number of developed grains

is related to

the distribution of the number of X rays incident on the receptor

When viewed on a **Viewbox** (source of back illumination), the developed grains reduce the light transmission through the film so that the viewed image is a **Negative**

Optical Density

The **Transmission** of light through a film or more precisely its inverse, the attenuation, is expressed as the **Optical Density** or **OD** which is defined by:

$$\mathbf{OD = \log_{10} (I_I / I_T)}$$

where $\mathbf{I_T}$ is the intensity of light transmitted through the film due to the intensity of light $\mathbf{I_I}$ incident on it

Thus for $\mathbf{I_I / I_T = 10}$ the corresponding optical density is **OD=1**

Optical Density

The transmissions of **two** absorbers placed in sequence in an optical path are **Multiplicative**

so that the logarithms are **Additive**

Thus two emulsions each of **OD=1** would have a total

OD=2

Optical Density

Although the reason for a **Negative** image is purely historical and determined by the nature of the photographic process

the display of images on **Digital** systems where there is freedom to change is usually very similar to that found with film

This is no accident as the **look** of radiographic film has been tailored over the years to be optimum for its purpose of displaying the image most efficiently to the human observer

Densitometry, Sensitometry

In a radiology department dependent on **Screen-Film**, the instability of film processing systems is the single most problematic aspect

Maintaining a match in characteristics between films processed from one day to the next, from morning to afternoon and from one film processor to another is an essential chore called

Film Quality Control

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Densitometry, Sensitometry

Film Quality Control requires the daily monitoring of the processor with the exposure of a film from a standard batch using a standardized step wedge with a **Sensitometer**, and the measurement of the developed film with a **Densitometer**

Essential to this approach is consistent testing at the **same time** each day, measurements of the temperature of the solutions and keeping them chemically fresh

Also essential is the need to have predetermined quantitative **Action Points** established, and associated actions to correct any deficiencies

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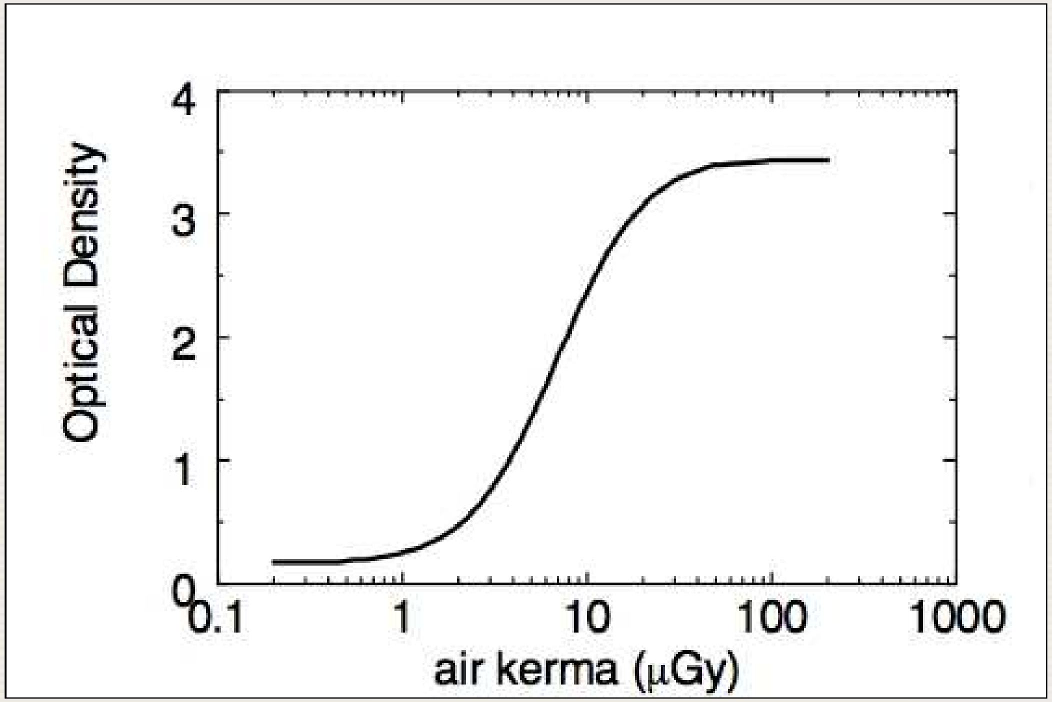
H & D Curve (Hurter & Driffield Curve)

The curve which relates the Optical Density of the film to the logarithm of the exposure is known as the

Characteristic Curve or

H&D curve

after Hurter and Driffield who first introduced its use



H & D Curve (Hurter & Driffield Curve)

The **form** of the curve can be explained as follows:

For photographic films there is some film density, known as **Base plus Fog**, even in the absence of any exposure

The **Base** component is the transmission of the film substrate or base

The **Fog** is the absorption in the emulsions due primarily to unwanted self-development of grains unexposed to light

H & D Curve (Hurter & Driffield Curve)

Together **Base plus Fog** in fresh films are **~0.3 OD** or less

A small amount of light will have difficulty in creating any kind of developable image due to the meta-stability of single electron excitations

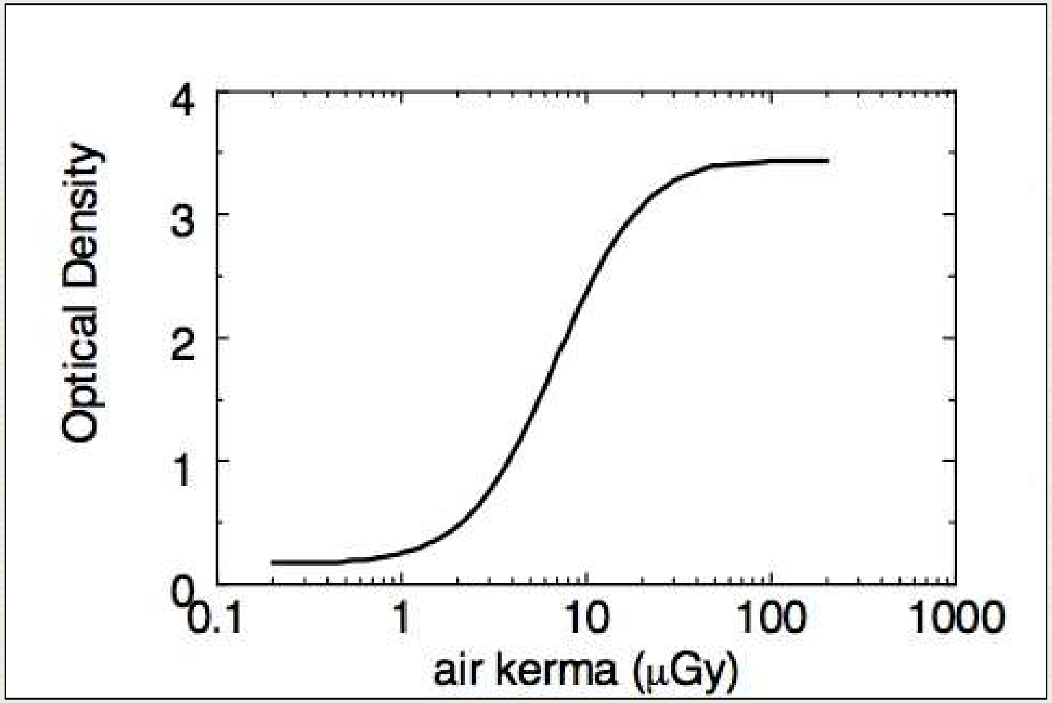
Thus, **small** amounts of light cause little darkening of the film

When enough light is incident, the film starts to develop (the **Toe**), then it responds more rapidly and the approximately **Straight Line** part of the curve emerges

H & D Curve (Hurter & Driffield Curve)

The **Gradient** or **Gamma** of the curve (i.e. its slope) actually varies continuously with optical density

Once there is sufficient light to develop most of the grains, then saturation begins, giving rise to the **Shoulder** of the curve, i.e. a flattening at high exposure



H & D Curve (Hurter & Driffield Curve)

The **Characteristic Curve** can be modified in many ways

The film designer can adjust the **Slope** and/or **Sensitivity** of the curve allowing adaptation to different applications by changing:

- the **Grains**
- **Size** distribution
- grain **Loading** of the film
- **Development** process

H & D Curve (Hurter & Driffield Curve)

The **Latitude** of a film is the range of exposures for which a sensible response of OD occurs

In practice it is measured between the end of the **Toe** and the beginning of the **Shoulder**

It is desirable that the **Gamma** be as large as possible to show low contrast objects while simultaneously maximizing the latitude

Satisfying these conflicting requirements is the **Art** of producing a satisfactory film design

Typical values of **Gamma** for radiography are in the range of **2-3**

H & D Curve (Hurter & Driffield Curve)

When X rays are used to irradiate film directly, an X ray interacting with a silver grain in the emulsion can deposit sufficient energy to create a primary electron

Which will in turn deposit its energy in the **immediate neighbourhood**

This will potentially create enough electrons within each of the 10-100 grains near to the primary interaction to **sensitise** each to the point that it will be developable

H & D Curve (Hurter & Driffield Curve)

Thus, the optical H&D response is bypassed and the initial response to X ray absorption is **Linear** in exposure

In addition there is little risk of **Image Fading** or **Reciprocity Failure** which accounts for its usefulness as an imaging radiation dosimeter

7.3.5 Reciprocity

An image receptor which produces the same response for a given exposure independent of the exposure time is said to exhibit **Reciprocity**.

Film has remarkably good reciprocity in the range of exposures normally used in photographic cameras, i.e. from **~0.001-0.1 s**

This also covers **most** exposure times encountered in medical radiography

However for the very long exposure times of **2-5 s** used in conventional film **Tomography** and **Mammography**, reciprocity failure can be important

The **reason** for reciprocity failure lies in the photographic process - the photographic grain is designed not to be sensitised unless sufficient light falls on it in a short time

Although this **saves** it from fogging in the dark, the by-product is reciprocity failure at long exposure times

Specifically this means that a long exposure will need a **longer** time than extrapolation by reciprocity from shorter exposure would indicate

This can be of the order of **30-40%** increase in time for 2-5 s exposures

There is a dependence of **Speed** (i.e. relationship between darkening of film to radiation) and **Beam Quality** for screen-film systems

which is usually compensated for by the **Phototimer** having knowledge of the generator settings

Thus it is not usually evident on a well calibrated system

For example for a Gd_2O_2S screen with a K-edge ~50 keV and absorption of ~40% at its **Sweet Spot** of 80 kV and usual filtration, the speed is maximized

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Speed drops somewhat with increased kV but more significantly at lower kV

Similar results are found for most screen-film combinations, with the **Sweet Spot** depending on the K-edge of the primary X ray absorber(s) in the screen

Key factors in the design of a screen-film cassette are:

- Excellent **contact** between the screen and the film to prevent blurring or loss of light
- The front surface of the cassette must be **easily** penetrated by X rays and not cause scatter

Often at the back of the cassette is a **Lead Layer** to control X ray **Backscatter** from external objects whose highly blurred image could otherwise be superimposed

Film requires **3-10** photons per grain before a developable grain can be created corresponding to an effective DQE for light of a few percent

Fortunately, the gain of the screen in terms of the light photons released per complete absorption of an X ray photon is of the order of

- 800 (for **Mammography** ~20 keV) and
- 2,000 (for diagnostic energies ~50 keV)

Thus the **Effective** quantum gain of X rays to developable grains in the film is 800-2000 x 1-2% or

- 8-16 grains/X ray for **Mammography** and
- 20-40 grains/X ray for **General Radiography**

This is the information on which the

Quantum Accounting Diagram

for Screen-Film was established

Each X ray interacting with the screen creates **many** light photons

but because of the **small size** of the film grains (<1 μm) compared to the **screen blurring** (~100-500 μm)

usually only one or at most a few light photons from the same X ray will interact with each **individual** grain

Thus there will be essentially no correlation between **individual** X ray events and **individual** grains

resulting in the same shape of the response curve for the **Screen-Film** system when irradiated by X rays as for the **Film** when irradiated by light

i.e. the optical H&D curve as measured with a **Sensitometer** colour matched to the emission of the phosphor

This is a key point in validating **Film Sensitometry** for Radiography

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The variation of the **probability** of X ray interaction with **depth** in the phosphor screen is **exponential**

so that the number of interacting quanta and the amount of light created will be proportionally greater near the X ray **entrance surface**

The highest-resolution screen-film systems are therefore generally configured from a single screen placed such that the X rays pass through the film **before** impinging on the phosphor

This **Back Screen** configuration improves the spatial resolution of the final image compared to the alternative **Front Screen** configuration

It can be noted that due to the thickness (~0.7 mm) of the standard **glass substrate** currently used for active matrix fabrication

and its consequent **significant absorption** of X rays

Flat-Panel systems are all configured in the **Front Screen** orientation

However, for all but the highest resolution requirements dual-screens are used as they can make a better trade-off between high **Quantum Efficiency** and high **Resolution**

Screen **Nomenclature** always stresses the **positive**:

A high resolution and low $\mathbf{A_Q}$ screen is referred to as a **High Resolution** screen

a low resolution, as a **High** $\mathbf{A_Q}$

MTF, NPS & DQE of Screen-Film Systems

The MTF of Screen-Film **primarily** depends on the need of the application

It can be excellent especially as measured by the single criterion of **Limiting Resolution**

usually defined at the frequency **f** for which the **MTF(f)** drops to **4%**

The **variation** of MTF with screen thickness/loading is very significant

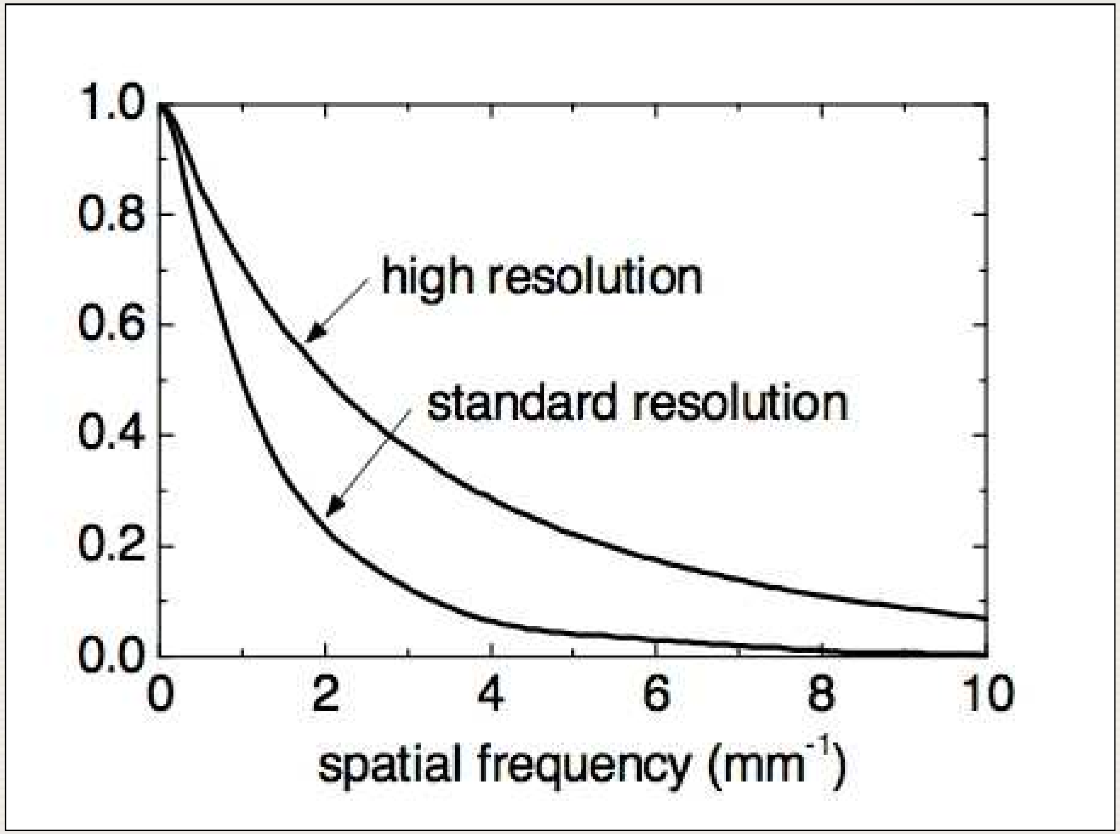
MTF, NPS & DQE of Screen-Film Systems

The variation of MTF with spatial frequency (**f**) for two double screen systems for different applications:

Lanex Fine for high resolution - bone - and

Lanex Regular for general radiography

and hence different screen thicknesses and optical design



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MTF, NPS & DQE of Screen-Film Systems

Only recently have **Digital** systems been able to approach their capabilities

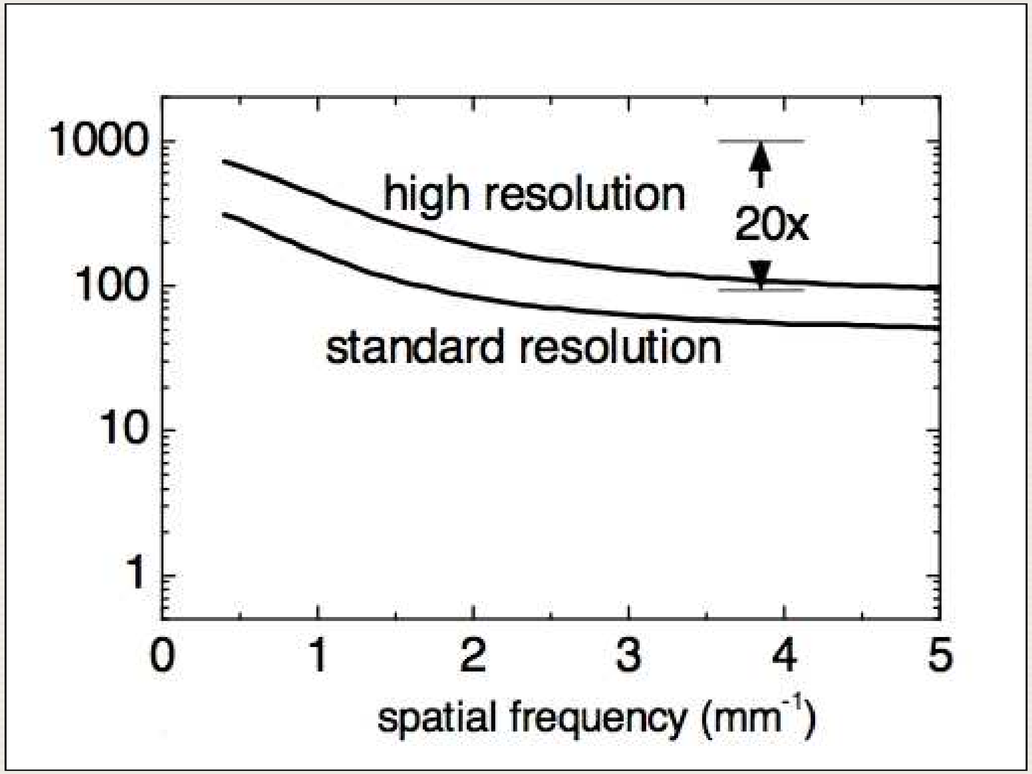
It was controversial for many years as to whether or not high frequency information available only in Screen-Films was necessary in **Mammography**

MTF, NPS & DQE of Screen-Film Systems

The answer to this question is obtained by looking at the **NPS** for Screen-Film

which demonstrates that the noise at high spatial frequency reaches an **asymptote**

where the noise is **white** and quantum noise is negligible due to the negligible MTF of the screens at these frequencies



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MTF, NPS & DQE of Screen-Film Systems

This is seen to be about a factor **20** times lower than the noise power extrapolated to zero spatial frequency

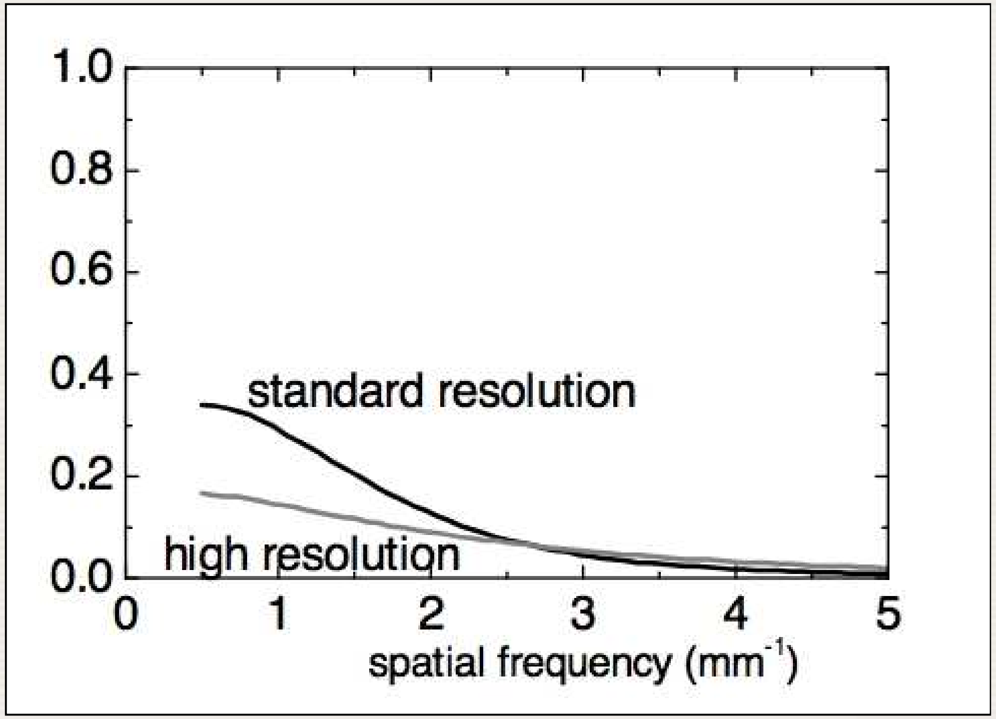
This result could have been predicted from the **Quantum Accounting Diagram**

which shows that the **Secondary Quantum Sink** is

20 silver grains per absorbed X ray

MTF, NPS & DQE of Screen-Film Systems

The dire consequence is that the **DQE** shows an extremely rapid drop off with **f**



which results in the **merging** of the standard and high resolution screens above **2.5 mm^{-1}** to an essentially negligible value

despite the high resolution screen showing a noticeable MTF even at **10 mm^{-1}**

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In a **Digital** imaging system, at some stage, 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 receptor elements or **dels**

These are usually **square**, and spaced at **equal intervals** throughout the plane of the receptor

The **Pixel** is the corresponding elemental region of the image

In the **Intensity Dimension**, the signal is digitized into one of a finite number of levels which are expressed in binary notation as **bits**

To avoid degradation of image quality, it is essential that the **Del Size** and the **Bit Depth n** (when **n** is given by $\mathbf{2^n}$) are appropriate for the requirements of the imaging task

The **Matrix Size** or the coverage of the array is different depending on the application and the size of the body part to be imaged and the magnification

The fractional area of the del which is active has an upper limit of unity but can often be smaller than that due to a reduced **Fill Factor**

The linear dimension of the active portion of each del defines an **Aperture**

The aperture determines the **Spatial Frequency Response** of the receptor

If the aperture is square with dimension, **d**, then the MTF of the receptor will be proportional to

$$|\mathrm{sinc}|(\pi f d)$$

times the intrinsic MTF of the equivalent analog receptor

where **f** is the spatial frequency along the **x** or **y** directions

the MTF will have its first zero at the frequency

$$f = 1/d$$

expressed in the plane of the receptor

For example

A receptor with **d** = 200 μm will have an MTF with its first zero at **f** = 5 cycles/mm

Also of considerable importance is the sampling interval, **p**, of the receptor, i.e. the **Pitch** in the receptor plane between corresponding points on adjacent dels

The **Sampling Theorem** states that only frequencies **f** in the object less than **<1/2p** (the Nyquist frequency, $\mathbf{f_N}$) can be faithfully imaged

Thus if the pattern contains higher frequencies, then a phenomenon known as **Aliasing** occurs

wherein the frequency spectrum of the image beyond the $\mathbf{f_N}$ is:

- mirrored or **folded** about the $\mathbf{f_N}$ in accordion fashion and
- **added** to the spectrum of lower frequencies, increasing the apparent spectral content of the image below $\mathbf{f_N}$

It is important to realize that **both** signal and noise can show aliasing effects

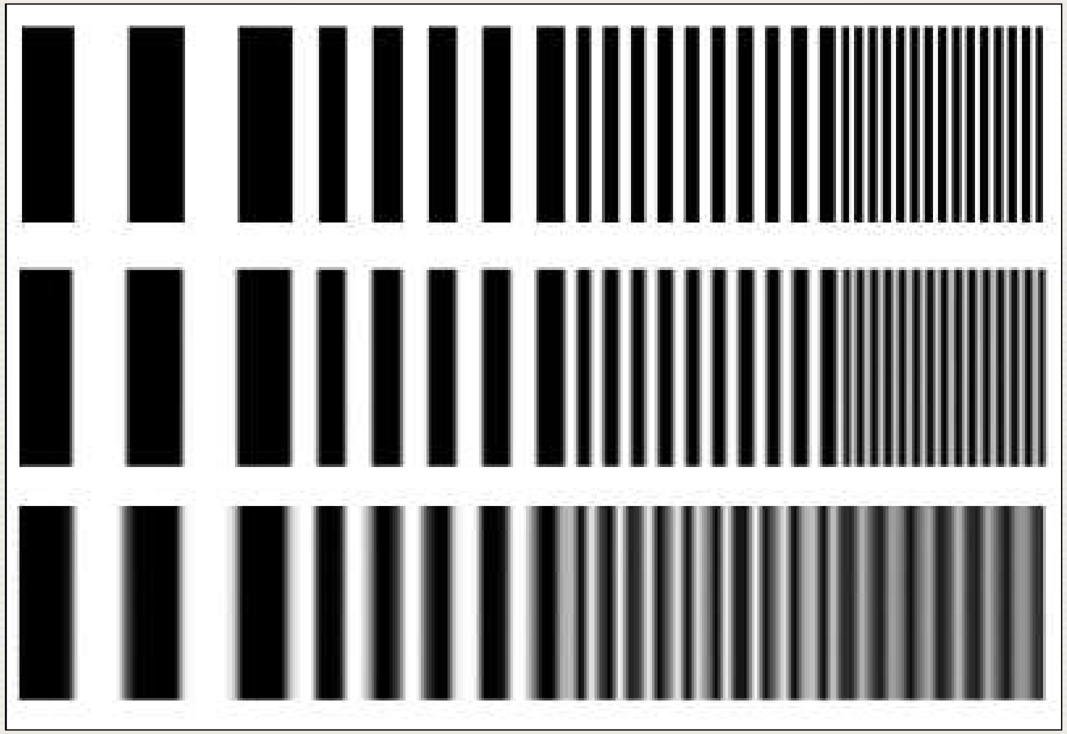
In a receptor composed of **discrete elements**, the smallest possible sampling interval in a single image acquisition is

$$\mathbf{p = d}$$

Even in this most favourable case, $\mathbf{f_N = 1/2d}$ while the aperture response only falls to zero at $\mathbf{2f_N}$

If the **Del Dimension** is less than the **Sampling Interval**, then the zero is at $\mathbf{>2f_N}$, further increasing the aliasing

Aliasing effect on the image of a bar pattern:



Top: original

Centre: pixel pitch sufficiently small, all bar patterns are resolved correctly despite some blurring

Bottom: pixel pitch too large to resolve the finest bar pattern, aliasing occurs

Aliasing can be avoided by **Band Limiting** the image before sampling

i.e. **attenuating** the higher frequencies such that there is no appreciable image content beyond $\mathbf{f_N}$

This may be accomplished by other blurring effects **intrinsic** to the receptor

These mechanisms

- can have **different** effects on noise and signal and
- may not necessarily reduce noise and signal aliasing in the **same** manner

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7.4 DIGITAL RECEPTORS

7.4.2 Computed Radiography (CR)

In **CR** the imaging plate or IP (a screen made using a **Photostimulable Phosphor**) is:

- **Positioned** in a light tight cassette
- **Exposed** to a patient X ray image and then
- **Produces** a digital image in a system which
  - **Extracts** the exposed plate from the cassette while protecting it from ambient light
  - **Reads** it out
  - **Erases** the image and
  - **Returns** it to the user in the cassette ready to be reused

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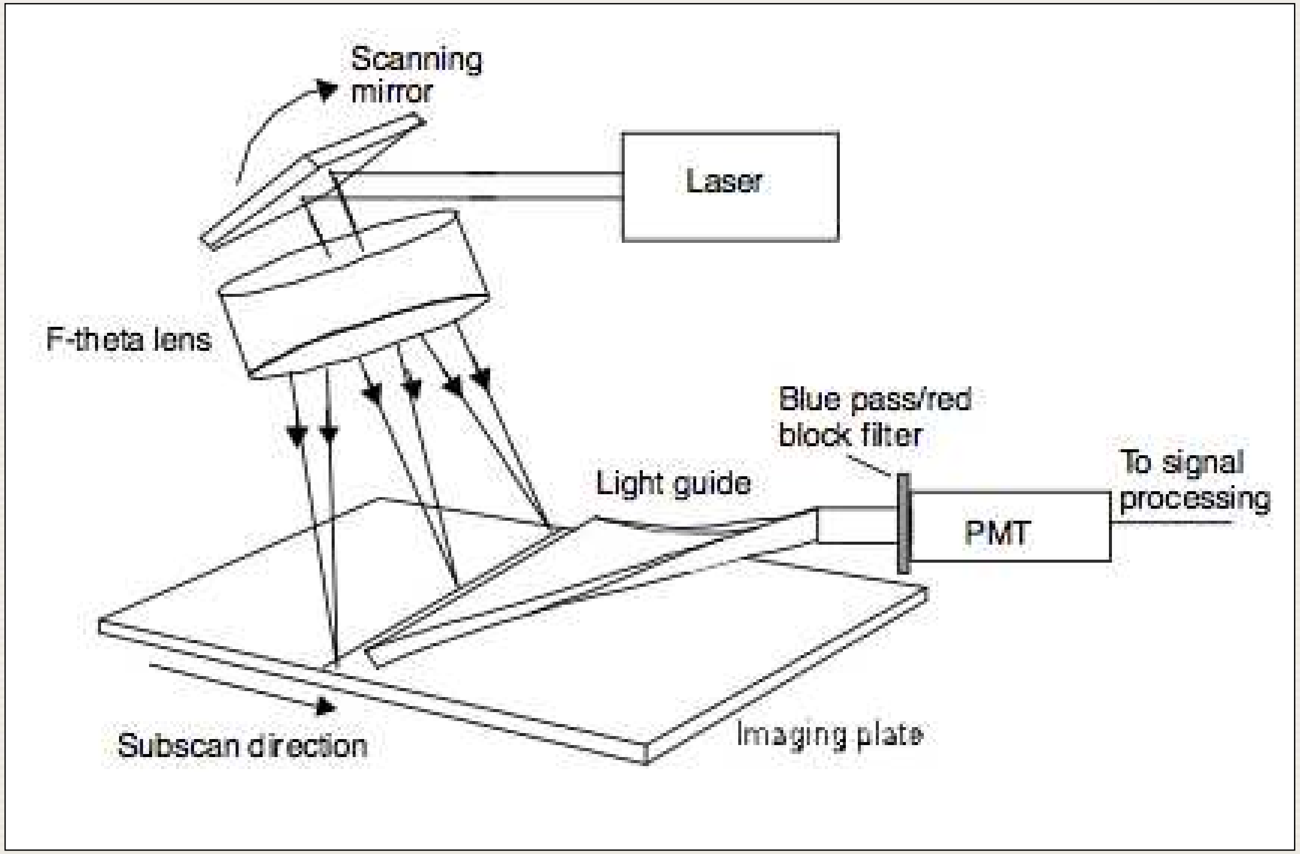
CR system based on the use of reusable **Photostimulable Phosphor** plates housed in cassettes

Readout of plate in **Laser Scanner**

with photostimulated light collected in **Light Guide** and detected by a **PMT**



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CR belongs to a class of systems which could be called **Reusable Plate Technologies**

and **directly replace** Screen-Film

There are currently no competing technologies for the reusable plate despite the fact that

- the **Image Quality** of CR is poorer than DR
- it requires a **larger exposure** to produce acceptable images

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Method of Latent Image Formation

The **Photostimulable Phosphor** screen in the IP is very similar to a conventional X ray screen except it uses a phosphor that contains **traps** for excited electrons

Most photostimulable phosphors are in the **Barium Fluorohalide** family

typically **BaFBr:Eu**

X ray absorption mechanisms are **identical** to those of conventional phosphors

Method of Latent Image Formation

The photostimulable phosphors **differ** in that the useful optical signal is

- not derived from the light emitted **in prompt response** to the incident radiation as in conventional screen-film systems
- but rather from subsequent **Stimulated Light Emission** when EHPs are released from traps

The initial X ray interaction with the phosphor crystal causes **EHPs** to be generated

Method of Latent Image Formation

Some of these electrons produce **blue/green** light in the phosphor in the normal manner but this is not used for imaging

Instead the phosphor is intentionally designed to contain **metastable EHP traps** that store a latent image as a spatial distribution of trapped electrons and holes

By stimulating the phosphor by irradiation with **red** light, these EHPs are

- released from the traps and are
- free to move in the valence band (**holes**) and conduction band (**electrons**)

Method of Latent Image Formation

These mobile EHPs subsequently trigger the emission of shorter wavelength (**blue**) light

CR screens (also called **Imaging Plates**) are exposed in a cassette with only one screen

This reduces the **Absorption Efficiency** compared to a dual screen combination

and this is an **intrinsic disadvantage** of CR compared to Screen-Film imaging

Image Readout

The **Readout** system for photostimulable phosphor plates uses a red laser beam **Flying Spot** scanning system to stimulate the screen on a point-by-point basis

exciting **Photostimulated** light

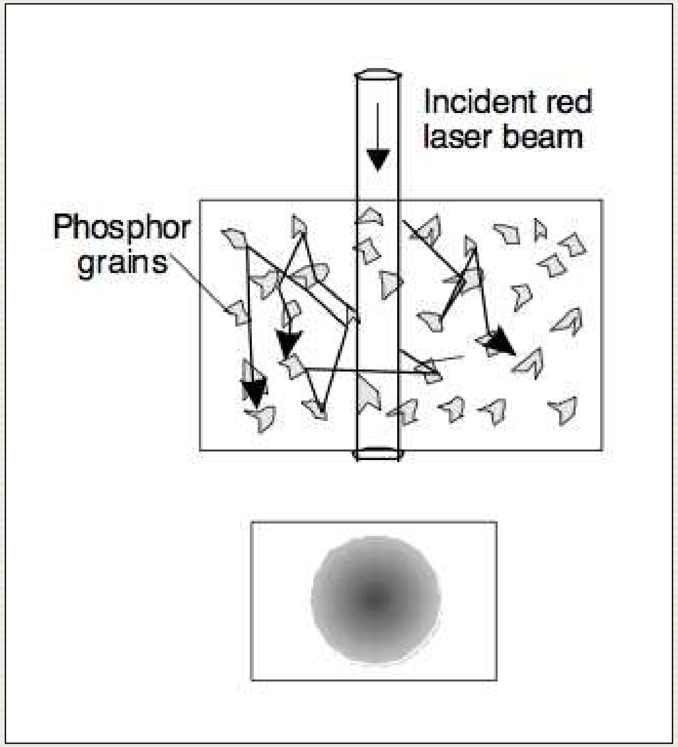
(usually **blue**)

from the screen in proportion to the previous X ray irradiation of that point

In practice, depending on the laser intensity, readout of a photostimulable phosphor plate yields only a **fraction** of the stored signal

Image Readout

The blue light is collected by a **Light Guide** that is a critically important component in the avoidance of a Secondary Quantum Sink



i.e. the light is funnelled to a PMT that detects and amplifies the signal

Note that in contrast to the situation with screen-film the **scattering** of the blue light does not degrade the image resolution

However the scattering of **red** light on its way to the photo centres does

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Image Readout

The **Primary Source** of light blurring in CR:

the spreading by scattering of the incident exciting light gives rise to the blurring

the light that comes out of the phosphor grains is irrelevant and does not contribute to resolution loss

Finally, the image plate is then flooded with light to **erase** any residual image and is then ready for **reuse**

Image Readout

The **advantages** of Photostimulable Phosphor systems are that they are digital systems with a very high **Dynamic Range**

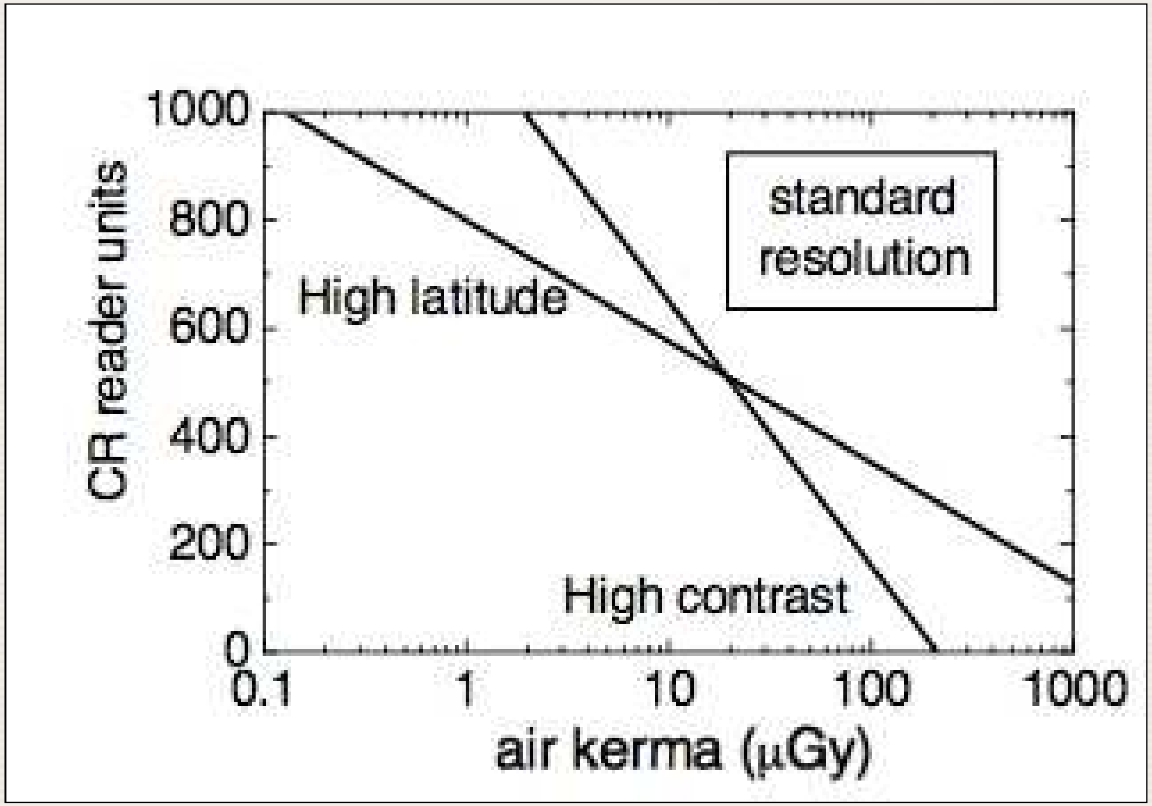


Image Readout

The **Quantum Accounting Diagram** for scanned readout of the flying spot laser readout of CR

shows that the incident absorbed X ray although originally represented by multiple quanta is at a critical stage in the chain represented by only **~5** electrons in the PMT

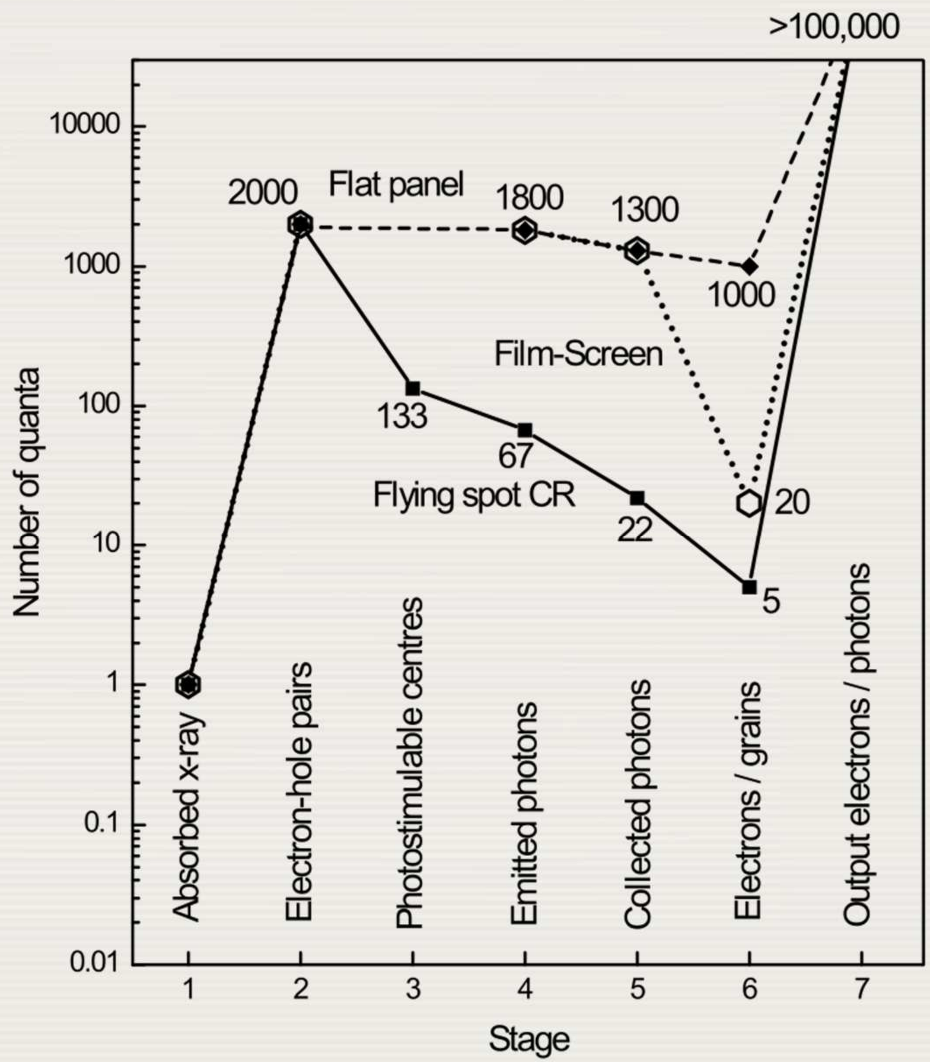


Image Readout

This is not adequate to keep the additional noise from the system negligible when the relatively poor MTF is also factored in

which can be seen in the **NPS** for CR:

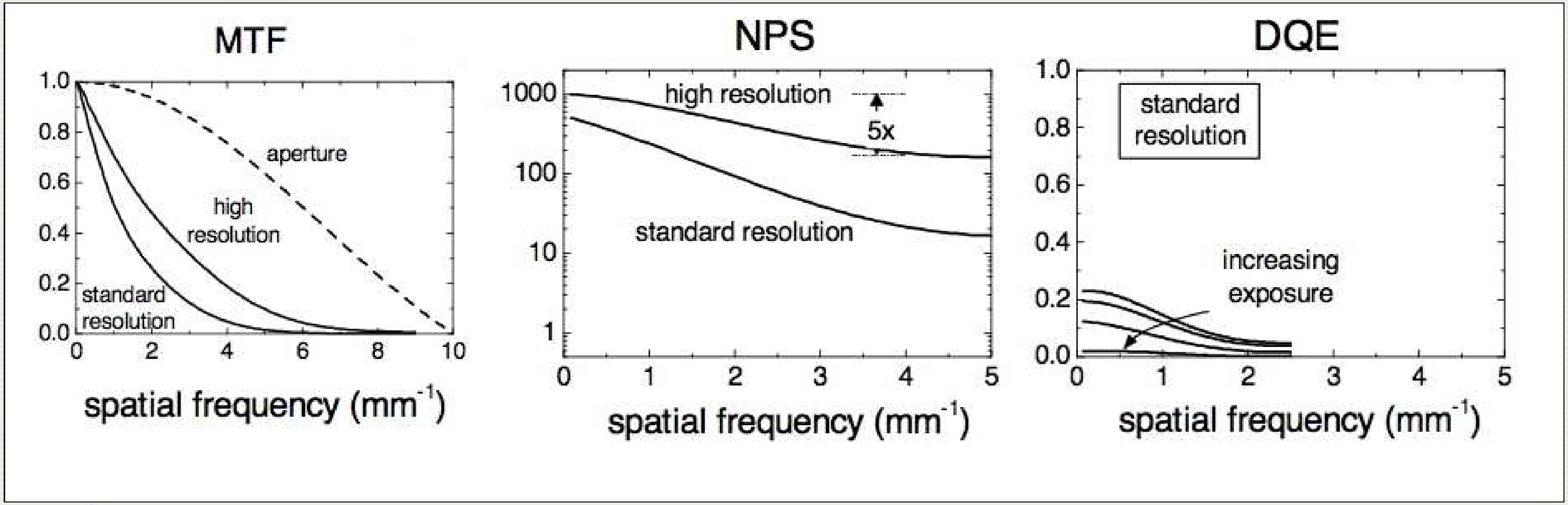


Image Properties

The underlying physical phenomenon of the **Photoluminescence** process is expected to be linear and this is demonstrated over the exposure range relevant for medical imaging

The variation of the signal with exposure is essentially linear over the **four orders of magnitude** relevant for diagnostic radiography

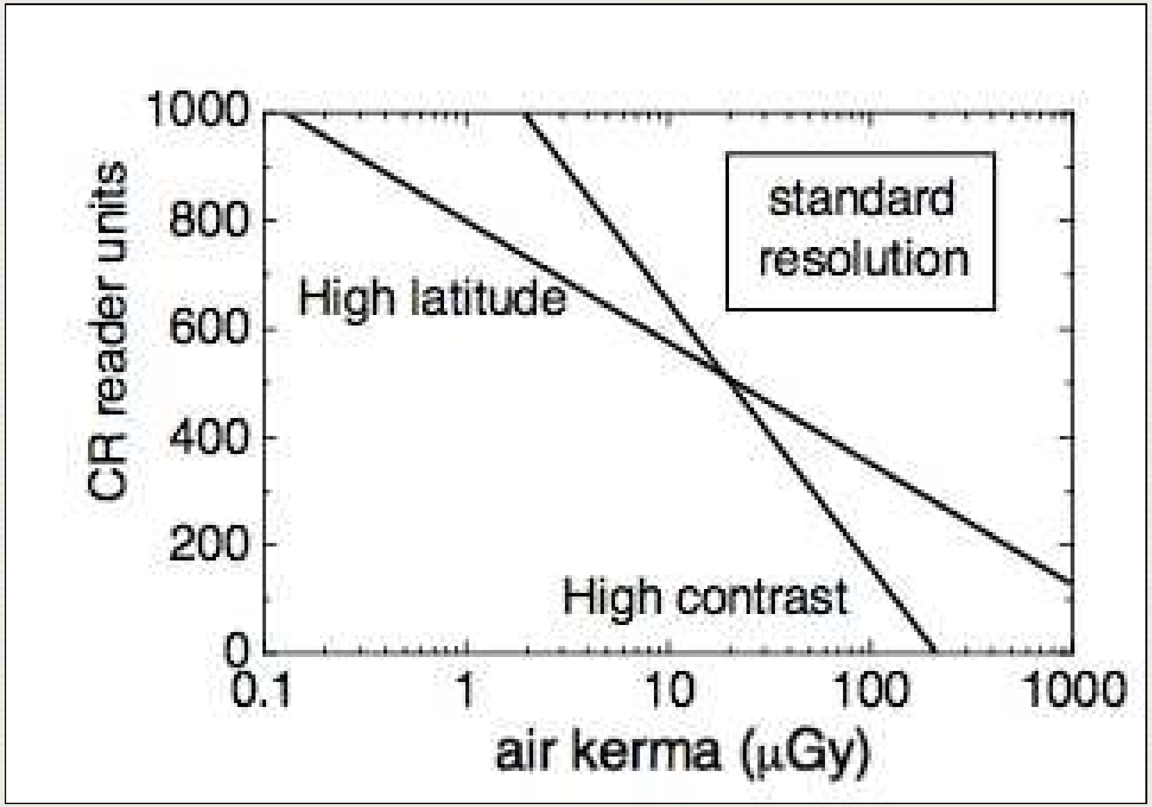


Image Properties

In practice, the **Photomultiplier** signal in most CR readers is processed **logarithmically** before being output

It appears at first sight that the **Characteristic Curve** for CR is radically different from all other imaging systems

This is due to the fact that images using both film-screen and CR are designed for radiography and it is conventional to use negative images (sometimes called **White Bone**)

Image Properties

The **conventional** mode of displaying the H&D curve hides this

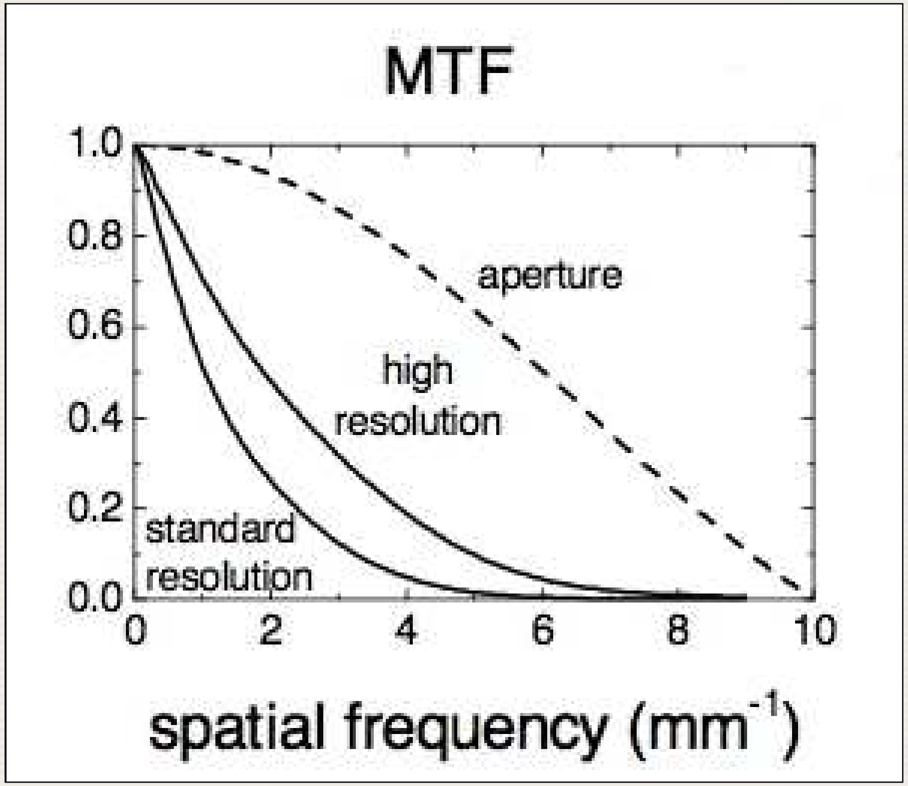
Depending on the **application** and the **plate type** used the relationship is different but consistent between settings

Various **Latitude** and **Speed** points can be set to mimic the behaviour of screen-film systems

Image Properties

The MTF is shown for high resolution and normal resolution IPs:

It can be seen that they are comparable to those for corresponding screen-film systems



This is not surprising as they are based on similar concepts for screens, and manufacturers can make any resolution simply by changing the **thickness**

The real test of equivalence will only come when we compare **DQE**

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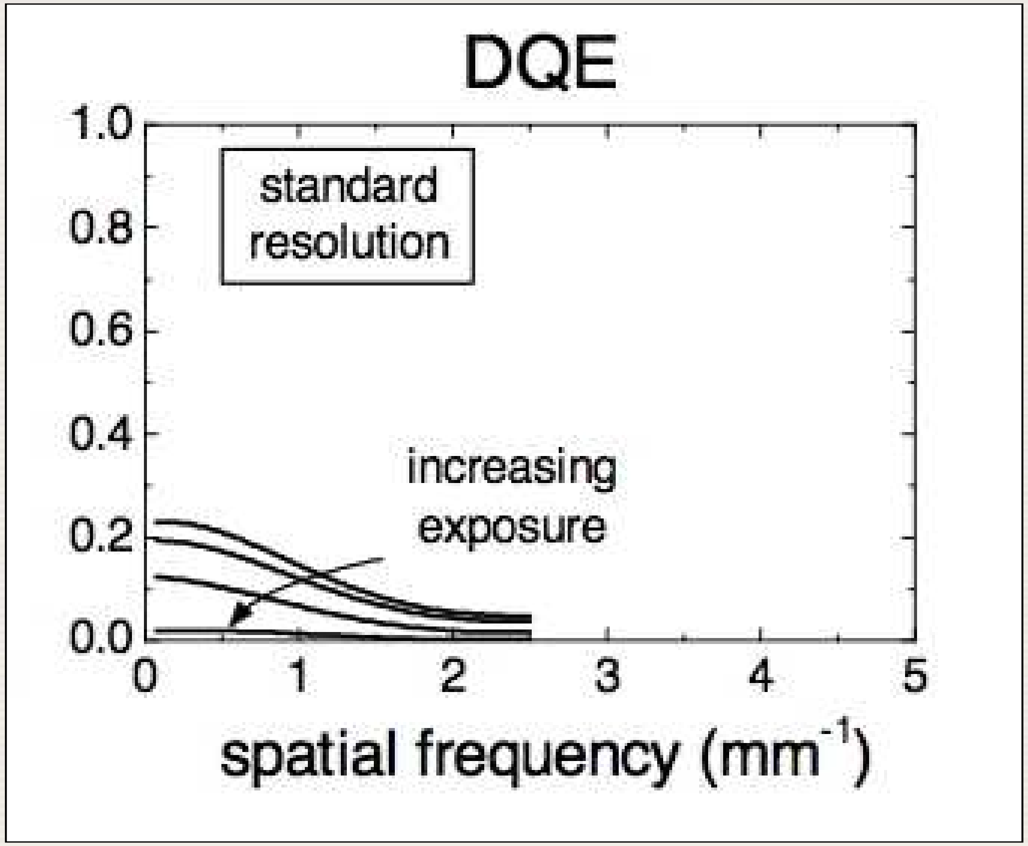
Image Properties

The overall image quality for CR can be evaluated from the **DQE** for a normal resolution plate

It is however interesting to see how closely the screen-film systems match the results for CR



This is related to the high level of secondary quantum noise evident in the **NPS** of CR, worse in fact than for screen-film, but not unexpected when the **Quantum Accounting Diagrams** for the two modalities are compared:

that both have Secondary Quantum Sinks in the range 5-20 quanta per X ray

Image Properties

All screens are made in similar ways and it is **impossible** to make them completely uniform in properties despite best efforts

The image from an entirely uniform exposure will therefore contain **as well as** Quantum Noise, noise due to the

- **Non-Uniformity** and
- **Granularity**

of the underlying structure

Image Properties

How does this manifest itself in images?

Suppose there was a 1% variation in effective thickness of the plate

then all images would reflect that change giving an effective signal-to-noise ratio of 1%

On the other hand, the SNR due to **Quantum Mottle** decreases as the square root of exposure

Thus, the **apparently surprising** conclusion that structural noise will be most important when the exposure is high

Image Properties

The **same effect** will be present in Screen-Film systems, although here variations in screen uniformity and film granularity will both be involved

For Screen-Film systems it is **impossible** to correct for structure noise, it could perhaps be accomplished for CR if efforts were made to ensure that the IP was always positioned exactly within the reader

However, this is not done and is **probably unnecessary** as the loss of DQE due to structural noise occurs at relatively high exposure levels, generally **outside** the clinically important regions of the image

The key digital technology permitting an advance in medical X ray applications is the

Flat-Panel Active Matrix Array

originally developed for laptop computer displays

The unique technology underlying active matrix flat-panel displays is large area integrated circuits called **Active Matrix Arrays** because they include an **Active** switching device - the **Thin Film Transistor** (TFT)

Active matrix arrays are an example of an important class of readout methods which are called **Self Scanned**

A self-scanned readout structure may be defined as one in which the image created in a certain plane is readout in that **same plane**

The advantage of such structures is that they are **thin** in the **third** dimension

Active matrix arrays use hydrogenated amorphous silicon as the semiconductors deposited on a thin (**~0.7 mm**) glass substrate which facilitate large area manufacture

The current scale on which the **Substrate** is made exceeds several metres on a side, thus permitting several sensors to be made on a single substrate which facilitates mass production of monolithic devices

Although long promised, it is **not yet feasible** to make all the readout components on the glass substrate

Gate drivers, readout amplifiers and ADCs for example are still made **separately** and bonded to the substrate

A complete **imaging system on glass** remains elusive

Another large **class** of self scanned receptors:

- **Charge Coupled Devices** (CCDs) and
- **CMOS** sensors

Although used extensively in optical systems incorporating demagnification

such as those for XIIs

they need to be made from single crystal silicon

Active matrix technology allows the deposition of semiconductors

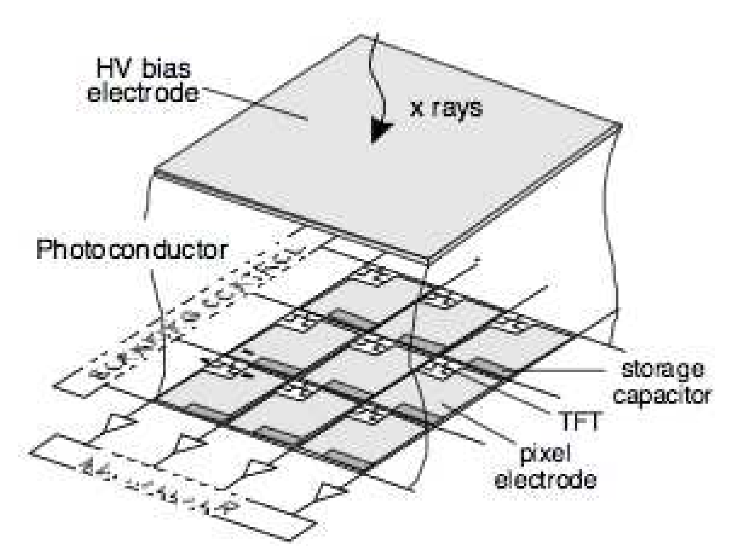
across **large area** substrates in a **well-controlled** fashion

such that the physical and electrical properties of the resulting structures can be **modified** and **adapted** for many different applications

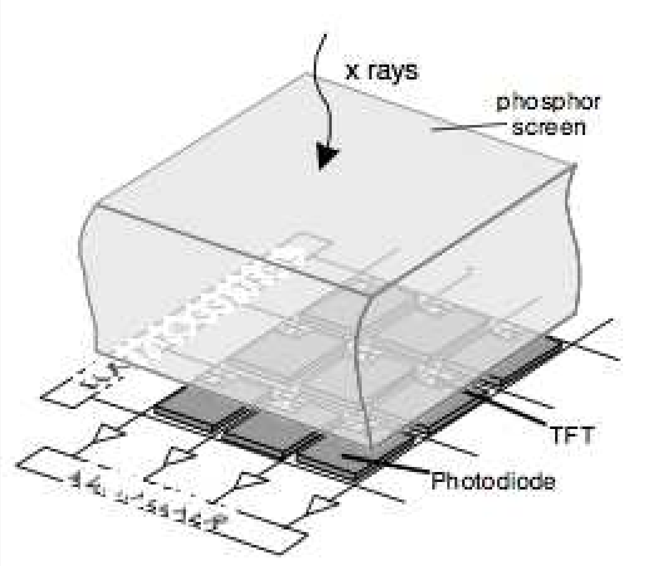
Coupling traditional X ray detection materials such as **Phosphors** or **Photoconductors** with a large-area active matrix readout structure forms the basis of flat-panel X ray imagers

7.4.3 Digital Radiography - DR

Large area active matrix array concept which is applicable for both
(a) **Direct** conversion (using a photoconductor layer) and
(b) **Indirect** conversion systems (using a phosphor layer)
depending on whether a pixel electrode or photodiode is used on the active matrix array



Direct conversion



Indirect conversion

In both cases thin film transistors (TFTs) are made using the semiconductor hydrogenated amorphous silicon (a-Si:H)

All the switches along a particular row are connected together with a single control line (**Gate Line**)

This allows the external circuitry to change the state of all the switching elements along the row with a single controlling voltage

Each row of pixels requires a separate switching **Control Line**

The signal outputs of the pixels down a particular column are connected to a single **Data Line** with its own readout amplifier

This configuration allows the imager to be readout **one horizontal line** at a time

7.4 DIGITAL RECEPTORS

7.4.3 Digital Radiography - DR

Unlike the **Charge Transfer Readout** method used in many modern CCDs

active matrix arrays **do not** transfer signal from pixel to neighbouring pixel

but from the pixel element **directly** to the readout amplifier via the Data Line

This is similar to **CMOS** imaging devices

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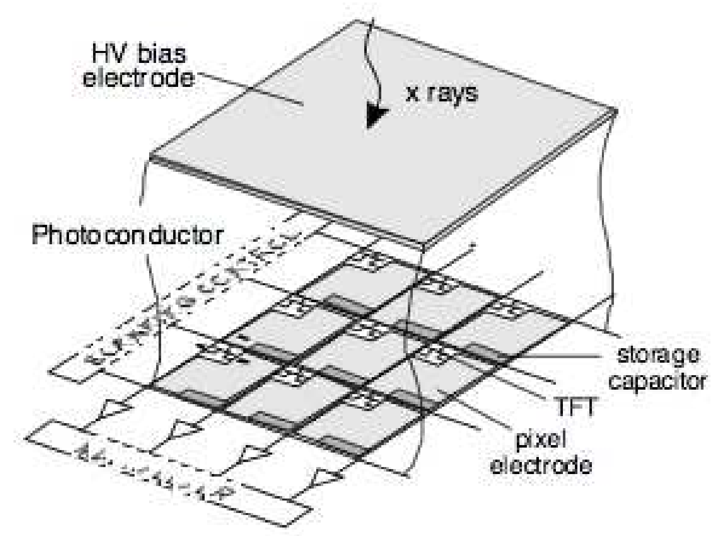
A distinction is made between flat-panel X ray imaging devices that:

- incorporate a photoconductor to produce electrical charges on detection of an X ray (**Direct Conversion**)
- those that use a phosphor to produce visible wavelength photons on detection of an X ray (**Indirect Conversion**)

In the **Direct Conversion** approach, a photoconductor is directly evaporated onto an active matrix array

The charge released in the bulk of the photoconductor is collected by a large applied field, which brings the electrons and holes to their respective **Electrodes**

Those reaching the upper continuously biased electrode are neutralized by the power supply providing the bias, effectively replenishing the loss of field which would otherwise be caused



Direct conversion

The **Charge** reaching the readout pixel electrode is stored temporarily on the **Capacitance** of the pixel until readout

The **Magnitude** of the signal charge from the different pixels contains the imaging information inherent in the intensity variations of the incident X ray beam

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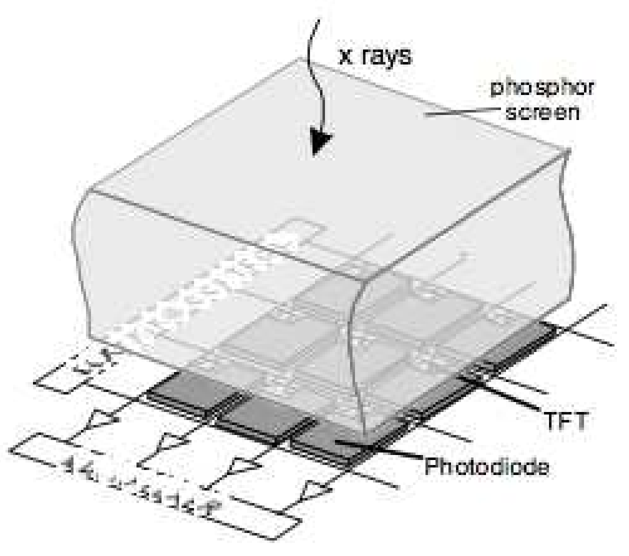
In the **Indirect Conversion** approach, a phosphor layer (e.g. a structured scintillator such as CsI:Tl) is placed in intimate contact with an active matrix array

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 receptor 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



Indirect conversion

This charge is **stored** in the pixel until the active matrix array is read out

There are **in principle** two advantages of direct conversion compared to indirect:

1) the fewer number of **Conversion Stages** make it possible to have a significantly higher conversion efficiency of X ray energy to EHPs on the active matrix

~8,000 direct ***cf*** 1,000-2,000 indirect

2) Much higher resolution due to the elimination of blurring during the charge (direct) or photon (indirect) **Collection Phase**

Direct Conversion

Proposed photoconductors include Hg_2I, CdTe and PbO

However, there is only one currently **practical** photoconductor, **a-Se**

Its **Conversion Efficiency** is in fact comparable to phosphors, and technical issues limit its thickness

so it is used mostly in **Mammography** where its low *Z* is in fact ideally matched to the absorption requirements and high resolution is attained

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Thus at the current time the dominant approaches to digital **General Radiography** use the indirect conversion approach due to the higher specific absorption of available phosphors compared to a-Se

In **Mammography** however, a-Se is superior due to the need for a low energy X ray spectrum increasing the quantum efficiency beyond what is possible with a phosphor and because it simultaneously increases resolution

Indirect Conversion

One of the main issues with the design of **Powdered Phosphor** screens is the balance between spatial resolution and X ray detection efficiency

As the phosphor is made thicker to absorb more X rays, the emitted light can spread further from the point of production before exiting the screen

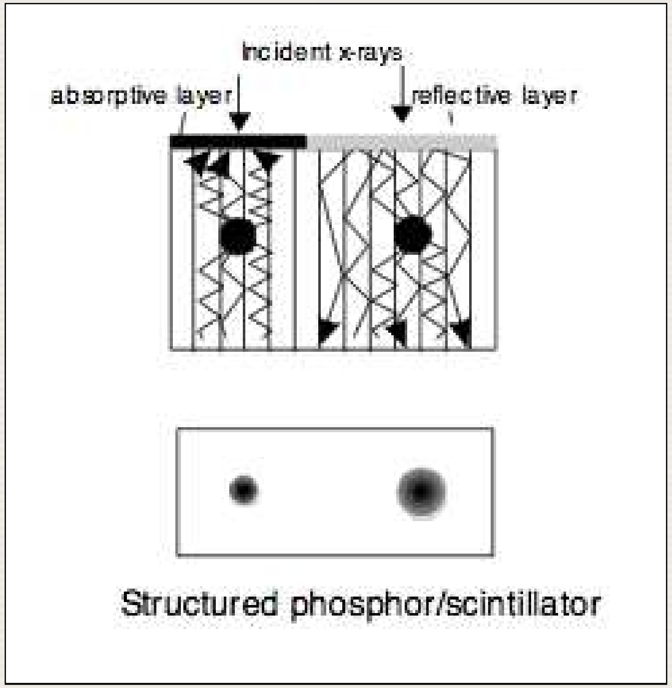
This conflict is significantly eased by the use of a **Structured Phosphor** such as CsI

When evaporated under the correct conditions, a layer of CsI will condense in the form of **needle-like**, closely packed crystallites

In this form the resolution is **better** than for a powder phosphor screen

However, resolution may be further enhanced by **fracturing** into thin pillar-like structures by exposure to a thermal shock

This has the disadvantage of reducing the effective density of the structured layer to ~80-90% that of a single CsI crystal



The hope was that these columns would act like fibre optic light guides due to the difference in refractive index **n** between:

- CsI (**n** = 1.78) and the
- Inert gas or air (**n**~1) which fills the gaps between the pillars

Taking this model literally, light photons produced by the absorption of an incident X ray will be guided towards either end of the pillar if they are emitted within the range of angles that satisfy conditions for **Total Internal Reflection**

Theoretical calculations predict that **~83%** of the isotropically emitted light will undergo internal reflection within a perfectly uniform pillar

The other **~17%** will scatter between pillars and cause a reduction in the spatial resolution

Actual layers of CsI have a somewhat **reduced** light collimating capability due to the:

- Unavoidable non-uniformity of the surface of the pillars
- Unavoidable contacts between adjacent pillars at various points within the layer and
- Defects in the cracking

In addition, in practice the layers form with an initial layer which is not columnar and the columns develop beyond a certain thickness (**~50 μm**)

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However, they maintain significantly **higher resolutions** for a given thickness of phosphor than powder screens

As a **Rule of Thumb** they seem to have **twice** the resolution of a powder phosphor screen of the same physical thickness and optical design

e.g. presence or absence of reflective backing

This corresponds to 3-4 times the **Mass Loading** for the structured phosphor layer when packing density and atomic number are accounted for

To increase the light collection capabilities of the layer, a **Reflective Backing** can also be added to the X ray entrance surface of the CsI to:

redirect the light photons emitted in this direction back towards the exit surface

This significantly increases the light output of the layer but at the cost of a reduced spatial resolution

MTF, NPS & DQE of DR Systems

The MTFs of the direct and indirect systems show a distinct **qualitative difference**

The overall MTF of each system is the product of the MTF of the X ray detection medium and the del aperture function which for a uniform response is the **sinc function**

The intrinsic spatial resolution of a photoconductor is extremely high, which means that the system MTF for a direct conversion receptor will be to all intents and purposes the sinc function as all other components are close to unity

MTF, NPS & DQE of DR Systems

In a system that utilizes a phosphor as the X ray detection medium, the spatial response of the phosphor is much poorer than the sinc function

The MTF of the combination will then be practically equivalent to the phosphor MTF

In other words for the a-Se system it is the **Aperture Function** which defines the **Presampling MTF** whereas for phosphor based system (even columnar CsI) the **phosphor blurring** dominates and defines the overall MTF

MTF, NPS & DQE of DR Systems

Furthermore, for the parameters usually chosen in practical medical systems the MTF at the $\mathbf{f_N}$ is:

- ~60% for the **Direct** system and
- closer to 10% for the **Indirect** system

This implies that noise aliasing will be severe for the direct

and

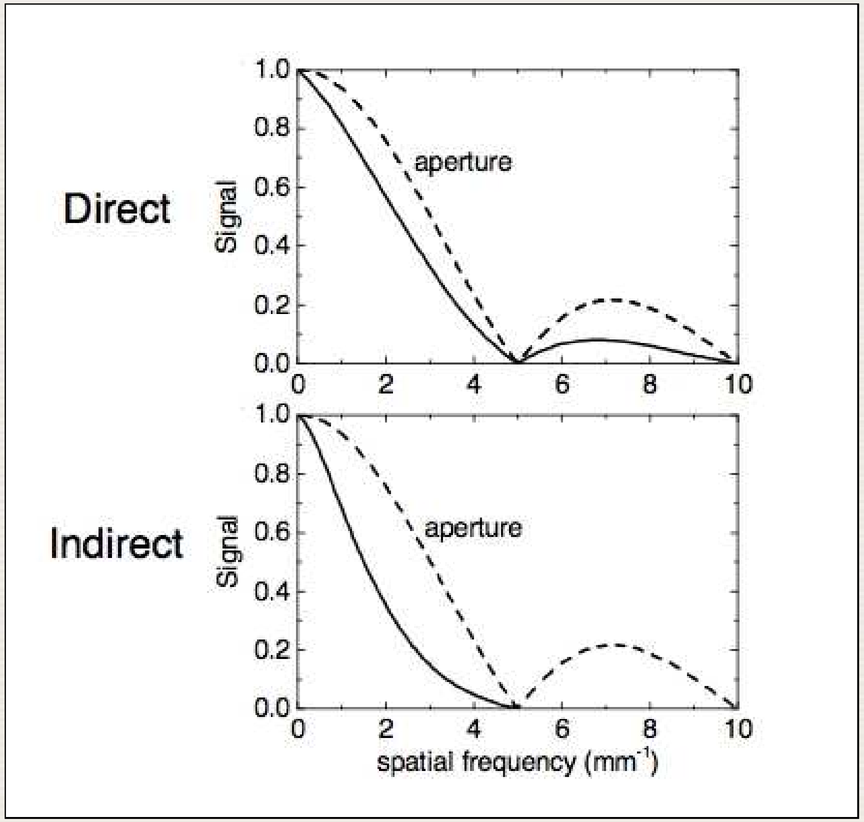
almost negligible for the indirect detection approach

in general

MTF, NPS & DQE of DR Systems

The degree of **Aliasing** permitted by the system designer can be established from an evaluation of the pre-sampled MTF and a calculation of the area above $\mathbf{f_N}$ compared to the area below

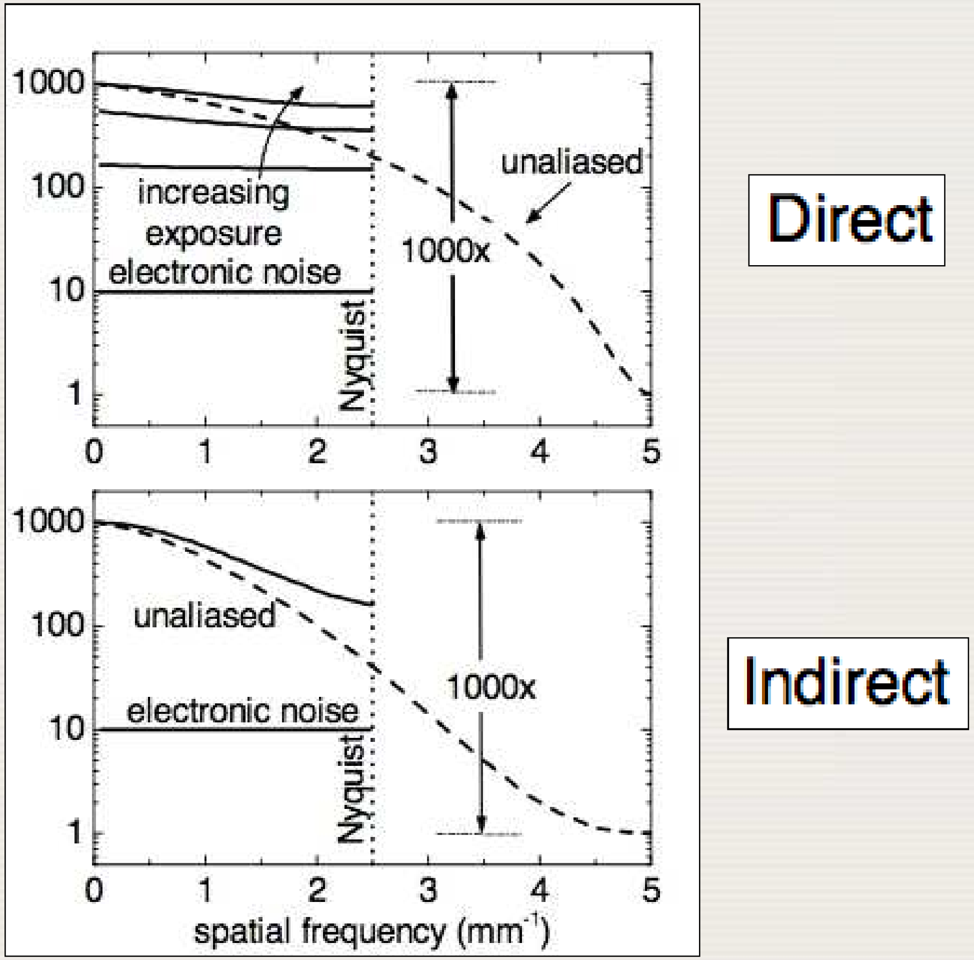
Examples of MTFs for well-matched systems using 200 µm pixels and 200 µm sampling pitch for both **Direct** and **Indirect** conversion systems at the same incident energy



MTF, NPS & DQE of DR Systems

The NPS of **Direct** and **Indirect** receptors also demonstrate striking differences

The NPS of the **Direct** receptor, due to its minimal spatial filtration prior to sampling, combined with aliasing of the frequencies above the $\mathbf{f_N}$ is almost **white** (i.e. is independent of spatial frequency)



MTF, NPS & DQE of DR Systems

In contrast the **Indirect** receptor shows a **marked drop** in NPS with increasing frequency, due to the greater presampling blurring inherent in the phosphor layer demonstrated by comparing the MTFs

The NPS for higher frequencies show a **Linear Exposure Dependence** but the NPS does not go to zero at zero exposure - this is the **Electronic Noise** which at the pixel level is of the order of several thousand electrons rms

This noise is to be compared with the signal per X ray which is of the order of **1,000**

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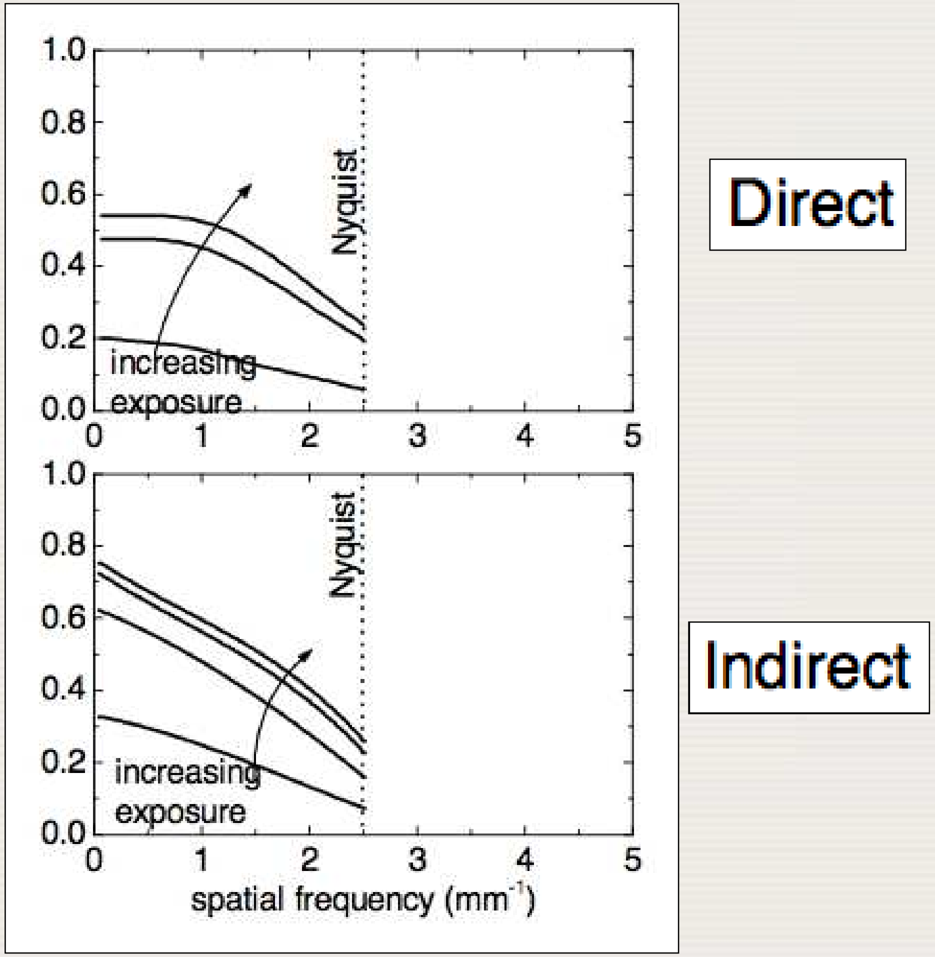
MTF, NPS & DQE of DR Systems

Finally by combining MTF and NPS the DQE(**f**) can be obtained:

In the particular **Direct** conversion receptor illustrated the DQE(0) is somewhat smaller due to the relatively poor X ray absorption efficiency of the photoconductor whereas DQE drops very little with spatial frequency due to the very high MTF (which is close to the ideal sinc function)



MTF, NPS & DQE of DR Systems

In contrast the DQE(0) of the **Indirect** conversion system is higher due to better X ray Absorption Efficiency

but the DQE drops more rapidly with increasing spatial frequency

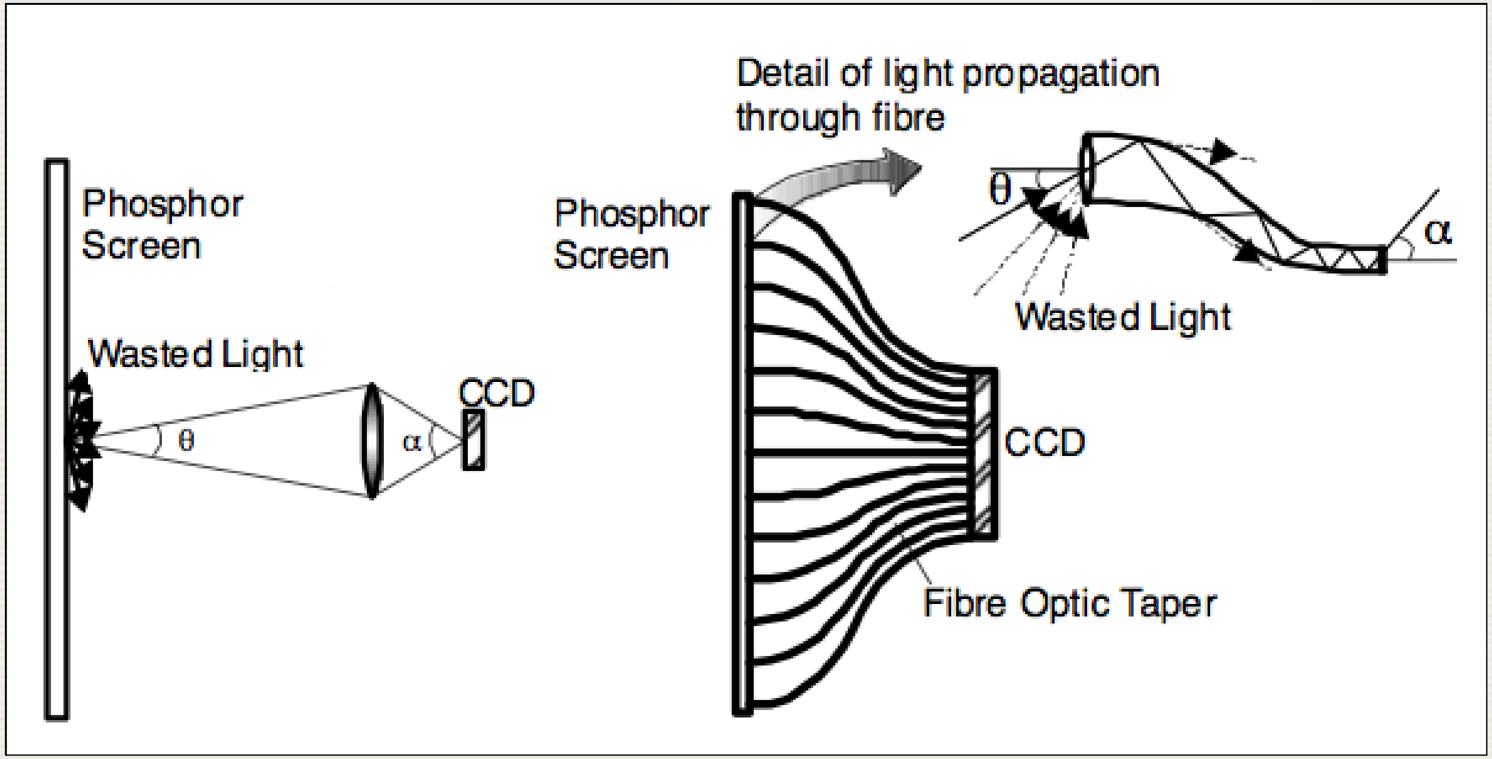
due to the poorer MTF

which is mitigated to some degree by the reduction of NPS with frequency due to the drop in MTF

Optically Coupled Systems

Indirect conversion flat panel systems use **Optical Coupling** of an imaging screen to an active matrix array

Earlier systems used smaller receptors arrays in conjunction with a fibre optic or a lens to couple the image to other optical devices, such as a CCD or a video camera



Lens Coupling Fibre Optic Coupling

Optically Coupled Systems

However, there is then a considerable loss of light depending on the **Collection Angle** of the optical system compared to the broad angle over which light is emitted from screens

These methods therefore have significant problems in maintaining **good noise properties**

This is because the **Collection Efficiency** of the light from the screen by the imaging device is generally rather poor

Optically Coupled Systems

For example, even in the best optically coupled lens system with a pair of **Relay Lenses** of the same focal length placed back to back, the coupling efficiency cannot practically exceed **20%** and this only with very high aperture lenses (f/0.75)

This is the case for 1:1 imaging

With demagnification **M** greater than unity, the coupling efficiency drops by $\mathbf{M^2}$

M: ratio of the side of the screen to the corresponding image sensor dimension

Optically Coupled Systems

Thus for commonly seen demagnifications of **20** the coupling efficiency is **0.05%**

Under these conditions only **~1** light photon on average represents the interaction of each X ray

This is a serious **Secondary Quantum Sink**

Optically Coupled Systems

This may manifest itself in **two** general ways:

1. The inability to fully represent the position of the X ray resulting in a decrease in both **DQE(0)** and a much more rapid decrease in **DQE** with **f** and

2. Due to the very small gain resulting in amplifier noise becoming dominant at much higher exposure levels than it would otherwise

Optically Coupled Systems

In the presence of these kinds of **Secondary Quantum Sinks** the transfer of light from the screen to the receptor becomes the limiting stage in the imaging chain and

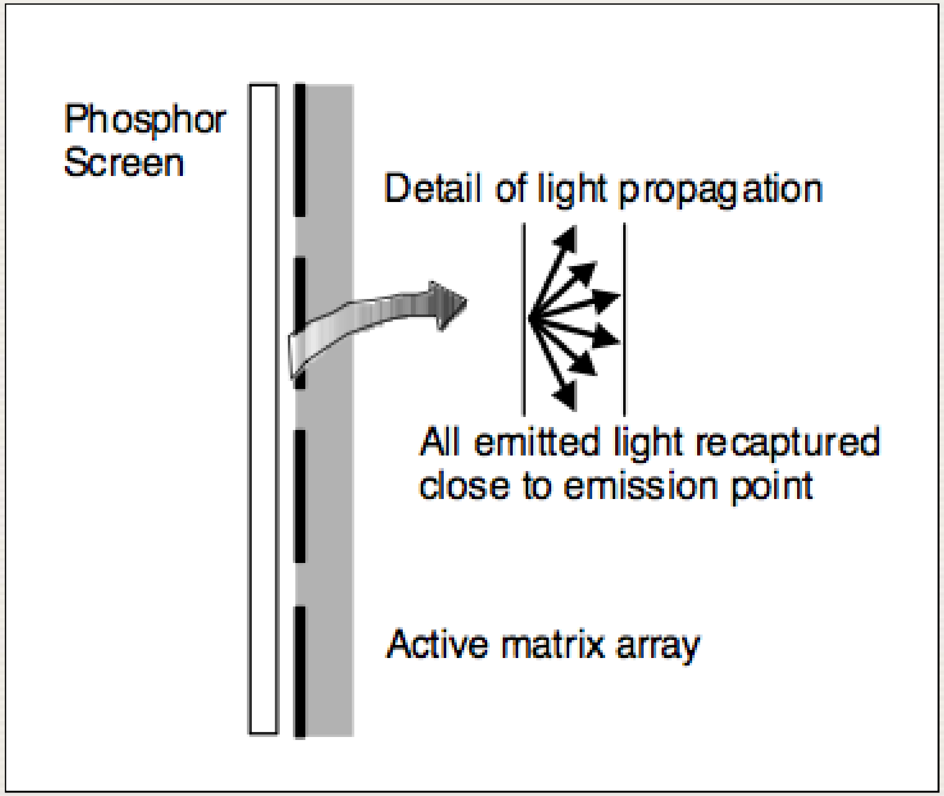
significantly degrades the overall performance of the system

The use of a **Fibre Optic Taper** can alleviate, but not eliminate the loss due to demagnification for the reason that the acceptance angle for light decreases in the same way as for lenses

Optically Coupled Systems

Thus a major technical advantage for the flat panel receptor (in an **Indirect** detection configuration) is that it can be placed in **Direct** contact with the emission surface of the screen

Its **Collection Efficiency** for the emitted light is consequently much higher (~50% and approaches 100% for special configurations) than with the demagnifying approaches



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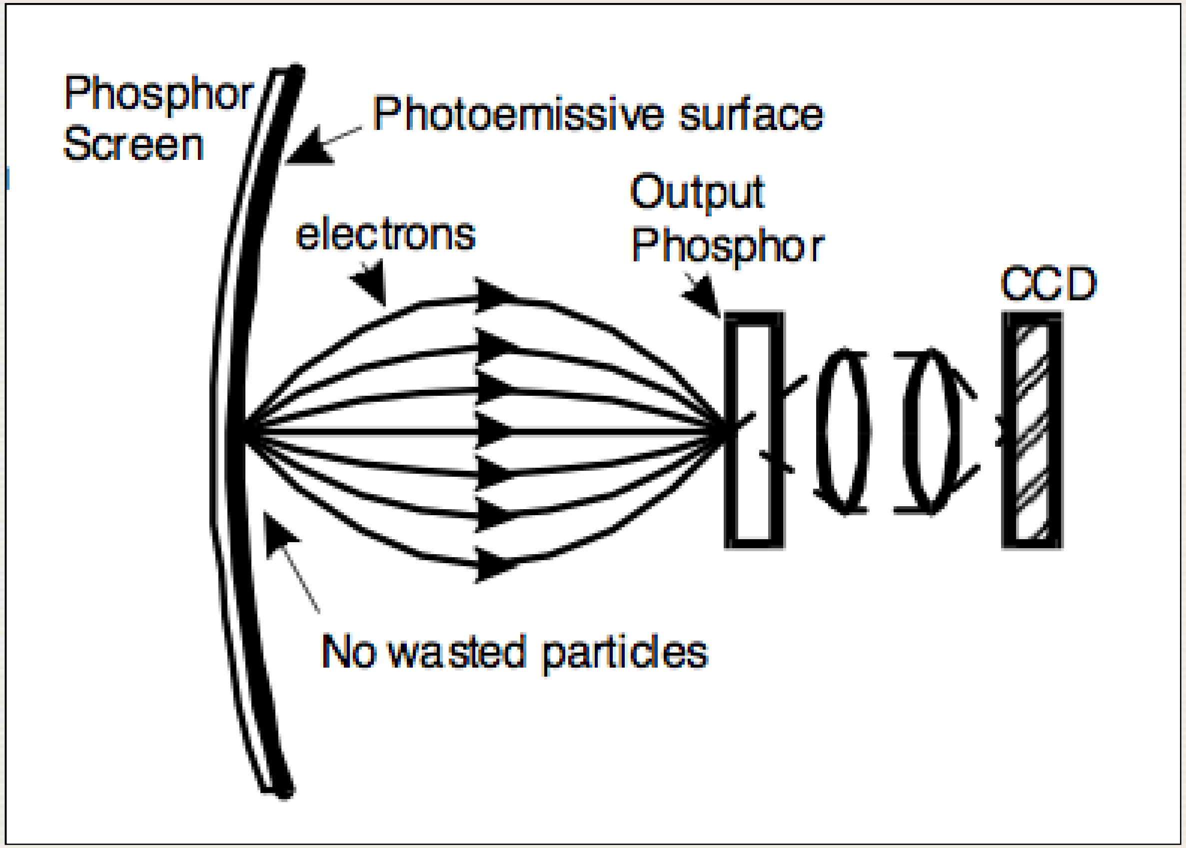
Optically Coupled Systems

It is interesting to compare the situation for an XRII:

Electron optical demagnification in an XRII



Optically Coupled Systems

Here

- by converting light photons from the Phosphor to electrons in the Photocathode in **direct** contact with the phosphor and
- the ability to bend the path of electrons, which is impossible for light

is critical in maintaining much higher Collection Efficiency and so avoiding a **Secondary Quantum Sink**

This is a critical reason why XRIIs were important for so long despite their other problems

Photon Counting

Photon Counting receptors and their close relatives **Photon Energy Discriminating** receptors interact with incident photons one by one and report back to the system that:

- a photon has been detected (a **Count**) or
- a photon within a specific energy range has been detected (**Energy Discriminated Count**)

The potential advantage in image quality of these systems is **significant**

Photon Counting

There are **two** reasons:

The **First** is that they can entirely eliminate the effect of amplifier noise

This is because the signal from a single X ray emerges in a very short time and the signal can thus be made to be **>5** times the noise

This advantage makes it possible to work at **very low exposure rates** where other systems would be dominated by amplifier noise

Photon Counting

The **Second** reason is that by knowing the size of the signal, the energy of the incident x ray may be estimated, and thus correction for **Swank noise** performed

And, in the context of the complete system, better **weighting** of the importance of:

- High Energy (highly penetrating, low contrast) versus
- Lower Energy (less penetrating, higher contrast)

X rays can be accomplished

Photon Counting

This can perhaps increase the SNR by factors of **~2**

Unfortunately these improvements pale in comparison to the increase of complexity of the circuitry necessary at each pixel, which may generally be of the order of **1,000-100,000** fold

This practical factor has to this point limited the application to **Mammography**

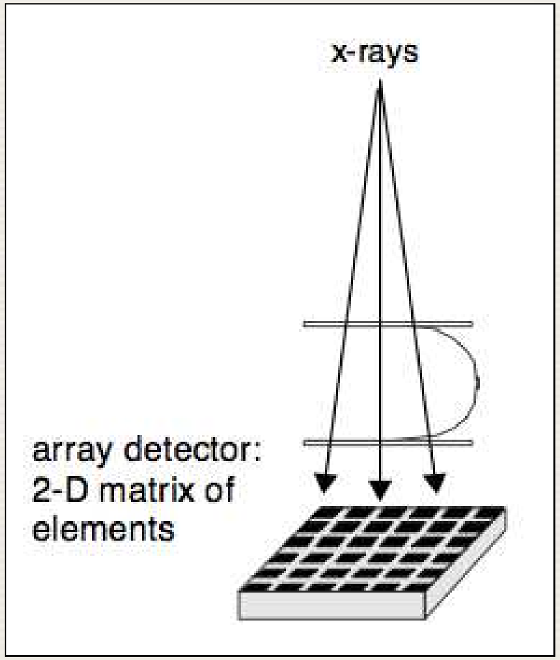
but improvements in microelectronic circuitry are reaching the point where more general application may be feasible in the near future

Scanning Geometries

What X ray geometry should be used?

is the first fundamental decision facing the designer of an X ray imaging system

Conventionally, producing an X ray image involves exposing the entire area of interest simultaneously and detecting it with an area sensor as in screen-film radiography



Scanning Geometries

Other approaches

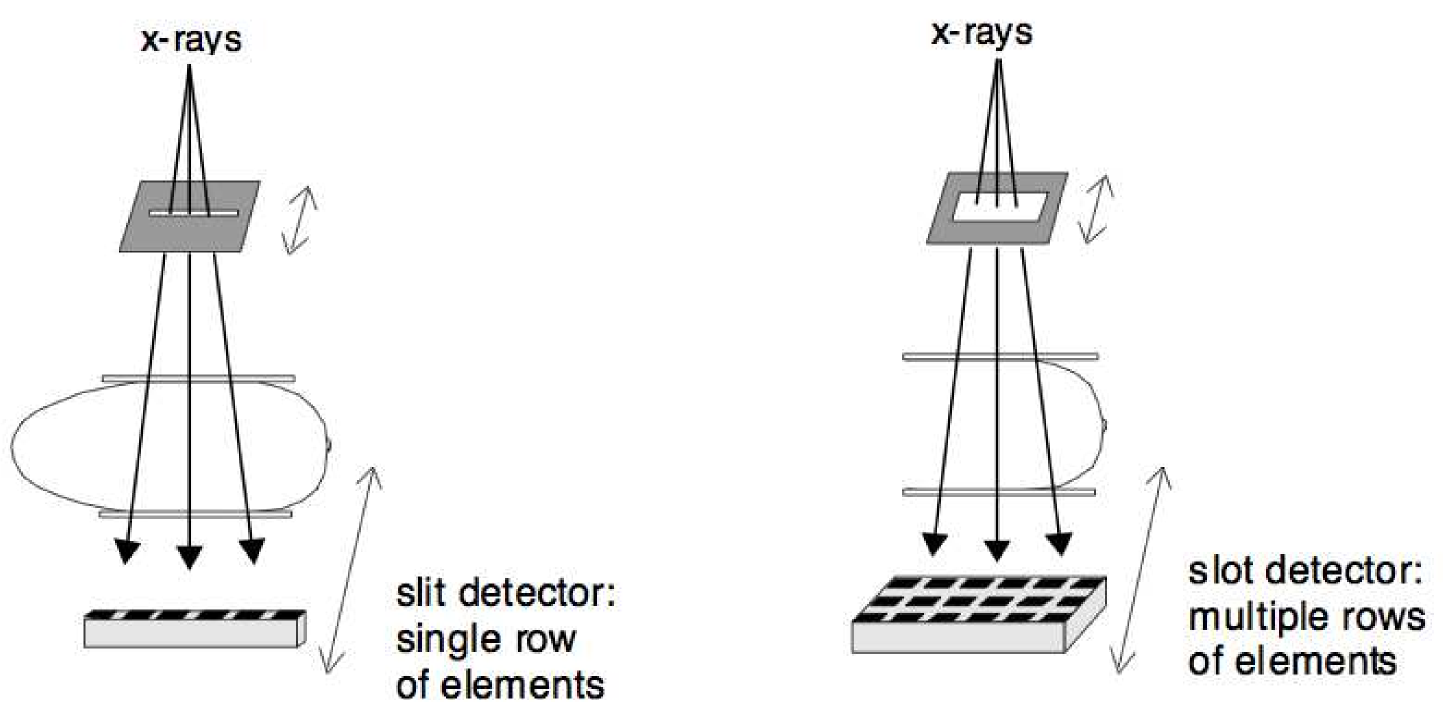
The **simplest** is to obtain a **Pencil Beam** of radiation and scanning it over the patient, one point at a time

pencil beam accomplished by collimating the broad area flux from the XRT by means of a lead blocker with a small hole in it

A single sensor aligned with the pencil beam creates an image of the patient

Scanning Geometries

Other variations between a pencil and area beams:



(a) Slit beam

(b) Slot beam

Other approaches

Scanning Geometries

Slit Irradiation is obtained with a fan beam of radiation and an aligned single line receptor which is scanned perpendicularly to the line across the patient

Both pencil beam and slit beam scanning are extremely **inefficient** in the utilization of X rays

Most of the X rays are removed by the collimator and a full scan imposes an enormous **Heat Load** on the tube

Scanning Geometries

Other approaches

It is possible to improve the efficiency of such systems by employing a multi-line or

Slot Receptor

where the X ray beam extends across the **full** image field in one dimension and

is several lines wide in the other

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Other approaches

Scanning Geometries

There are **two** types of Slot receptors:

1. A slot is moved discontinuously across the width, a single exposure made and the **multiple lines** readout - this process is repeated until the entire area is covered

2. In **Time Domain Integration**, TDI, the slot beam receptor moves continuously and the image is read-out one line at a time

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Scanning Geometries

Why use these **complicated** scanning methods to produce images when it appears that irradiating a static area receptor is much simpler?

Several concepts must be balanced:

- Simplicity of Construction
- Scatter
- Tube Loading

History has shown that when **Area Receptors** are feasible they are preferred

Scanning Geometries

The first is relative **Simplicity of Construction**

In an X ray scanning system to image patients there must be no significant wasted radiation, which means that very accurate pre-collimation of the X ray beam must be used

This is made more difficult by the requirements of scanning the system

In the **early** development of digital radiographic systems it was only technically feasible to create **Linear Receptor Arrays** and no practical area arrays existed, thus the mechanical complexities were acceptable since there were no alternatives

Scanning Geometries

Image Quality

Each method has image quality advantages and disadvantages

but the most important consideration is

Scattered Radiation

A reduced area receptor can, be much more efficient than area receptors in **eliminating** scatter

Scanning Geometries

Tube Loading

The shortest exposure and the least loading of the tube

are huge strengths of the area receptor which make it the preferred option

unless control of **Scatter** is paramount

Scanning Geometries

Existing systems which use scanning systems are highly specialized devices where **Scatter Control**:

1) to ensure quantitative imaging is essential

For example

Dual-Energy Imaging for Bone Densitometry

where the complete and exact elimination of scatter overwhelms other concerns

Scanning Geometries

2) to permit **Photon Counting** approaches

where the technical demands of the additional counters and discriminators needed at a **per pixel** basis are currently **prohibitive** for an area receptor

but are **feasible** for example by using several silicon receptors in an edge-on configuration

The raw image information acquired from the current generation of flat-panel receptor systems is unsuitable for immediate image display

It must be processed to remove a number of **Artefacts** to obtain a diagnostic quality radiograph

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Moiré

A particularly visually disturbing effect, which is to be avoided at all costs, is **Moiré Fringing**

arising from spatial interference between the periodic structure of flat panel receptors and a **Stationary Grid**

Moving the grid perpendicularly to the grid lines during the exposure using a **Potter-Bucky** grid arrangement should eliminate these problems

Ghosting & Lag

Corrections for:

- Image carry over or **Lag** - effects seen in dark field exposure after prior exposure or
- **Ghosting** - effects producing change in gain related to prior images and so are seen in flood field images

may sometimes be necessary

Ghosting & Lag

These phenomena may be particularly problematic

- After large exposures to the imager or
- When the imager is used in **Mixed Mode** (i.e. receptor designed to be capable of both fluoroscopic and radiographic imaging)

and the system is moved to **Fluoroscopy** after a large **Radiographic** exposure

Advantages of Digital over Analogue Systems for Radiography

Advantages related to Image Quality and Dose:

- Lower dose needed
- Higher resolution
- Greater dynamic range

Advantages related to **Convenience** in use:

- Elimination of handling and carrying of cassettes
- Immediate evaluation of images for image quality and positioning
- Transmission of digital images
- Digital archiving, searching PACS
- Elimination of unique image (film)
- Image processing to more optimally present the image information to the reader
- Elimination of distortion and shading (c.f. XRIIs)
- Enabling advanced applications (e.g. digital tomosynthesis, cone beam CT, dual energy imaging and CAD)

The image quality of radiographic detectors has experienced a quantum jump in the last decades as **Flat Panel Imagers** have become feasible

However, there is still no completely satisfactory system, and the **cost** is very high compared to the systems they have replaced

There is still much to be done to provide **Quantum Limited** performance for all radiographic imaging receptors at reasonable cost

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