

Chapter 6: Projection Radiography

Slide set of 124 slides based on the chapter authored by
J. Poletti
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 a basic treatment of X ray
image formation and the effects of scattered radiation



IAEA
International Atomic Energy Agency

Slide set prepared
by K.P. Maher
following initial work by
S. Edyvean

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6.1 INTRODUCTION

In its simplest form X ray imaging is the collection of

Attenuation Shadows

that are projected from an ideal X ray point source onto an image receptor

This simple form is true for all X ray imaging modalities, including complex ones that involve source and receptor movement such as CT

6.1 INTRODUCTION

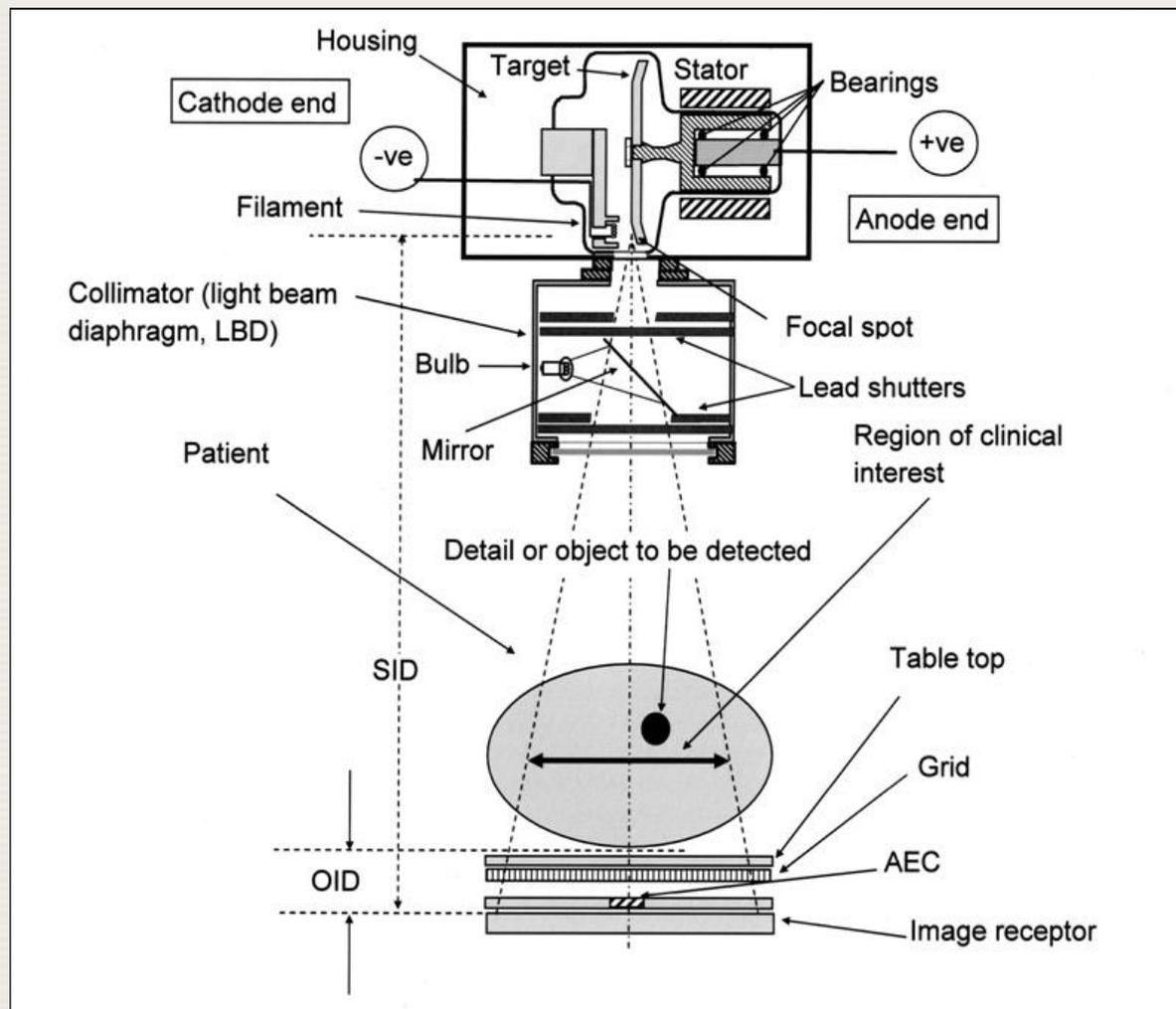
This simplified view however is made vastly more **complex** by:

- the **Non-Ideal** point source
- the consequences of **Projecting** a 3D object onto a 2D detector
- the presence of **Scattered Radiation**, generated within the patient, which will degrade any image that is captured

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

The **Principal** components of a system for X ray projection radiography:



Components such as shaped filtration, compression devices or restraining devices may be added for special cases

Grid and AEC are optional



6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

When considering the **Ideal** imaging task (the detection of a detail, against a uniform background) the ideal x-ray spectrum is **Monochromatic**, when the three constraints patient dose, image quality and XRT loading are considered

Any particular projection may consist of more than one such task, each with a different ideal monochromatic energy

The choice of **X Ray Spectrum** for each task is therefore always a compromise

So that the actual Bremsstrahlung and Characteristic Radiation spectrum provide the best **Approximation** to the ideal monochromatic spectrum for the particular projection

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

For an ideal imaging task:

$$\text{Contrast, } C = \Delta B/B$$

where **B** is the image brightness (or shade of gray) in a background region and ΔB is the difference in brightness for a small detail

For small values of ΔB , **Linearity** of brightness with X ray intensity (**I**) is assumed, so the contrast

$$C = \Delta I/I$$

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

This is generally valid for a particular monochromatic spectrum

For a **Real** polychromatic spectrum:

- An approximation can be a monochromatic spectrum with the **Average Energy** of the actual spectrum or
- The result can be **Integrated** over all spectral energies

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

The X ray intensity is related to thickness by the attenuation law, therefore the **Primary Contrast**:

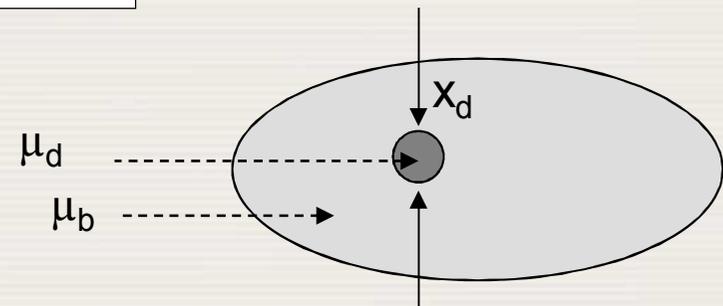
$$C_p = 1 - e^{-(\mu_d - \mu_b)x_d}$$

where:

x_d : thickness of the detail

μ_d : linear attenuation coefficient of the detail

μ_b : linear attenuation coefficient of the background material



6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

To find the **Average Contrast** for a particular detail, the equation should be integrated over the detail

For **Thin Spherical Details** (e.g. micro calcifications in mammography, solitary pulmonary nodule in chest x-rays), this is straightforward and results in a contrast which is **2/3** the contrast obtained for a ray passing through the centre of the detail

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

The relationship between the **Linear** Attenuation Coefficient and the **Mass** Attenuation Coefficient tells us that contrast exists for details that differ in:

Mass Attenuation Coefficient and/or **Density**

Contrast will depend on the **Thickness** of the detail not the thickness of surrounding tissue

Contrast is **Inversely** related to the kV setting

Since values of μ reduce as photon energy increases

Thus **kV** may be considered to be the contrast control

where contrast is strictly the detail contrast



6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

For **Screen-Film** imaging the difference in optical density due to the detail is proportional to the

Subject Contrast multiplied by the **Gamma**

of the screen-film system

For **Digital** image receptors, the relationship is more complex, since the contrast of the displayed image is **Adjustable** independently

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

Energy absorbed in a small region of the image receptor:

E_p : energy absorbed due to primary rays

E_s : energy absorbed due to secondary rays

Scatter may be quantified by:

Scatter Fraction $SF = E_s / (E_p + E_s)$

Scatter to Primary Ratio $SP = E_s / E_p$

The relationship between the two is $SF = ((SP^{-1}) + 1)^{-1}$

6.2 X-RAY IMAGE FORMATION

6.2.1 Components of an Imaging System

In the presence of scattered radiation, the **Primary Contrast** equation becomes

$$C_P = 1 - e^{-(\mu_d - \mu_b)x_d} \frac{1}{1 + SP}$$

Minimization of scatter is therefore important

6.2 X-RAY IMAGE FORMATION

6.2.2 Geometry of Projection Radiography

The **Primary Effect** of projection radiography is to record an image of a 3D object (the patient) in 2D, resulting in superposition of the anatomy along each ray

This leads to a number of effects that need to be considered in:

- the **Design** of equipment
- the **Production** of the images and
- their **Interpretation**

In particular, for each projection there will be a region of clinical interest, **Somewhere** between the entrance and exit surface of the region to be imaged

6.2 X-RAY IMAGE FORMATION

6.2.2 Geometry of Projection Radiography

Considerable training and experience is required for the **Radiographer** to correctly choose the geometrical variables to image this region, based on superficial visible or palpable landmarks

These variables include **Focus-to-Image Distance (FID)**, **Object-to-Image Distance (OID)**, projection direction (lateral, cranio-caudal, etc.) or **Angulation**, **Centring Point** and beam **Collimation** area

In some cases, the correct projection of **Joint Spaces** needs also to be considered

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Superposition

Leads to a significant loss of **Image Contrast**

This provided one of the prime motivations for the development of CT scanners

Also leads to a **Loss** of all depth information and **Ambiguity** in the relative sizes of objects at different depths

It directly **Overlays** objects in such a way that it can become difficult or impossible to distinguish one from the other, or even to identify some of the objects



6.2 X-RAY IMAGE FORMATION

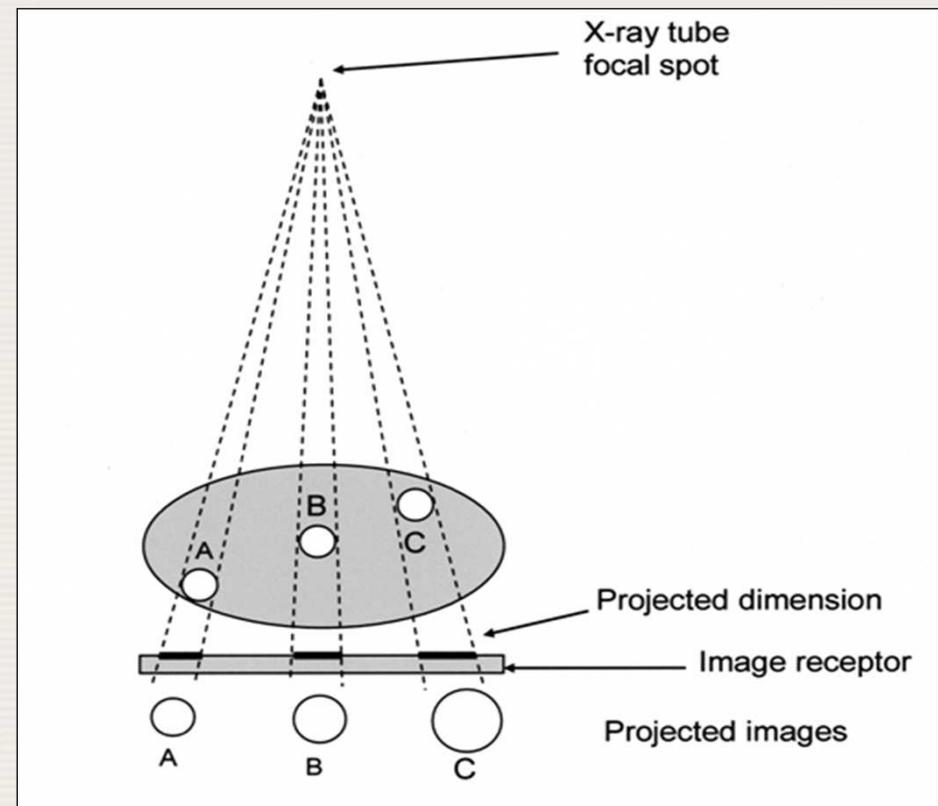
6.2.3 Effects of Projection Geometry

Geometrical Distortion - Position

All objects are magnified by an amount related to the **OID**

The further away from the OID the larger the object appears

In diagram all objects A, B and C are the **Same** size, but they appear progressively larger due to differences in position



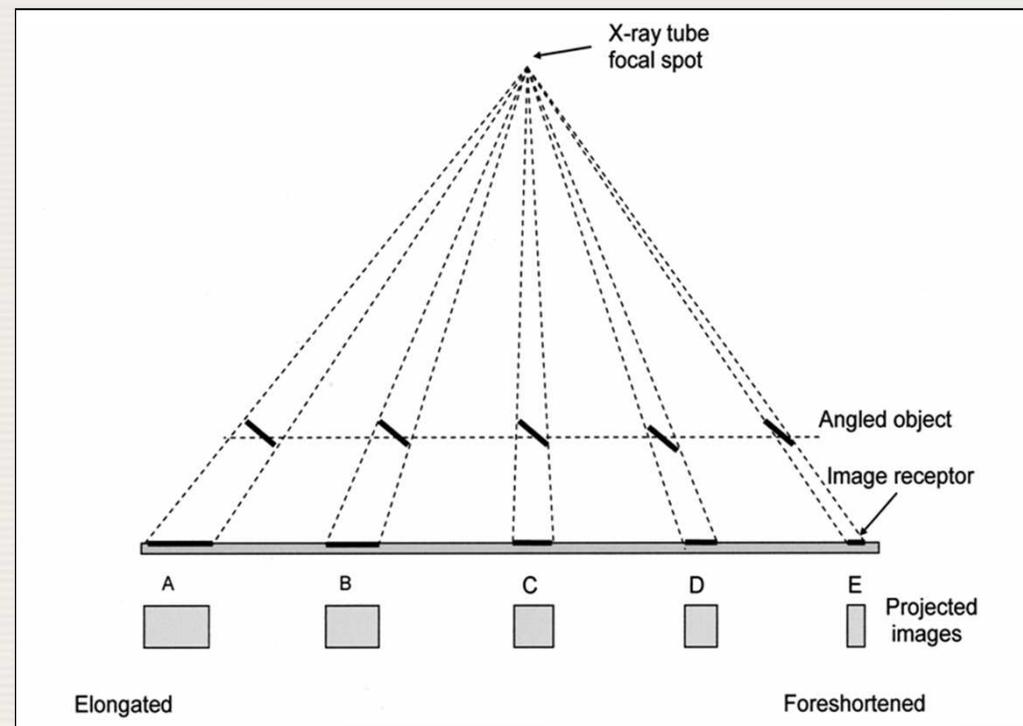
Effect of depth of objects on their projected size

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Distortion - Shape

Tilted object is shown projected at a range of angles, illustrating the increasing degree of foreshortening as the angle increases



Effect of angulation on projected length of an angled object

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Inverse Square Law

For an isotropic **Point Source**, the X ray beam intensity is inversely proportional to the square of the distance from the source

An XRT with its attached collimator is a good approximation to a point source for distances greater than **~50 cm** from the focal spot

Only at low kV settings, such as those typical of **Mammography**, does air attenuation affect the inverse square relationship

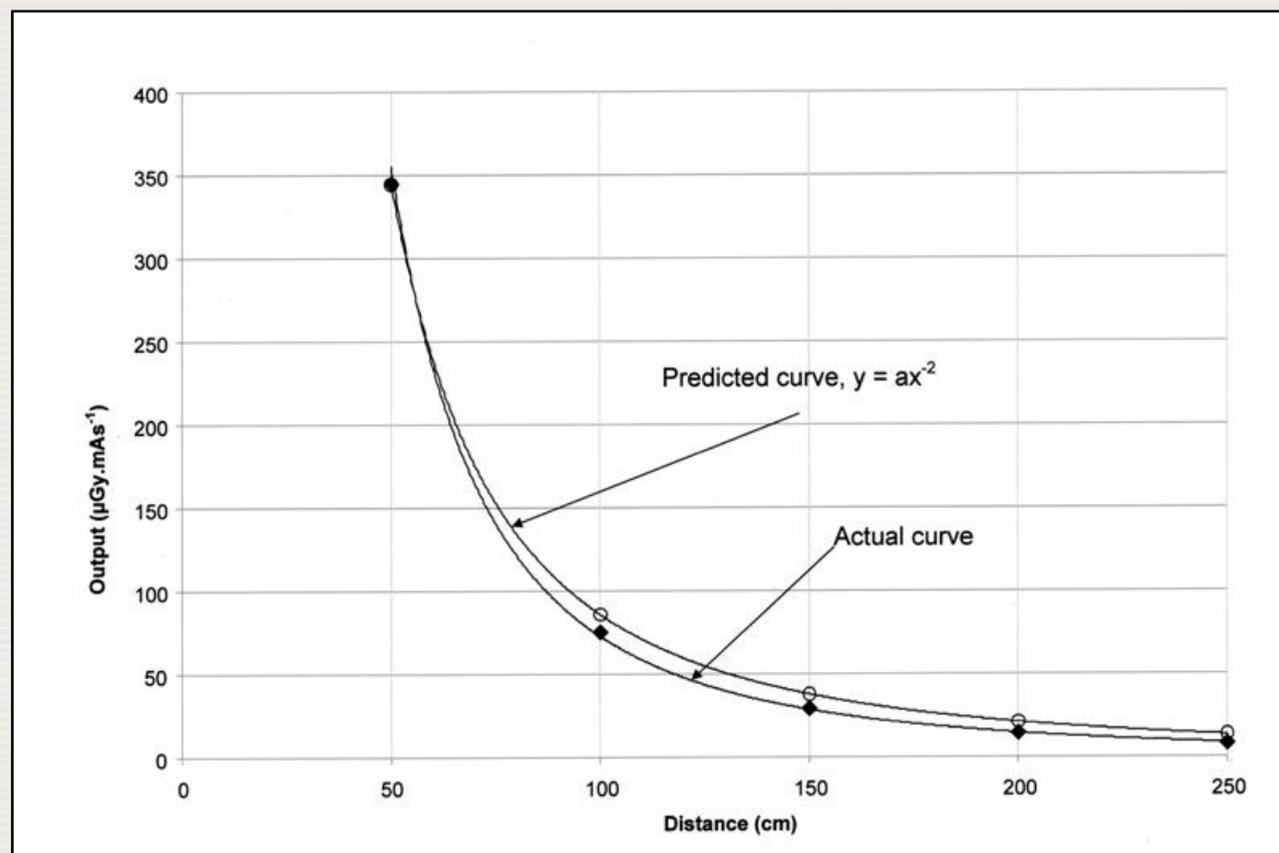
6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Inverse Square Law

Measured and calculated air kerma curves over FID range 20-250 cm

Deviation from inverse square law due to air attenuation, for a tungsten-target X ray beam with 0.5 mm aluminium added filtration at a voltage setting of 30 kV, no compression paddle



6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

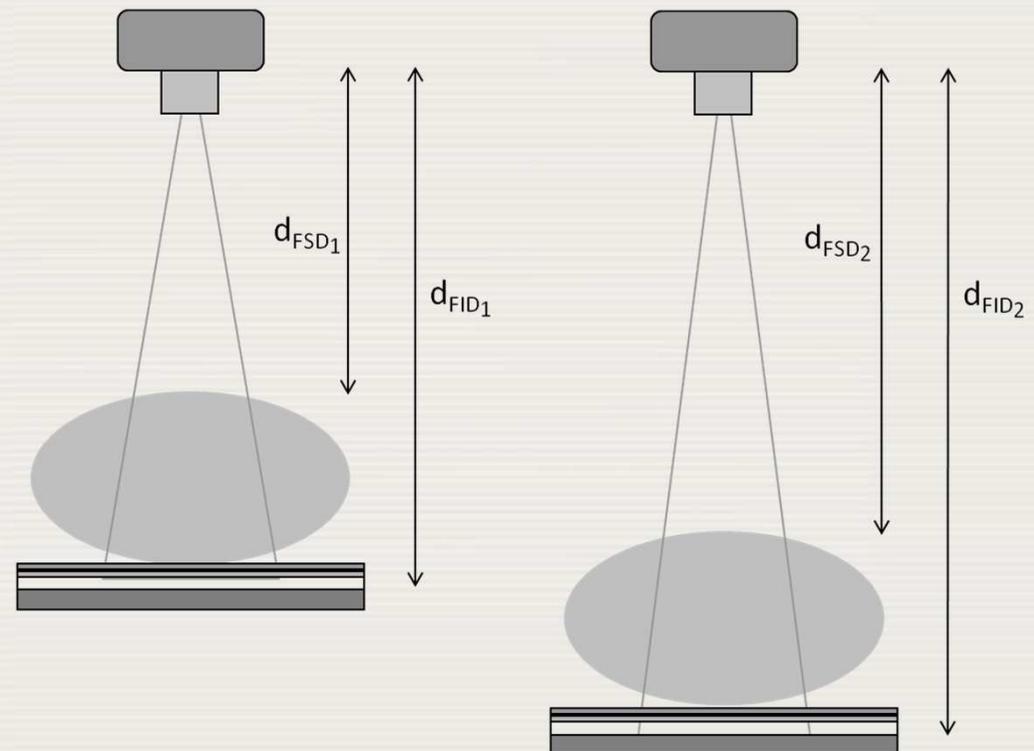
Inverse Square Law

To maintain the same air kerma at the image plane - an increase in the mAs is required as the FID is increased

The increase in mAs is given by:

$$\frac{mAs_2}{mAs_1} = \left(\frac{d_{FID_2}}{d_{FID_1}} \right)^2$$

where d_{FID} is the **FID**



6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

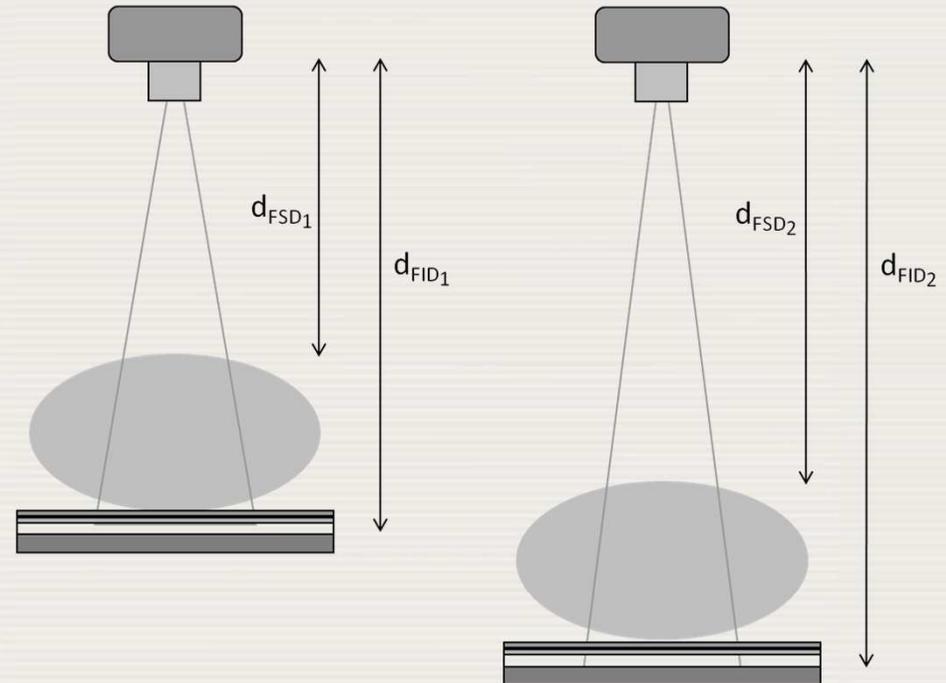
Inverse Square Law

Air kerma at patient entrance surface is **Greater** than that at the image receptor (neglecting attenuation) by the ratio:

$$\left(\frac{d_{FID}}{d_{FSD}} \right)^2$$

Entrance air kerma (K_i) decreases with increased FID according to:

$$K_{i_2} = K_{i_1} \left(\frac{d_{FID_2} d_{FSD_1}}{d_{FID_1} d_{FSD_2}} \right)^2$$



Effective Dose ~ same

as increase in entrance surface beam size needed

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Ideal image **Sharpness** would be produced by a **Point Source**

The spatial resolution in such a case being limited by the image receptor factors such as

- **Phosphor layer Thickness,**
- **Lateral Spread** of light in scintillators, and the
- **Image Matrix**

However, due to the restriction on the permissible temperature of the focal spot and focal track of the anode, typical focal spot sizes of 0.6 mm to 2.0 mm are required

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Most XRTs also have a **Fine** focal spot for high resolution images of small body parts

Typically the fine focal spots are **0.3-1.0 mm**, but must use lower mAs to protect the tube from heating effects

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

The **Spatial Resolution** depends on the focal spot size and the image receptor and **both** need to be considered

For the demagnified image, the width of the penumbra, or more correctly the **Edge Gradient**, due to a focal spot of size X_F is given by the geometric unsharpness (U_g) divided by the magnification:

$$U_g = X_F \frac{d_{OID}}{d_{FID}}$$

where d_{OID} is the OID

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

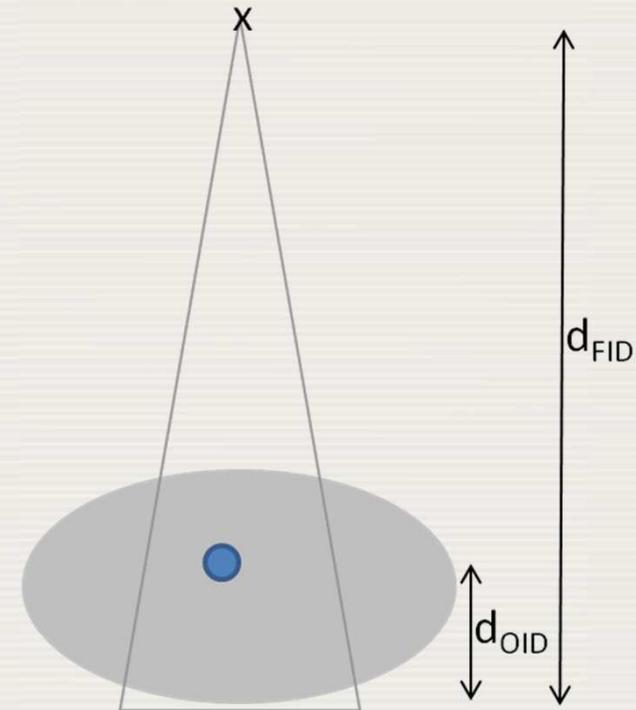
Geometrical Unsharpness

Since the **Magnification** of the object at the image receptor is given by:

$$m = d_{\text{FID}} / (d_{\text{FID}} - d_{\text{OID}})$$

the geometric unsharpness can be rewritten as:

$$U_g = X_F \frac{d_{\text{OID}}}{d_{\text{FID}}}$$



6.2 X-RAY IMAGE FORMATION

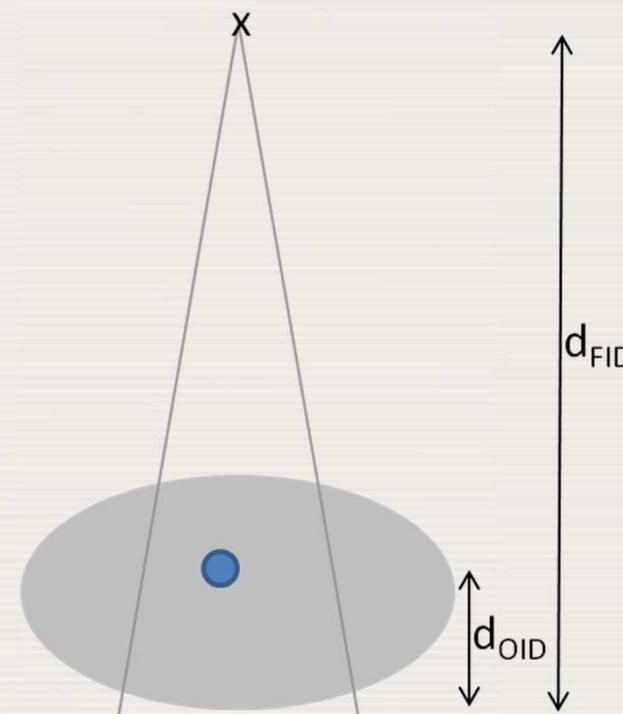
6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Change in FID

to maintain the same focal spot resolution, the new focal spot size may be determined from:

$$X_{F_{new}} = X_{F_{old}} \frac{d_{FID_{new}}}{d_{FID_{old}}}$$



Magnification also will change which will affect the overall image sharpness because of the effect of the image receptor **Blur**

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Change in FID The overall unsharpness (**U**) is given as

$$U = U_r \sqrt{\left(\frac{1}{m^2} + \left(1 - \frac{1}{m}\right)^2 \left(\frac{X_F}{U_r}\right)^2 \right)}$$

U_r is the **Intrinsic Image Receptor Unsharpness** (i.e. for **m=1**)

It is assumed that the geometric and receptor unsharpness can be added in **Quadrature**

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Optimization of projection radiographs involves choosing an appropriate focal spot size

This requires a **Compromise** between the exposure time and the resolution

For Example, a very small focal spot will provide good spatial resolution, but only permit a low tube current, therefore requiring a long exposure time, leading to increased risk of motion blur

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

Whilst it may be considered that **Quantum Noise** limits the detectability of fine details

There is Some Evidence that Smaller Focal Spots than are Currently Employed may Lead to Improved Spatial Resolution

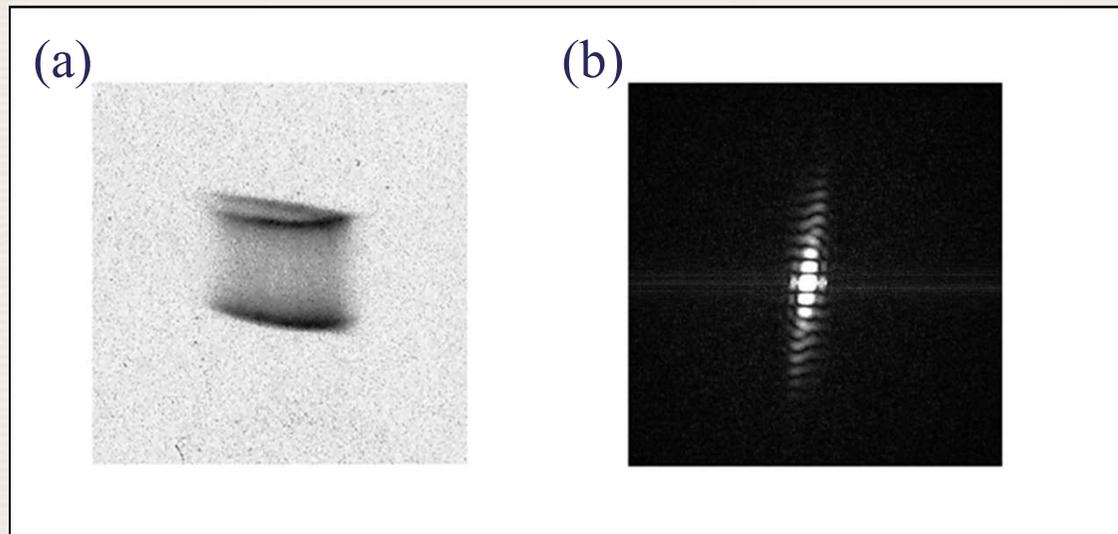
This is because the system **DQE** is affected by the focal spot MTF

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

The **Focal Spot MTF** may be measured using a pinhole to determine the PSF or a slit to determine the LSF, and calculating the normalised modulus of the Fourier Transform of the spread function



(a) A pinhole image a 2.0 mm focal spot showing a typical distribution of the X ray intensity

(b) the corresponding 2D MTF

6.2 X-RAY IMAGE FORMATION

6.2.3 Effects of Projection Geometry

Geometrical Unsharpness

The MTF of a focal spot is given by **Convention** for a magnification factor of 2.0

To correct the MTF for the true magnification, the frequency axis must be **Scaled** according to:

$$f_{new} = f_{old} \left(\frac{m_{old} - 1}{m_{old}} \times \frac{m_{new}}{m_{new} - 1} \right)$$

6.2 X-RAY IMAGE FORMATION

6.2.4 Magnification Imaging

Magnification is a relatively uncommon technique

Probably the most important example of its use is in magnification **Mammography**

However, there are instances when significant magnification is **Unavoidably Present** in standard radiographic projections

These include the:

- Lateral hip projection
- Lateral cervical spine projection

6.2 X-RAY IMAGE FORMATION

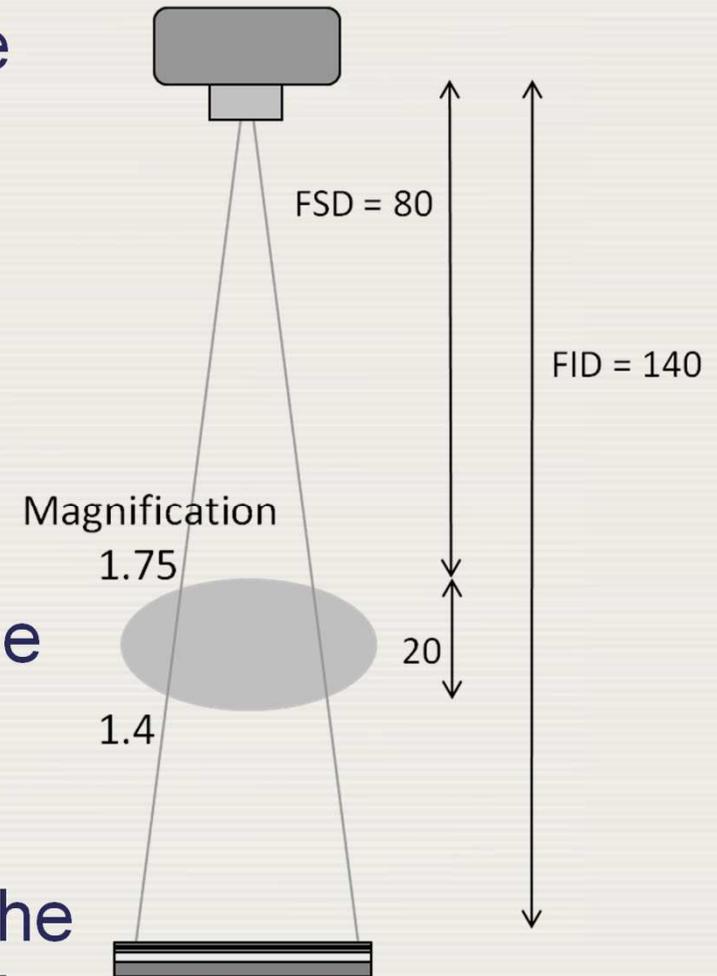
6.2.4 Magnification Imaging

Magnification is achieved by increasing the OID which generally requires an increase in the FID as well

The actual magnification achieved varies with depth in the patient

Example: Patient thickness is 20 cm, the FID 140 cm and the FSD 80 cm

the magnification varies between 1.4 at the **Exit** side of the patient to 1.75 at the **Entrance** side



6.2 X-RAY IMAGE FORMATION

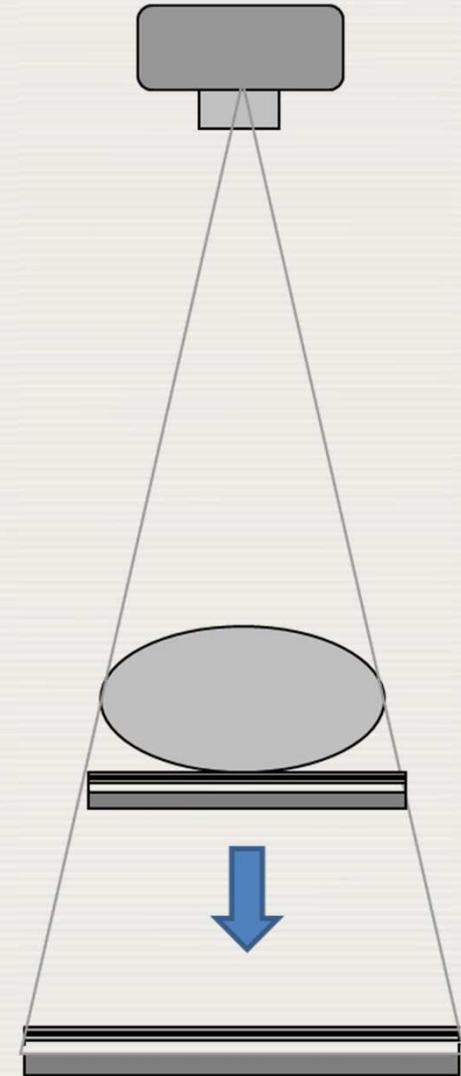
6.2.4 Magnification Imaging

Magnification requires employment of a **Larger** image receptor

For large body regions this may **Not** be possible

The use of magnification has consequences for:

- Dose
- Spatial Resolution and
- SNR



6.2 X-RAY IMAGE FORMATION

6.2.4 Magnification Imaging

Dose

A number of **Effects** occur when increasing the OID

There is a substantial reduction in the **Scatter Fraction** at the image receptor, because the scattered rays are generally directed away from the receptor

To maintain the dose to the image receptor, an increase in the mAs and hence the patient dose would be required

Mainly because of the loss of scatter but also due to the increase in FID due to the inverse square law

6.2 X-RAY IMAGE FORMATION

6.2.4 Magnification Imaging

Dose

Because of the reduction in scatter fraction, magnification may usually be performed **Without** the use of an anti-scatter grid

This leads to a reduction in mAs in proportion to the **Bucky Factor**

The Bucky Factor is the ratio of the mAs with scatter reduction method divided by the mAs without scatter reduction method (typically 3-6)

6.2 X-RAY IMAGE FORMATION

6.2.4 Magnification Imaging

Unsharpness

An **increase** in the OID leads to a **reduction** in image sharpness

due to the geometric blur of the focal spot

Use of magnification techniques requires a

Significant Reduction

in focal spot size

compared to contact methods



6.2 X-RAY IMAGE FORMATION

6.2.4 Magnification Imaging

Unsharpness

Improvement in the overall sharpness of the complete system is generally because of the increase in size of the image compared to the **Unsharpness** of the image receptor

From effects such as:

- **Light Spread** for screen-film systems and the
- **Pixel Size** for digital systems

Magnification can therefore **Improve Spatial Resolution**, compared to the result of a simple zoom of a digital image which enlarges the pixels as well as the image



6.2 X-RAY IMAGE FORMATION

6.2.5 Contrast Agents

The **Subject Contrast** for many X ray examinations is **low**, due to the similarity in the atomic number and density of soft tissues and blood

The contrast of organs and of the circulatory system may be substantially increased with the use of higher atomic number **Contrast Agents**

These generally employ **Barium** compounds for study of the GI tract, and **Iodine** compounds for soft tissues and the circulatory system

6.2 X-RAY IMAGE FORMATION

6.2.5 Contrast Agents

Iodine and barium have considerable photoelectric attenuation due to their **K-Edges** being in the diagnostic X ray energy range

I: 33 keV; Ba: 37 keV

The maximum contrast that can be achieved will occur for photon energies **Just Above** the K-edge of these elements

This in turn requires the choice of a kV setting that produces exit spectra with the majority of photon energies within the appropriate region of the spectrum

Optimum kV settings are between **60-70 kV** for I and up to **80 kV** for Ba

6.2 X-RAY IMAGE FORMATION

6.2.5 Contrast Agents

Examinations of the GI tract sometimes employ **Air** as well as barium

It should be noted that there is a small but unavoidable incidence of **Adverse Reactions** to contrast media

Generally minor but occasionally serious or even fatal

Recently, targeted contrast agents have been developed based on **Gold Nanoparticles** that have superior contrast enhancement to traditional iodine based agents with minimal toxicity and negligible negative reactions

6.2 X-RAY IMAGE FORMATION

6.2.6 Dual-Energy Imaging

An **Alternative**, and sometime an additional, method of improving image contrast is with the use of **Two** quasi simultaneous images of the body using different X ray spectra and processing them into separate images:

- one more reflecting the **Photoelectric Process** and
- one the **Compton Effect**

6.2 X-RAY IMAGE FORMATION

6.2.6 Dual-Energy Imaging

By combining these images using specific **Weightings**, differences between:

- Bone tissue and soft tissue or between
- Air and soft tissue

can be displayed

In order to make these separations, X ray images acquired at different

- **Tube Voltages** and/or
- **Filtrations**

are required

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

With **Screen-Film** systems, technique selection is relatively straightforward:

- The choice of **kV** setting is based largely on the required contrast
- the **mAs** is chosen to produce a suitable optical density for the region of clinical interest

generally ~1.0 net OD

With **Digital** systems, the direct link between technique setting and image appearance has been lost, making correct technique selection much more difficult

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

To determine whether a detail will be detectable in the image, **Noise** must be considered

The primary **Source** of noise is generally the random arrival of photons at the image receptor, a **Poisson** process

From **Rose's** expression, the number of detected photons required per unit area, to image a detail of size **d** and contrast **C** with a signal to noise ratio of **k**, is:

$$N = k^2/C^2d^2$$

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

The value of k required to be certain that an observed detail is real and not due to chance fluctuations in the number of photons is often taken to be **5**

Thus as C is increased, the number of photons required at the image receptor is reduced so that

a Reduction in kV will Produce an Image of Satisfactory Quality at a Lower Image Receptor Dose

provided that the contrast range does not exceed the dynamic range of the image receptor



6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

However, This Reduction in kV will Require an Increase in the mAs, Leading to an Increase in Patient Dose

The dose to the image receptor depends approximately on kV^5 , and is linear with mAs

The patient dose (K_i), is proportional to mAs and approx to kV^2

The overall effect on patient dose therefore is approximately proportional to kV^{-3}

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

For Example, consider a setting of 60 kV at 40 mAs

Using the **15% Rule**, this could be changed to 69 kV at 20 mAs

The patient dose will then be reduced to $(69/60)^{-3} = \mathbf{66\%}$

However, the increase in kV will result in a reduction in the **CNR**, which may be acceptable, in which case a worthwhile reduction in dose will have been achieved

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

If the image receptor dose is considered to be a variable, then there is a wide range of **kV** and **mAs** combinations that will produce a diagnostically acceptable image

but at a Wide Range of Patient Dose Levels

In order to manage **Digital** imaging systems; suitable levels of image receptor dose have been determined by all of the manufacturers of such systems, expressed in a variety of proprietary **Exposure Indices** (EI) to represent the dose to the image receptor

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

Generally there will be a selection of indices suitable for imaging extremities, trunk and chest

For **CR**

these correspond approximately to screen-film system speeds of 100, 200 and 400 respectively

Direct and indirect **Digital** systems are somewhat faster
allowing higher nominal speeds

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

For all **Digital** systems the choice of suitable kV and mAs combinations requires that for each projection:

- the **kV** and **mAs** produce the correct value of the **EI** and that
- the maximum value of the **kV** is chosen that will allow diagnostically acceptable **CNR**

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

This is readily **Demonstrated** in practice

If a suitable phantom is radiographed at a low kV with suitable mAs, and a range of further images is obtained at increased kV settings at the same mAs, the images will appear **Very Similar**

This is because the reduction in contrast with increasing kV is matched by the increased number of photons detected, resulting in **Similar CNR** for each image

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

Each increase in kV will cause an increase in patient dose by kV^2 , so such a procedure is clearly clinically unacceptable

If Instead, each kV increase is accompanied by a reduction in mAs to **Maintain** the image receptor dose and EI, then the image quality will become **Steadily Worse** as kV is increased, until a point is reached at which the image quality is no longer acceptable

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Effect of Tube Voltage on Contrast, Noise & Dose

A corollary to this is that images exposed with a kV below the optimum level and an mAs above the optimum level look better

leading to the phenomenon of

Exposure Creep

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Matching Technique to Study

The choice of **Suitable kV** setting for any body region involves **Two** steps

The **First** is to choose a suitable image receptor dose and speed to produce acceptable levels of image noise

For Example, a wrist X ray may require a nominal speed of 100, whereas a PA chest may permit 400 speed

Regions of low subject contrast such as the abdomen then require a relatively low kV setting, whereas regions of high contrast such as the chest require a high kV setting

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Matching Technique to Study

Guideline kV settings are widely available

Such as those given in the quality criteria documents of the EU

The kV setting should then be increased gradually, with appropriate mAs reduction to maintain the **CNR** or film optical density

Until the Loss of Image Quality is Just Tolerable



6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Matching Technique to Study

For screen-film imaging this also requires matching the **Dynamic Range** of the image receptor system to the range of the input signal

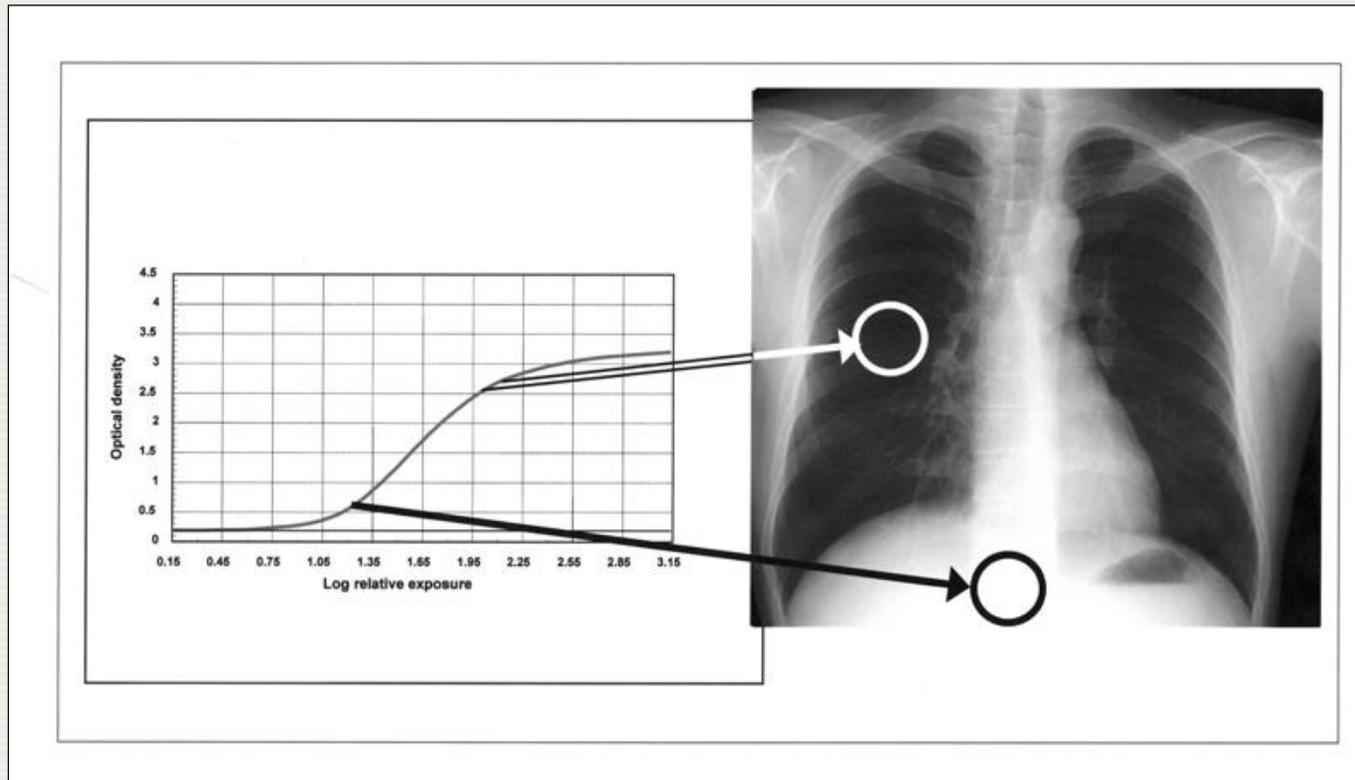
This is illustrated in the following figures for the **Two** extreme cases:

- **Chest Radiography**, which has high subject contrast with a wide latitude image receptor system
- **Mammography**, which features low subject contrast with a narrow latitude image receptor system

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Matching Technique to Study

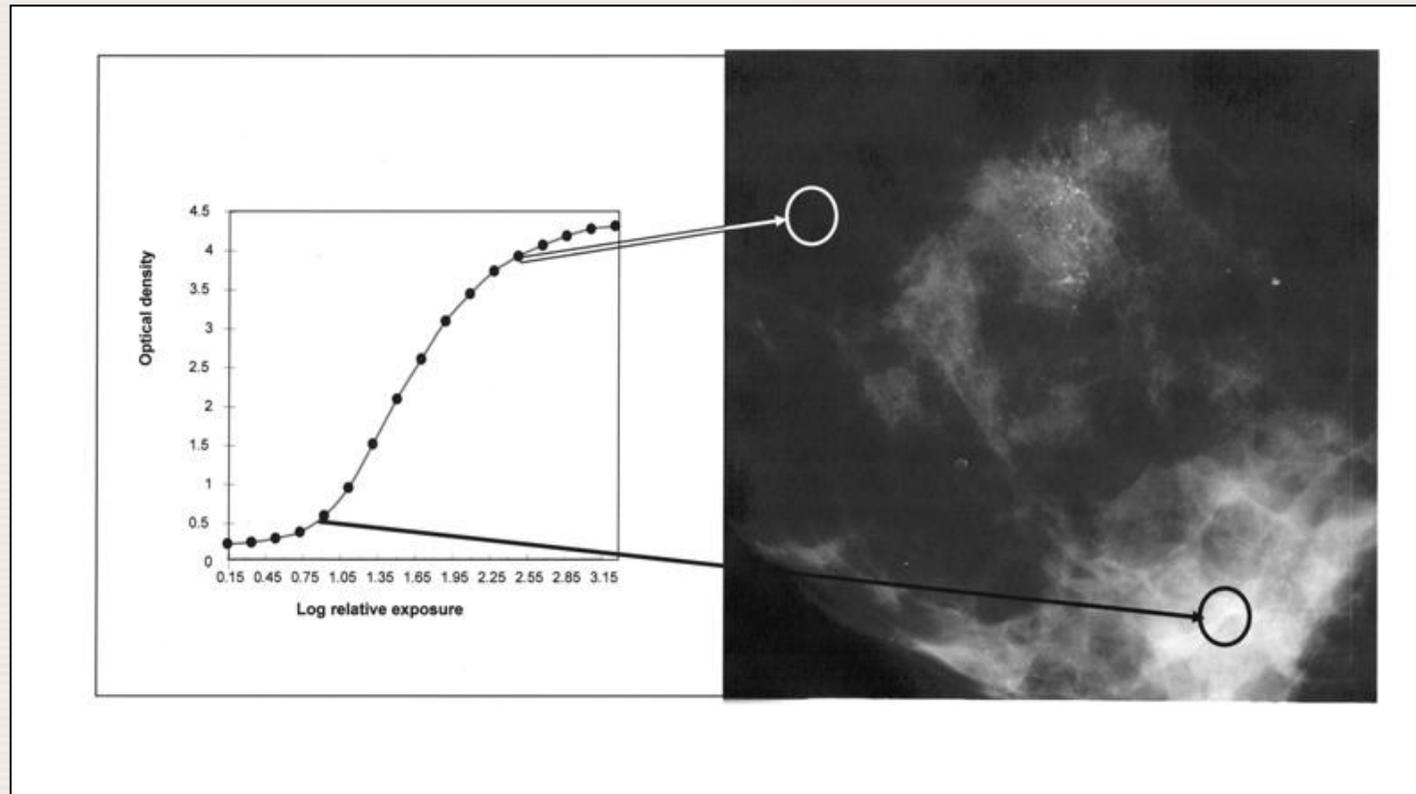


Matching kV setting and mAs to dynamic range of the image receptor for a study of a high contrast region of the body (a chest x-ray)

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Matching Technique to Study



Matching kV setting and mAs to dynamic range of the image receptor for a study of a low contrast region of the body (a mammogram)

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Relationship between kV & mAs

Given that the image receptor dose is proportional to **mAs** and to **kV⁵**, some simple exposure rules may be derived

First it is observed that an increase in kV of 15% results in an increase in image receptor dose by a factor of **two**

Hence the so-called **15% Rule**, that an increase in kV of 15% is equivalent to a doubling of the mAs and a reduction by 15% to halving the mAs

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Relationship between kV & mAs

Furthermore an increase in kV of 5% results in an increase in image receptor dose of 30% leading to the **5% Rule**:

**That a 5% Increase in kV is Equivalent
to 30% Increase in mAs**

a reduction of 5% in kV
to a reduction in mAs by 30%

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Relationship between kV & mAs

Finally, since 15% increase in kV is about 10 kV between 60 and 80 kV, another commonly used rule is that

10 kV Increase is Equivalent to Doubling the mAs

and 10 kV reduction to halving the mAs

None of these rules are **Exact**, but their use is satisfactory because of the tolerance for small exposure errors due to the latitude of screen-film systems, and because of the very wide dynamic range of digital systems

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Automatic Exposure Control (AEC)

Even with the most skilled of practitioners, manual setting of technique factors results in inconsistent exposures, so that:

- Optical densities vary in **Screen-Film** imaging and
- Image noise levels vary with **Digital** systems

In addition a number of **rejects** and **repeats** are unavoidable due to exposure errors

AEC Systems are intended to increase exposure consistency and reduce reject and repeat rates



6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Automatic Exposure Control (AEC)

The **Principle** is to

- measure the X ray **Flux** at the image receptor and to
- **Terminate** the exposure when sufficient **Energy** has been absorbed

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Automatic Exposure Control (AEC)

However, the advantages of AEC systems may be achieved only if the systems are **Correctly** calibrated and **Properly** used

Calibration is required for a number of reasons, including energy dependence and beam hardening

Energy Dependence is due to the varying sensitivity of the AEC detectors and the image receptor system at different kV settings

Correction Factors are included in the control system to allow for kV setting

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Automatic Exposure Control (AEC)

More difficult is the **Beam Hardening** caused by the patient

The system is not able to measure the amount of beam hardening, and hence cannot correct for beam hardening errors

Therefore, AEC systems include controls for **Manual** correction by the **Radiographer**

6.2 X-RAY IMAGE FORMATION

6.2.7 Technique Selection

Automatic Exposure Control (AEC)

These generally include **Compensation Settings** such as:

-3, -2, -1, 0, +1, +2, +3

Each setting increasing the mAs delivered by a constant factor such as $\sqrt{2}$

There is also a **Patient Size Button**, with settings for a thin patient, an average patient and a large patient

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

It is often stated that photon **Scattering** is of no benefit for projection radiography, leading only to **Fogging** of the image

However this is incorrect, as the appropriate contrast for each and every projection is chosen by setting a suitable kV to provide the **Correct Proportion** of photoelectric and scatter interactions

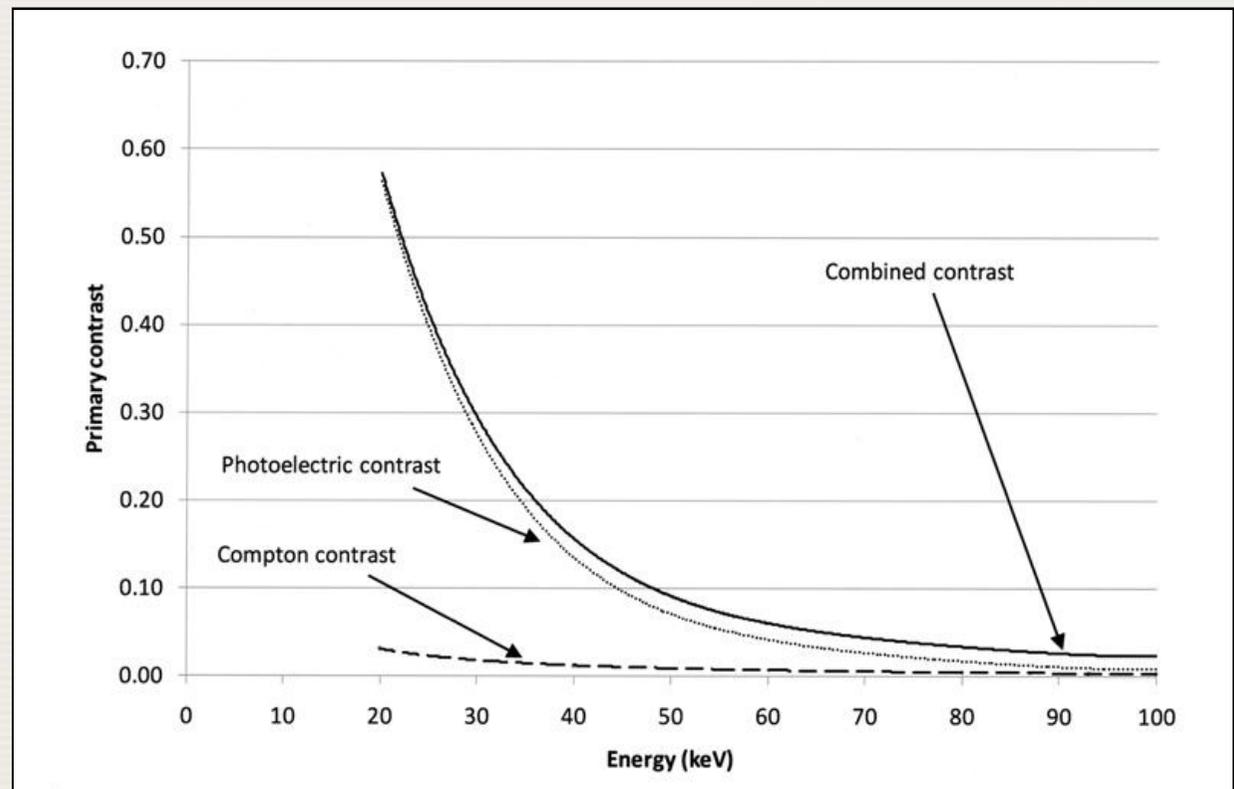
At low kV settings the contrast is high, due to the predominance of the **Photoelectric Effect**, while at high kV it is low, due to the predominance of **Scattering** interactions

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

These effects are illustrated below - showing the contribution of photoelectric and scattering interactions for the primary contrast of a **1 mm sphere** of calcium embedded in water, calculated over a range of energies from 20-100 keV

Calculated using:

$$C_P = 1 - e^{-(\mu_d - \mu_b)x_d}$$



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.1 Origins of Scattered Radiation

Both the **Incoherent** and **Coherent** interactions lead to scattered radiation impinging on the image receptor

The **Former** is by far the more significant

For Example

for a 10 cm patient thickness and a 20 x 20 cm² X ray field only 19% of the **Scatter Fraction** is due to coherent scatter at 50 kV

and 9% at 120 kV

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

The **Magnitude** of the scatter depends on many variables

The dependences on:

- Radiographic procedure,
- X ray beam size,
- Patient thickness and
- Position in three-dimensions

are described in the following sections

The magnitude is also widely claimed to depend on the **kV setting**, but as shown here, this is the case only for very inefficient image receptors

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence upon Radiographic Procedure

The radiographic procedure itself has a strong influence on the proportion of scatter, depending on whether the subject is a region consisting largely of **bone**, or **soft tissue**, or some intermediate **combination**

This is because the **Scatter** interactions are predominant for soft tissue, but the **Photoelectric** interaction is predominant for bone over much of the diagnostic energy range (except mammography)

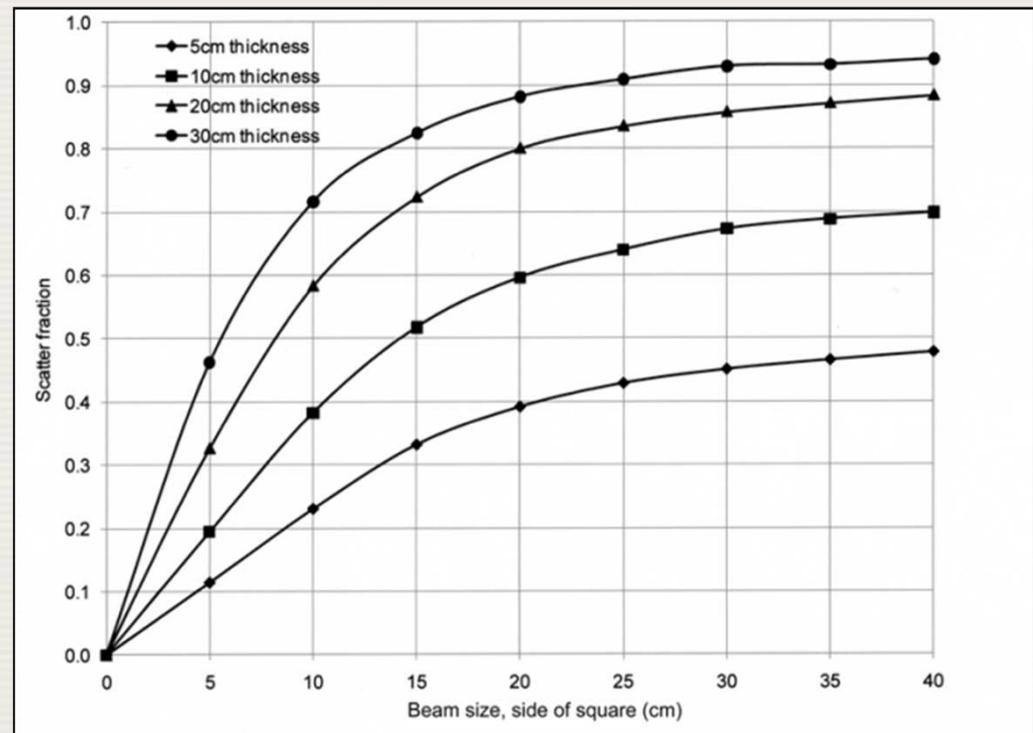
6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Field Size

As the field size increases from a narrow beam with almost no scatter to very large, the **Scatter Fraction** increases until a saturation level is reached, beyond which little or no further increase in scatter fraction occurs

Dependence of scatter fraction on beam area for four phantom thicknesses at 80 kV

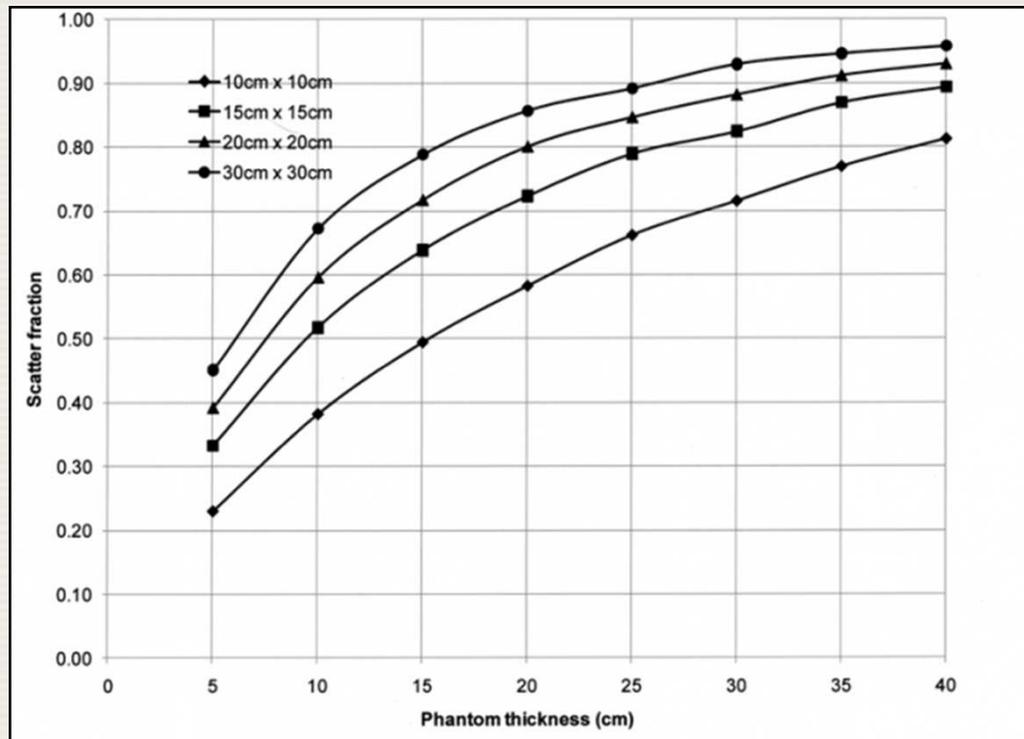


6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Thickness

The **Scatter Fraction** increases rapidly with patient thickness, but tends to saturate for very large patients



Dependence of scatter fraction on phantom thickness for four X ray field sizes at 80 kV

These data also demonstrate the necessity for **Scatter Reduction Methods**, especially for large patients

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

Whilst the **Scatter Fraction** (at the image receptor) has been considered so far, it is also important to quantify the scattered radiation in all directions from the patient, as this affects the dose to **Personnel**

Knowledge of scatter levels is required in order to determine appropriate **Radiation Shielding** levels

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

It is also useful to consider the proportion of **Backscatter** at the patient's entrance surface as this:

- May contribute significantly to the **Skin Dose** and
- Complicates measurements of patient dose when **Free-in-Air** measurements are used

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

Scatter may be categorized as:

- **Forward-Scatter**

affecting mainly the **Image Receptor**

and

- **Side- and Back-Scatter**

affecting dose to **Personnel**

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

If a small volume of soft tissue is considered, and we make the simplifying assumption that the scatter angular distribution is given by the **Klein-Nishina formula**:

the Proportions of Forward-Scatter and Back-Scatter can be seen to be Similar for Photons in the Diagnostic Energy Range

with the proportion of forward scatter increasing as the photon energy is increased

At **90°** the differential cross-section is approximately half that
at **0°** and at **180°**

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

However, for the large volumes of tissue typical in projection radiography there is considerable attenuation, so that back-scatter can be **Significant**

For larger body regions the proportion of photons which undergo more than one scatter is significant, leading to a **More Isotropic** scatter distribution than predicted by the Klein-Nishina formula

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

It is found experimentally that, to a close approximation:

- The magnitude of scattered radiation follows the **Inverse-Square Law** at distances greater than ~500 cm from the patient and
- That the magnitude is **Directly** proportional to the X ray field size

A Rule of Thumb for scatter levels is that the scatter dose at 1 m from the patient is 1/1000 of the dose in the primary beam at the patient entrance surface

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Position (in 3D)

With respect to **Position** in the image, the proportion of scattered radiation is found to be:

- Greatest in the centre
- Reduced at the edges and
- Reduced further still at the corners of the field

see table 6.1

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Energy

In soft tissue, as the energy is increased:

- **Photoelectric** cross section reduces approximately as $1/E^3$, whereas
- **Scattering** cross-section reduces as $1/E$

Hence the **Probability** of scattering events increases relative to photoelectric events as the energy increases, but the overall probability of scattering events decreases

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Energy

Conversely, the energy of the scattered photons increases as the energy increases

So that they are less likely to be attenuated in the body and are more likely to escape

The overall effect is that side-scatter, back-scatter and forward-scatter all increase as energy is increased

However, the primary beam attenuation also decreases as energy is increased



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Energy

Consequently, the scatter fraction reaching the detector is found to show **Little** dependence on energy

In the case of **Efficient** image receptors, almost all of the primary beam and scatter are absorbed, so the scatter fraction for the receptor is similar to that reaching the receptor

However, **Inefficient** image receptors are more sensitive to scattered photons than to primary photons

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Energy

This is because the primary beam is incident approximately **Perpendicular** to the receptor, so the path length is similar to the phosphor thickness

However the scattered photons are incident **Obliquely**, resulting in greater path-length on average and a greater probability of absorption

They may also be of **Lower** energy, which will generally also increase the **Probability** of absorption

This effect becomes more significant as energy is increased, so the detected scatter fraction also increases with energy

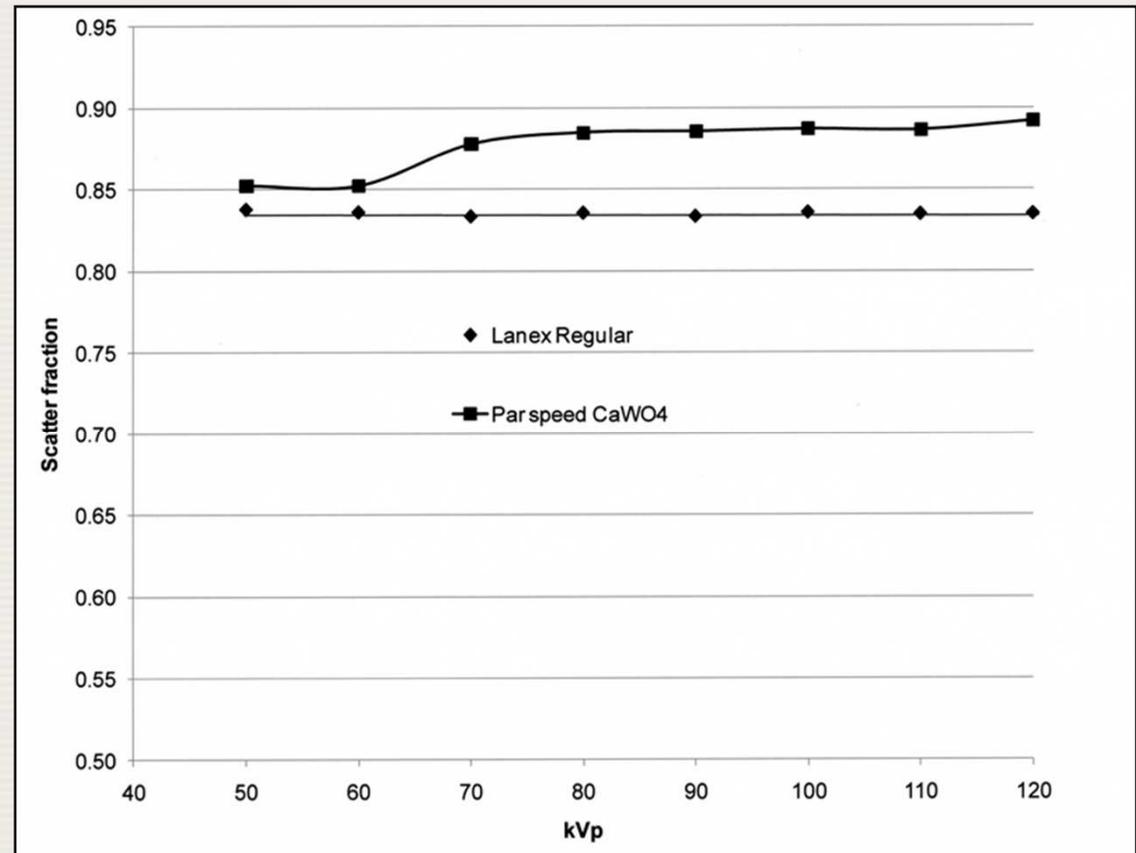
6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.2 The Magnitude of Scatter

Dependence on Energy

Scatter Fractions for Kodak Lanex Regular and a CaWO_4 par-speed screen from 50-120 kV:

The scatter fraction for an efficient rare-earth screen, is independent of kV setting, whereas the scatter fraction increases with energy for the (**Obsolete**) par speed screen



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.3 The Effect of Scatter

Contrast Reduction

The effect of scatter on contrast is quantified by the **Contrast Degradation Factor (CDF)**, which is given by:

$$CDF = \frac{1}{1+sp}$$

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.3 The Effect of Scatter

Noise

The **Quantum Noise** in the image arises from **both** primary and scattered photons

Both of these image contributions therefore affect the SNR

Digital chest imaging with a grid shows that **SP** varies between
~**2** in the region of the mediastinum,
to ~**0.6** in the lung fields

leading to a value of SNR per pixel
behind the mediastinum of ~**20**, and
in the lung fields ~**60**

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

The scatter component of the image may be considered to consist of the primary image convolved with a **Scatter Spread Function** which gives a highly blurred version of the image

The resulting image may be considered to be the **Sum** of these two images

Efforts are being made to employ this idea for **Computerized Scatter Removal**, rather than using grids or other methods

This approach is complicated because the scatter spread function is not shift invariant



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

In the absence of computerized methods, the use of **Anti-Scatter Grids** is routine for the vast majority of radiographic projections, apart from those of the extremities

Grids vary greatly in terms of the

- **Degree** of scatter rejection, and in the
- **Increase** in dose to the patient

that their use requires

All are designed to **Allow** a large proportion of the primary photons to reach the image receptor whilst **Removing** a good proportion of the scattered photons from the radiation field

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

Apart from the **Special** cellular grids used in some mammography systems, a grid generally consists of an array of **Thin Lead Strips** aligned to allow:

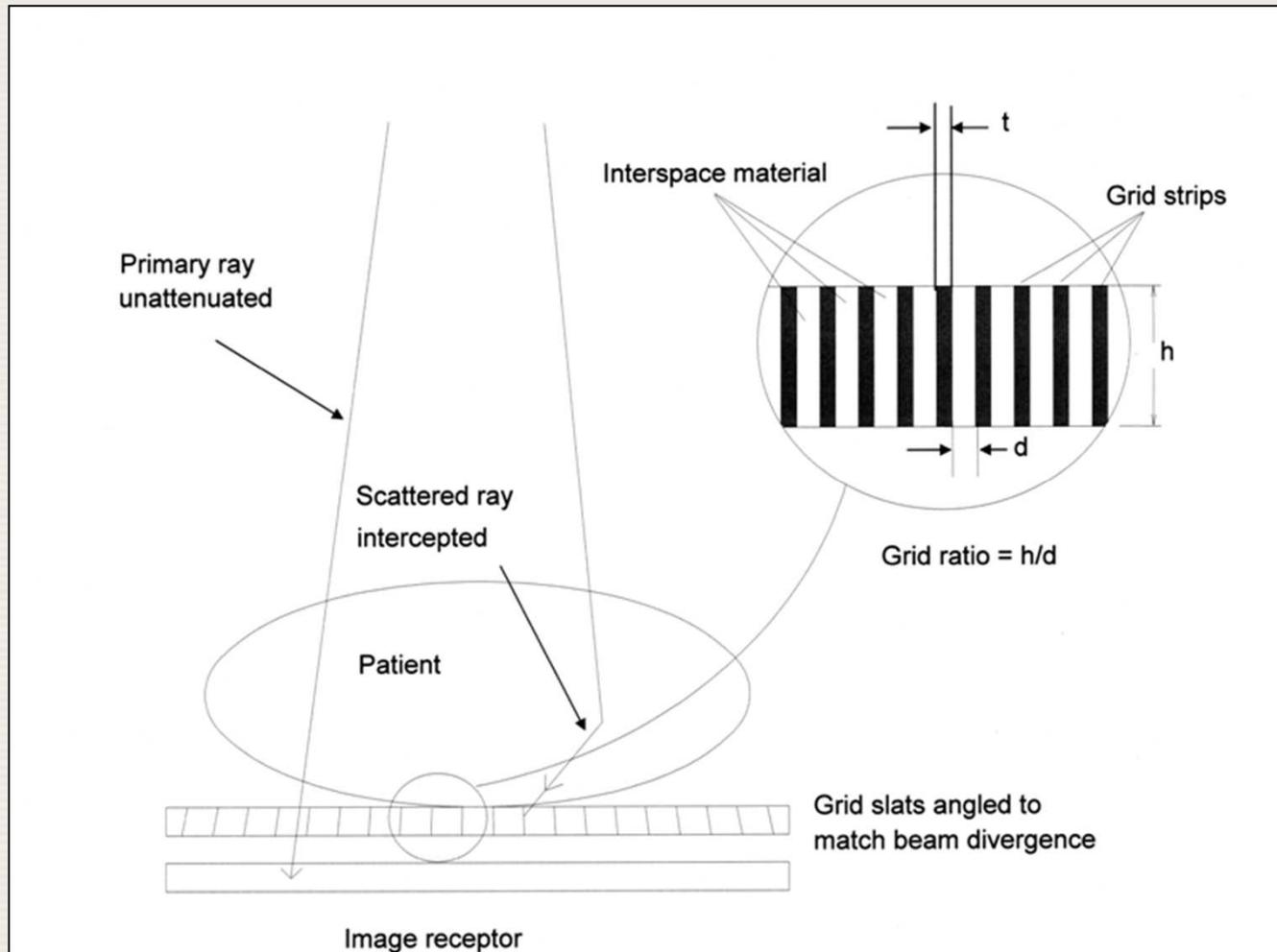
- **Passage** of the primary beam, but
- **Interception** of most of the scattered photons

The lead strips are separated by an **Interspace Material** and have protective covers on the top and bottom

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction



Construction and principle of operation of a focused anti-scatter grid (not to scale)

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

The number of lead strip lines per cm is known as the **Strip Frequency**

The ratio of the height of the lead strips to the width of the interspace material is known as the **Grid Ratio**, which is therefore given by:

$$\text{Grid ratio, } r = \frac{h}{d}$$

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

Scattered photons whose angle of incidence is less than $\tan^{-1} r$ will hit a lead strip, so that scatter rejection will increase with increasing grid ratio

On the Other Hand, the transmission of primary photons through the grid will decrease because of the increased thickness of the interspace

If the **Lead Strips** are too thin, they can be penetrated by the scattered X-ray photons and if they are too thick, they will stop too many primary photons

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

Thus the design of the grid is a **Compromise** between the requirements of:

- Good **Scatter Rejection** and
- High **Primary** photon **Transmission**

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

Usually the lead strips are **Tilted** to match the **Divergence** of the primary beam, at a chosen distance from the focus called the **Focal Length** of the grid

The grid is then referred to as a **Focused Grid**

Such grids must be used at the correct focal distance within a **permissible tolerance**

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

If a grid is used at the **Wrong** distance the tilt of the lead strips will not match the angle of divergence of the primary beam

The primary beam will then be **Attenuated** progressively more towards the edge of the image

This is termed

Grid Cut-Off

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

Parallel Grids have parallel strips

These always have some degree of **Grid Cut-Off** and

Should Not be Used at Short Distances

The **Degree** of cut off will be affected by the:

- Field Size used and the
- Grid Ratio

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

The grid **Interspace Material** is plastic, carbon fibre or other low atomic number material

Older grids used **aluminium**

The material of the **Grid Covers** should also be of low atomic number

Since the interactions of the X rays with the lead strips will mainly be photoelectric, for photons of energy above the K-edge of 88 keV the emission of **K-Fluorescent X Rays** must be considered in grid design

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

TYPICAL CONSTRUCTION DATA FOR THREE ANTI-SCATTER GRIDS

Grid Ratio	Strip Freq. (cm ⁻¹)	h (mm)	d (mm)	t (mm)
10:1	33	2.5	0.25	0.050
10:1	40	2.0	0.20	0.050
12:1	57	1.56	0.13	0.045

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Construction

The appearance on the image of the **Grid Lines** in radiographic images is generally **Unacceptable** for diagnosis

Furthermore, for **Digital** systems, the image of the grid lines may interfere with the pixel matrix, generating **Moiré Patterns** in the image

Consequently, a mechanism is usually provided to **Oscillate** the grid in a direction perpendicular to the grid lines to blur them out during the exposure

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Measures of Grid Performance

Grid **Performance** is specified in terms of parameters which relate to:

- the associated **Dose Increase** which is necessary and
- the **Improvement in Contrast** which is achieved

A **Good** grid will:

- Eliminate 80-90% of the scatter, while
- Transmitting at least 75% of the useful beam

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Measures of Grid Performance

Definitions

The **Primary Transmission**, T_p , is a narrow beam measurement of the ratio of X ray intensity with and without the grid present

The **Secondary Transmission**, T_s , is a broad beam measurement of the scattered radiation intensity with and without the grid

The **Total Transmission**, T_t , is a measurement of the total intensity of x-rays with and without the grid

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Measures of Grid Performance

Definitions

The Grid Factor or **Bucky Factor** (BF) is the dose increase factor associated with the use of the grid:

$$BF = \frac{\text{Exposure(mAs) required with grid}}{\text{Exposure(mAs) required without grid}} = \frac{1}{T_r}$$

The **Selectivity** is a measure of the effectiveness of the grid, given by:

$$\Sigma = \frac{T_p}{T_s}$$

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Measures of Grid Performance

Definitions

The **Contrast Improvement Factor** is given by:

$$\text{CIF} = \frac{\text{Contrast with grid}}{\text{Contrast without grid}} = \frac{T_p}{T_i}$$

It should be noted that as well as removing scatter, the grid will **Harden** the X ray beam and calculations of the CIF, should in principle allow for this effect - usually no more than a few per cent - and to a good approximation, the CIF can be calculated as:

$$\text{CIF} = (1 - \text{SF}_g) / (1 - \text{SF}_{ng})$$

where, SF_g and SF_{ng} are the scatter fractions with and without the grid

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Selection

BF AND CIF AT THE CENTRE OF THE IMAGE RECEPTOR FOR A SELECTION OF RADIOGRAPHIC GRIDS, USING A 25.4 CM X 25.4 CM X RAY FIELD WITH THE LUCAL CHEST PATIENT EQUIVALENT PHANTOM - 120 KV

Grid Ratio	SF	SP	BF	CIF
No grid	0.390	0.640	-	1.0
6:1	0.188	0.231	1.53	1.33
8:1	0.150	0.176	1.62	1.39
10:1	0.123	0.141	1.69	1.44
12:1	0.103	0.115	1.75	1.47
16:1	0.076	0.082	1.85	1.51

Low Bucky factor is because scatter level in chest radiography is low as the lungs act as a large air gap, and the thickness of other tissues is relatively low

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Selection

BF AND CIF AT THE CENTRE OF THE IMAGE RECEPTOR FOR A SELECTION OF RADIOGRAPHIC GRIDS, USING A 25.4 CM X 25.4 CM X RAY FIELD WITH THE LUCAL ABDOMEN PATIENT EQUIVALENT PHANTOM (SOFT TISSUE REGION) - 70 KV

Grid Ratio	SF	SP	BF	CIF
No grid	0.712	2.472	-	1.0
6:1	0.371	0.588	2.79	2.18
8:1	0.302	0.433	3.20	2.42
10:1	0.256	0.345	3.53	2.58
12:1	0.221	0.283	3.81	2.71
16:1	0.174	0.211	4.27	2.87

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Selection

BF AND CIF AT THE CENTRE OF THE IMAGE RECEPTOR FOR A SELECTION OF RADIOGRAPHIC GRIDS, USING A 25.4 CM X 25.4 CM X RAY FIELD WITH THE LUCAL ABDOMEN PATIENT EQUIVALENT PHANTOM (SPINE REGION) - 70 KV

Grid Ratio	SF	SP	BF	CIF
No grid	0.837	5.155	-	1.0
6:1	0.472	0.894	2.79	3.25
8:1	0.388	0.636	3.20	3.76
10:1	0.327	0.486	3.53	4.14
12:1	0.276	0.382	3.81	4.45
16:1	0.203	0.254	4.27	4.91

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Selection

From the data in the **Tables**, it is clear that for each radiographic projection there is an optimum grid ratio that will provide adequate scatter projection, with acceptable increase in dose to the patient

For Example, chest and abdomen projections on adults would require ratios of 10:1 or 12:1

Practically however, the grid is permanently fitted to the cassette holding device in radiographic tables and wall mounted devices, and these are generally of at least 10:1 ratio

Consequently grid use is generally **Far From Optimized**



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Artefacts & Alignment

There are several possible **Misalignments** that will lead to **Artefacts** in projection images

Additionally, a **Damaged** grid will generate artefacts and must be replaced

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

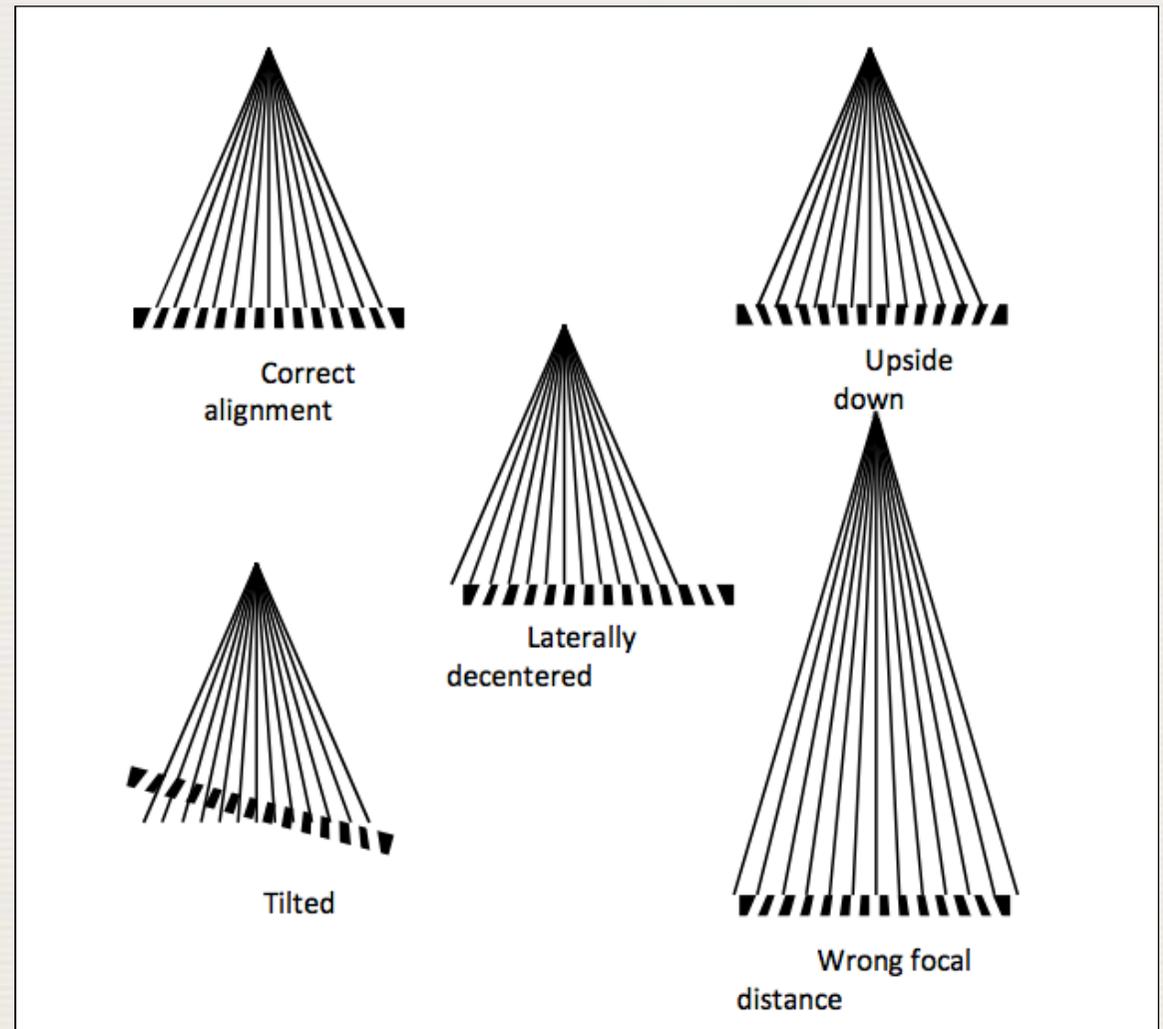
6.3.4 Methods for Scatter Reduction - Anti Scatter Grids

Grid Artefacts & Alignment

Possible misalignments
of the grid:

In practice it is possible for a
number of these to be
present **at once**

Note that the moving grid is
laterally **De-Centred** during
operation, although the
degree of offset is small on
average



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

While the use of a grid is effective at reducing scattered radiation, image quality may be further improved by:

- Careful **Collimation** and
- Patient **Compression**

Alternatives to anti-scatter grids include the use of

- Air Gaps and
- Slit Scanning Systems

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Collimation

Since smaller **Field Sizes** reduce the scatter fraction, it is good practice to

Collimate the X Ray Beam to as Small an Area as Possible for Each Projection

improving image quality
and
reducing patient dose

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Collimation

This requires

- Good **Knowledge** of anatomy and
- Good **Judgment** of the wide variety of patients

by **Radiographers**

to ensure that the **Region** of clinical interest is included

without **Needlessly** irradiating tissues that are not of interest

Care is required not to collimate too tightly increasing the possibility of the region of clinical interest being **Cut-Off** resulting in a **Repeated** exposure and increased dose

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Compression

Scatter Fraction increases with **Patient Thickness**

Therefore, if the patient thickness can be **Reduced** during exposure

by applying a **Compression Band** for example

then the amount of scatter will be reduced

This has further benefits, because a **Shorter** exposure time can be used, reducing movement blur and patient dose

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Compression

Alternatively the kV setting may be reduced, improving the contrast, or SNR in the image

Compression is routinely used in **Mammography** for these reasons among others

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Air Gap

The use of an increased OID, or **Air Gap**, results in:

- Magnification and a
- Reduction in scatter fraction

This is because the **Divergent** scattered rays:

- Will be increasingly less likely to strike the image receptor as the OID is increased and are therefore
- Much less likely to strike the image receptor than the primary rays

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

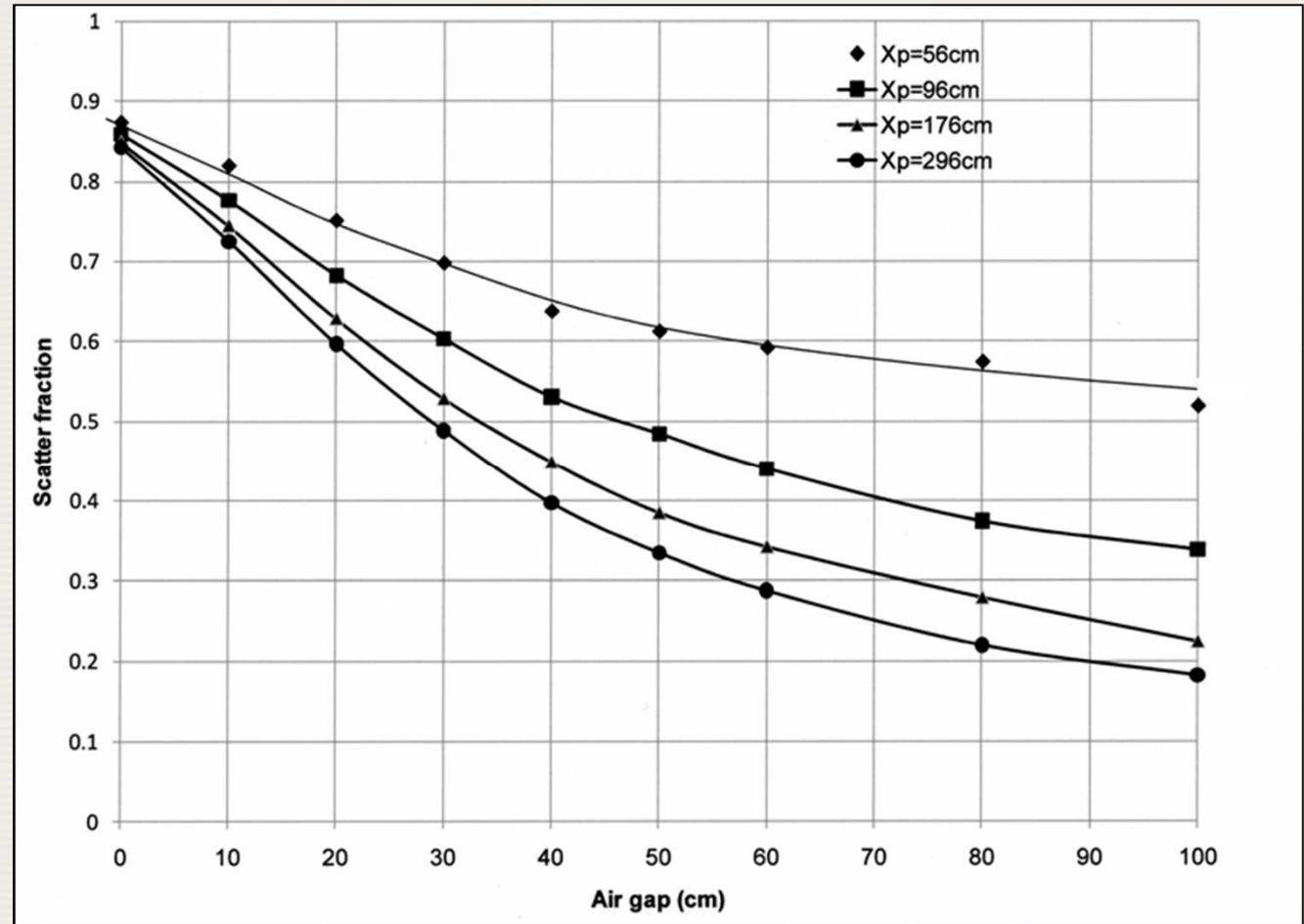
6.3.5 Other Methods for Scatter Reduction

Air Gap

Effect of the OID
on Scatter Fraction:

X_p is the source to patient
exit distance

PMMA phantom of 20 cm
thickness, X ray field size of
20 x 20 cm², 90 kV



6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Scanning Slit

Scatter Fraction depends on the X ray **Field Size**

Slot Scanning systems take advantage of this to minimize the scatter fraction by using a very small area x-ray field with a slit aperture, which must be scanned across the patient to produce the image

Such systems generally feature a

- **Pre-Patient Collimator** to define the fan-shaped beam, and a
- **Post-Patient Collimator** to intercept any scattered radiation produced

6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Scanning Slit

For **Digital** systems the area detector may be replaced by silicon or similar strip detectors

In which case the post-patient collimation is not required

These systems may feature a **Single** fan beam scanned across the region of interest or a **Number** of fan beams

The **Latter** allows for faster scanning and shorter exposure times

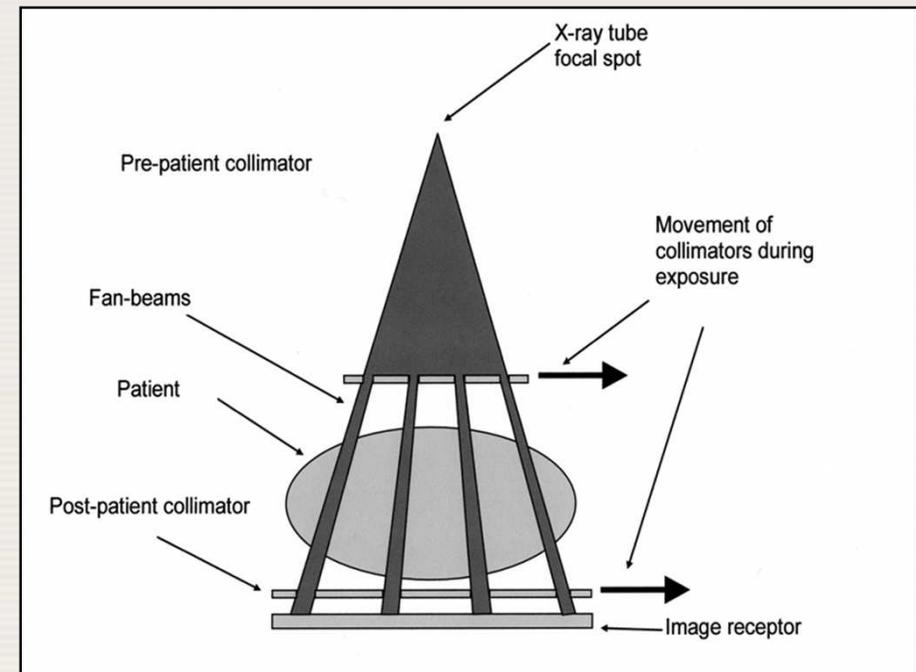
6.3 SCATTERED RADIATION IN PROJECTION RADIOGRAPHY

6.3.5 Other Methods for Scatter Reduction

Scanning Slit

Basic principle:

These systems are capable of good scatter rejection without the necessity for a grid, but require **Smooth** and precise movement of the collimator systems and **Stable** X ray generator performance



There is increased probability of artefacts compared to conventional methods, and the longer exposure time required increases the risk of movement blur in the images and will reduce the life of the XRT

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