Chapter 6: Projection Radiography


*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

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

6.2 X-Ray Image Formation
   6.2.1 Components of an Imaging System
   6.2.2 Geometry of Projection Radiography
   6.2.3 Effects of Projection Geometry
   6.2.4 Magnification Imaging
   6.2.5 Contrast Agents
   6.2.6 Dual-Energy Imaging
   6.2.7 Technique Selection

6.3 Scattered Radiation in Projection Radiography
   6.3.1 Origins of Scattered Radiation
   6.3.2 Magnitude of Scatter
   6.3.3 Effect of Scatter
   6.3.4 Methods for Scatter Reduction – Anti Scatter Grids
   6.3.5 Other Methods of Scatter Reduction

Bibliography
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.
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.
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.
For an ideal imaging task:

**Contrast, \( C = \frac{\Delta B}{B} \)**

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

For small values of \( \Delta B \), **linearity** of brightness with X ray intensity (\( I \)) is assumed, so the contrast

\[ C = \frac{\Delta I}{I} \]
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.
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
- \( \mu_d \): linear attenuation coefficient of the detail
- \( \mu_b \): linear attenuation coefficient of the background material
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.
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 $\mu$ reduce as photon energy increases.

Thus **kV** may be considered to be the contrast control where contrast is strictly the detail contrast.
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
Energy absorbed in a small region of the image receptor:

\[ E_p: \text{energy absorbed due to primary rays} \]
\[ E_s: \text{energy absorbed due to secondary rays} \]

Scatter may be quantified by:

Scatter Fraction \[ SF = \frac{E_s}{E_p + E_s} \]

Scatter to Primary Ratio \[ SP = \frac{E_s}{E_p} \]

The relationship between the two is \[ SF = ((SP^{-1})+1)^{-1} \]
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.
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.
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
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.
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 \(~50\text{ cm}\) 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
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:

\[
\frac{K_{i2}}{K_{i1}} = \left( \frac{d_{FID_2}}{d_{FID_1}} \right)^2
\]

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

\[
K_{i2} = K_{i1} \left( \frac{d_{FID_2}}{d_{FID_1}} \cdot \frac{d_{FSD_2}}{d_{FSD_1}} \right)^2
\]

Effective Dose ~ same

as increase in entrance surface beam size needed
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
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.
Since the **Magnification** of the object at the image receptor is given by:

\[ m = \frac{d_{FID}}{d_{FID} - d_{OID}} \]

the geometric unsharpness can be rewritten as:

\[ U_g = X_F \frac{d_{OID}}{d_{FID}} \]
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.
**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**

The overall unsharpness $U$ is scaled to a magnification of 1
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
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:

\[ f_{\text{new}} = f_{\text{old}} \left( \frac{m_{\text{old}} - 1}{m_{\text{old}}} \times \frac{m_{\text{new}}}{m_{\text{new}} - 1} \right) \]
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
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.
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
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.
Because of the reduction in scatter fraction, magnification may usually be performed \textbf{Without} the use of an anti-scatter grid.

This leads to a reduction in mAs in proportion to the \textbf{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).
**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.
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.
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.
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.
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.
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**
By combining these images using specific \textit{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

- \textit{Tube Voltages} and/or
- \textit{Filtrations}

are required.
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 \( \sim 1.0 \) net OD

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

\[
N = \frac{k^2}{C^2 d^2}
\]
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.
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 $kV^{-3}$.
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 \((69/60)^3 = 66\%\)

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
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.

IAEA
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.
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**
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
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.
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
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.
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
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
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)
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)
Given that the image receptor dose is proportional to \(\text{mAs}\) and to \(kV^5\), 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 \(\text{mAs}\) and a reduction by 15% to halving the \(\text{mAs}\).
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%
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.
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
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.
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**.
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.
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.
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:

\[ C_P = 1 - e^{-(\mu_d - \mu_b)\chi_d} \]
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 cm$^2$ X ray field only 19% of the **Scatter Fraction** is due to coherent scatter at 50 kV

and 9% at 120 kV
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.
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
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.
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**
  affects mainly the *Image Receptor*

  and

- **Side- and Back-Scatter**
  affects dose to *Personnel*
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°
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.
### 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
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.
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.
Consequently, the scatter fraction reaching the detector is found to show \textbf{Little} dependence on energy.

In the case of \textbf{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, \textbf{Inefficient} image receptors are more sensitive to scattered photons than to primary photons.
This is because the primary beam is incident approximately \textit{Perpendicularly} to the receptor, so the path length is similar to the phosphor thickness.

However, the scattered photons are incident \textit{Obliquely}, resulting in greater path-length on average and a greater probability of absorption. They may also be of \textit{Lower} energy, which will generally also increase the \textit{Probability} of absorption.

This effect becomes more significant as energy is increased, so the detected scatter fraction also increases with energy.
Scatter Fractions for Kodak Lanex Regular and a CaWO₄ 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.
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

\[ \sim 2 \] in the region of the mediastinum,

\[ \sim 0.6 \] in the lung fields

leading to a value of SNR per pixel behind the mediastinum of \( \sim 20 \), and

in the lung fields \( \sim 60 \)
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.
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.
Grid Construction

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

\[ \text{Grid ratio}, \quad r = \frac{h}{d} \]
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.
Grid Construction

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

- Good **Scatter Rejection** and

- High **Primary photon Transmission**
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
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.
# Grid Construction

## Typical Construction Data for Three Anti-Scatter Grids

<table>
<thead>
<tr>
<th>Grid Ratio</th>
<th>Strip Freq. (cm⁻¹)</th>
<th>h (mm)</th>
<th>d (mm)</th>
<th>t (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>10:1</td>
<td>33</td>
<td>2.5</td>
<td>0.25</td>
<td>0.050</td>
</tr>
<tr>
<td>10:1</td>
<td>40</td>
<td>2.0</td>
<td>0.20</td>
<td>0.050</td>
</tr>
<tr>
<td>12:1</td>
<td>57</td>
<td>1.56</td>
<td>0.13</td>
<td>0.045</td>
</tr>
</tbody>
</table>
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.
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
Measures of Grid Performance

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

\[
BF = \frac{\text{Exposure (mAs) required with grid}}{\text{Exposure (mAs) required without grid}} = \frac{1}{T_f}
\]

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

\[
\Sigma = \frac{T_p}{T_s}
\]
The **Contrast Improvement Factor** is given by:

\[
\text{CIF} = \frac{\text{Contrast with grid}}{\text{Contrast without grid}} = \frac{T_p}{T_t}
\]

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:

\[
\text{CIF} = \frac{(1 - \text{SF}_g)}{(1 - \text{SF}_{ng})}
\]

where, \(\text{SF}_g\) and \(\text{SF}_{ng}\) are the scatter fractions with and without the grid.
## 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

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

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.
### 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 LOCAL ABDOMEN PATIENT EQUIVALENT PHANTOM (SOFT TISSUE REGION) - 70 KV

<table>
<thead>
<tr>
<th>Grid Ratio</th>
<th>SF</th>
<th>SP</th>
<th>BF</th>
<th>CIF</th>
</tr>
</thead>
<tbody>
<tr>
<td>No grid</td>
<td>0.712</td>
<td>2.472</td>
<td>-</td>
<td>1.0</td>
</tr>
<tr>
<td>6:1</td>
<td>0.371</td>
<td>0.588</td>
<td>2.79</td>
<td>2.18</td>
</tr>
<tr>
<td>8:1</td>
<td>0.302</td>
<td>0.433</td>
<td>3.20</td>
<td>2.42</td>
</tr>
<tr>
<td>10:1</td>
<td>0.256</td>
<td>0.345</td>
<td>3.53</td>
<td>2.58</td>
</tr>
<tr>
<td>12:1</td>
<td>0.221</td>
<td>0.283</td>
<td>3.81</td>
<td>2.71</td>
</tr>
<tr>
<td>16:1</td>
<td>0.174</td>
<td>0.211</td>
<td>4.27</td>
<td>2.87</td>
</tr>
</tbody>
</table>
### 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 LOCAL ABDOMEN PATIENT EQUIVALENT PHANTOM (SPINE REGION) - 70 KV

<table>
<thead>
<tr>
<th>Grid Ratio</th>
<th>SF</th>
<th>SP</th>
<th>BF</th>
<th>CIF</th>
</tr>
</thead>
<tbody>
<tr>
<td>No grid</td>
<td>0.837</td>
<td>5.155</td>
<td>-</td>
<td>1.0</td>
</tr>
<tr>
<td>6:1</td>
<td>0.472</td>
<td>0.894</td>
<td>2.79</td>
<td>3.25</td>
</tr>
<tr>
<td>8:1</td>
<td>0.388</td>
<td>0.636</td>
<td>3.20</td>
<td>3.76</td>
</tr>
<tr>
<td>10:1</td>
<td>0.327</td>
<td>0.486</td>
<td>3.53</td>
<td>4.14</td>
</tr>
<tr>
<td>12:1</td>
<td>0.276</td>
<td>0.382</td>
<td>3.81</td>
<td>4.45</td>
</tr>
<tr>
<td>16:1</td>
<td>0.203</td>
<td>0.254</td>
<td>4.27</td>
<td>4.91</td>
</tr>
</tbody>
</table>
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.
There are several possible **Misalignments** that will lead to **Artefacts** in projection images.

Additionally, a **Damaged** grid will generate artefacts and must be replaced.
**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
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
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
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
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.
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
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 cm$^2$, 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
**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.
DOI, K. et al., Physical and clinical evaluation of new high-strip-density radiographic grids, Radiology 147 (1983) 575-582