CT Image Quality

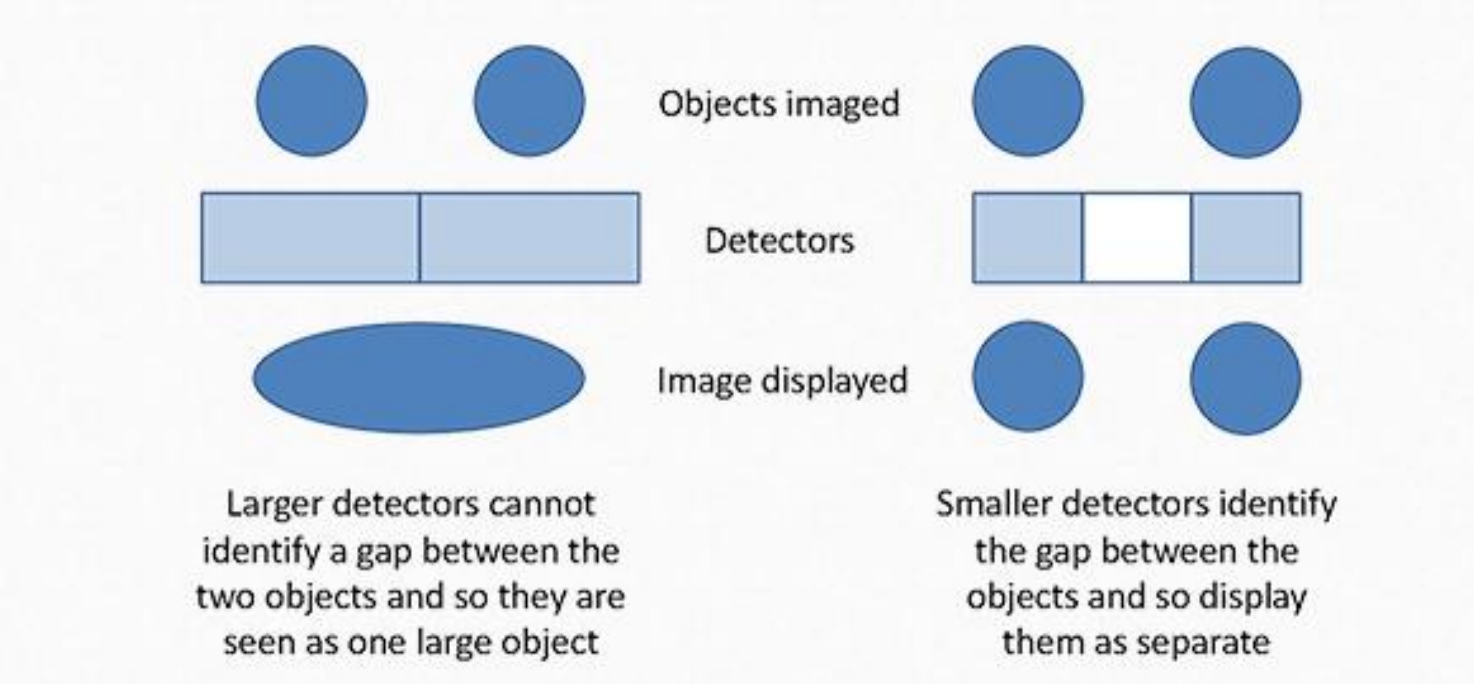
In CT scanning, image quality has many components and is influenced by many technical parameters. While image quality has always been a concern for the physics community, *clinically acceptable* image quality has become even more of an issue as strategies to reduce radiation dose – especially to pediatric patients– become a larger focus.

The image quality is mainly determined by **3 factors**:

- **Resolution**
- **Noise**
- **Contrast**

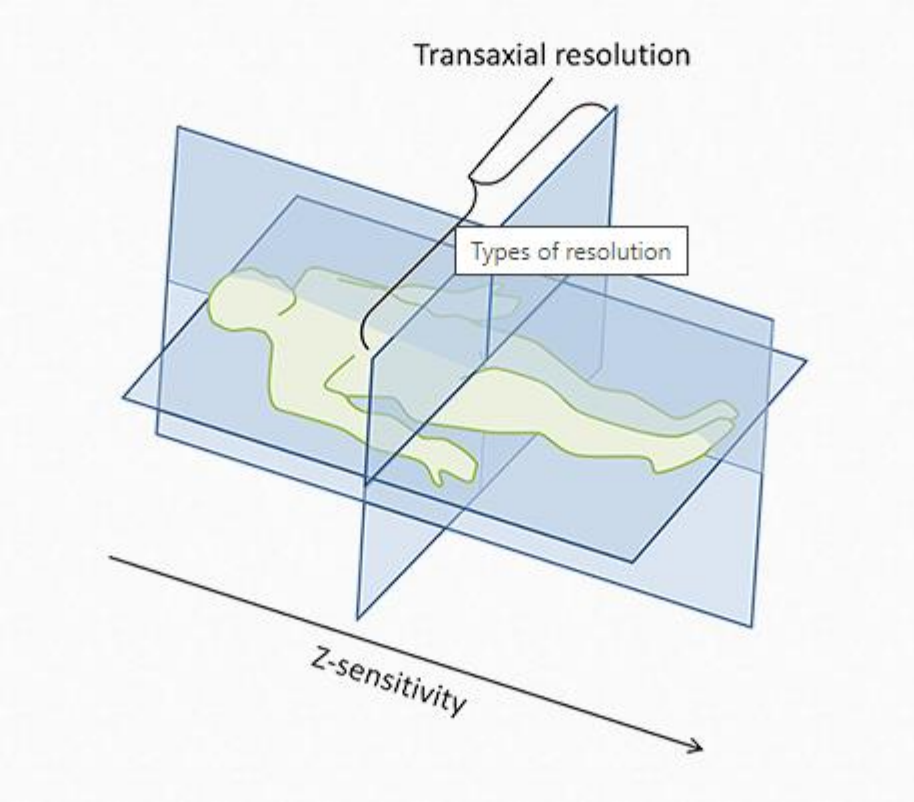
Resolution is the measure of how far apart two objects must be before they can be seen as separate details in the image. For two objects to be seen as separate the detectors must be able to identify a gap between them.

Resolution is measured in line pairs per centimeter **(lp/cm)** i.e. the number of line pairs that can be imaged as separate structures within one centimeter.



Trans-axial resolution

The minimum trans-axial resolution is determined by **the actual detector size**, however it is often quoted as the "**effective detector width**" at the isocenter of the scanner (Centre of the bore of the scanner). The "effective detector width" and the actual detector size are slightly different due to the divergence of the beam. The smaller the "effective detector width" the higher the resolution.



Scanner factors

1. Focal spot

•Size

Smaller focal spots give higher resolution, but the **max mA** is limited to prevent damage to the anode.

- There are usually two available focal spot sizes on CT scanners, for example:

  - **Fine = 0.7 mm**
  - **Broad = 1.2 mm**

•Properties

Flying focal spot: the position of the focal spot is rapidly altered in the trans-axial plane and/or the Z-axis. Each focal spot position increases the number of projections sampled and improves spatial resolution. For example, if the position of the focal spot moves in the X-Y plane, then the in-plane resolution increases.

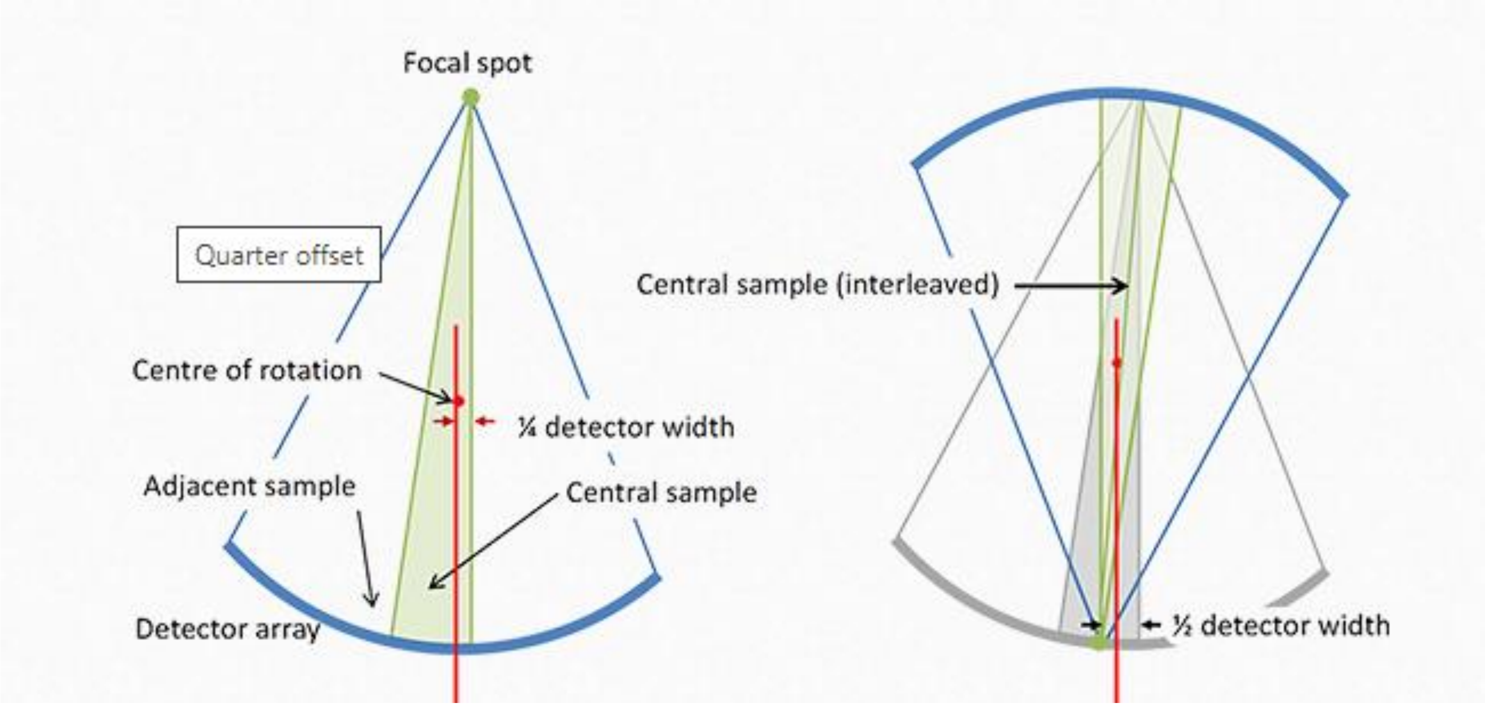
- **Focus-detector distance (FDD)**
- **Focus-iso-centre distance (FID)**

2. Detector size

Smaller detectors give higher resolution but more detectors within an area also means more partitions (dead space) and a reduced overall detection efficiency.

3. Detector design properties

Quarter ray detector offset: the centre of the detector array is offset from the centre of rotation by one quarter the width of an individual detector. As the gantry rotates to 180° the centre of the detector array is now offset by half the width of a detector giving an interleaved sampling of the patient.



Scan parameters

1. Number of projections

•Larger number of projections gives finer resolution (up to a point).

2. Reconstruction filter

•Higher resolution or "sharp" kernels (e.g. bone reconstruction) have better spatial resolution than soft kernels (e.g. soft tissue reconstruction).

•However, higher resolution kernels do not average out high spatial frequency signals and therefore produce more noise.

3. Pixel size

The pixel size (d) in mm is give by the equation: **d = FOV/n……(1)** where: **FOV** = field of view (mm)
n = image matrix size

The highest spatial frequency that can be obtained (fmax) is called the Nyquist limit and is given by:

fmax = 1/2d…….(2)

•From this equation you can see that the higher the pixel size, the lower the maximum spatial frequency.

•To improve spatial frequency we can:

- Reduce the field of view (**smaller FOV = smaller pixel size as seen in the first equation**). We can do this retrospectively by a targeted reconstruction of the original data into a small field of view.
- Increase the matrix size (**larger n = small pixel size as seen in the first equation**)

Factors Affecting Spatial Resolution

Focal Spot	Smaller focal spot, SR improves
Detector width	Smaller Detector Width, SR improves
Number of Projections	More projections, SR improves
Slice thickness	Smaller ST, SR improves
Pitch	Lower pitch, SR improves
Pixel Size	Smaller Pixel Sixe, SR improves
FOV	Decreasing FOV(everything else constant), SR improves
Patient Motion	Decreased Patient motion, SR improves

Pixel Size

- the spatial resolution can be no greater than the size represented by the pixel length.
- In reality, pixel size should be 1.5 to 2 times smaller than the desired resolution.
- Unless a matrix element exactly coincides with an object, the object representation will be averaged over two or more pixels and thus may not be visualized.
- It must be realized that the pixel size refers to the FOV (or body), not the viewing screen or film.

Contrast Resolution

- ability to differentiate the attenuation coefficients of adjacent areas of tissue.

- In the computation of any single pixel value, there is error in the form of statistical variation; it is this variation that limits the ultimate contrast resolution.
- This variation (called *image noise*) is manifested as a grainy background, or mottle.
- The parameter used to evaluate this variation is the **standard deviation (SD).**

Factors Affecting Contrast Resolution

mAs	More mAs, CR improves
Pixel Size	FOV and pixel size increase, CR improves
Slice thickness	ST increases, CR improves
Reconstruction filter	Using Soft tissue improves CR
Patient Size	For larger patients, at same technique, more attenuation, detected photons decreases, CR degrades.

Temporal resolution

- refers to the ability of a CT scanner to capture objects that change shape or position over time and depends primarily on the gantry rotation speed and the reconstruction method used.

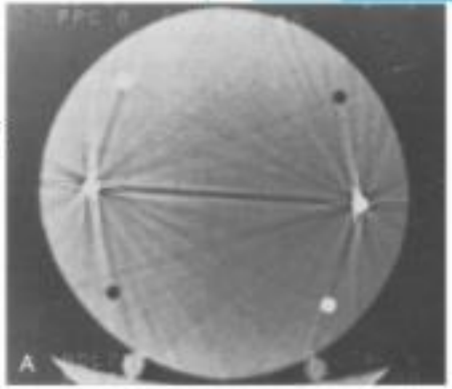
- depends on:

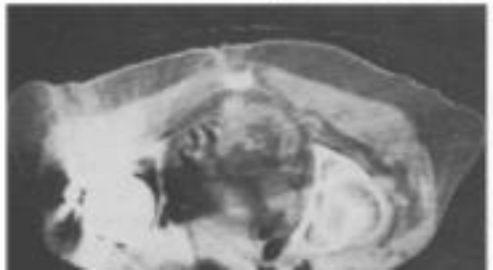
  1. gantry rotation speed
  2. spiral interpolating algorithm used during reconstruction.

Metal and Bone Artifacts

- The presence of objects having an exceptionally high or low attenuation can create artifacts by forcing the detector to operate in a nonlinear response region.

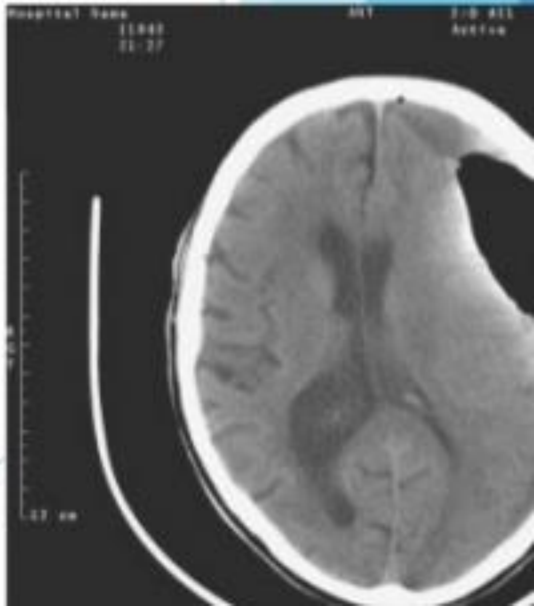
- Because this incorrect response occurs at specific directions of the beam through the object, incomplete cancellation of the back-projected rays during reconstruction occurs and yields streaking artifacts.





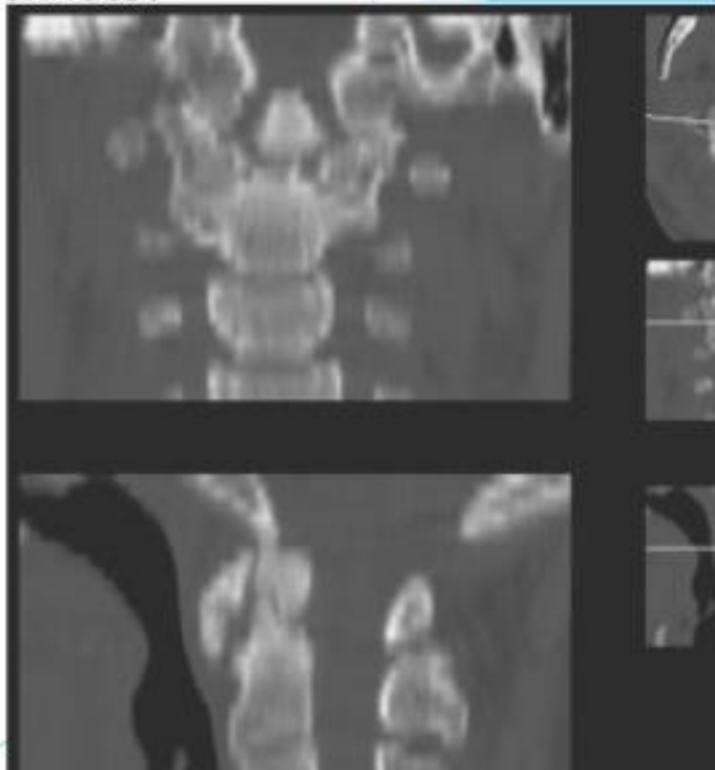
Beam-Hardening artifacts

- result from the preferential absorption of low-energy photons from the beam.
- average beam photon energy is progressively increased.
- toward the end of the x-ray path, the attenuation is less than at the beginning because the attenuation coefficient is smaller with higher energy.
- The reconstruction program, however, assumes a monochromatic beam and attributes any change in beam intensity to a change in tissue composition rather than to the result of a shift in average photon energy.
- The assigned attenuation coefficients are thus in error, and the densities seen on the image are in error.
- The effect is most pronounced in regions of large attenuation, such as bone.



Stair-Stepping Artifact

- occur when in one direction the pixel of the reformatted image has the same length as the axial image
  but in the other direction the pixel length is the same as the slice thickness.
- Because pixel length in most scans is considerably smaller than slice thickness, the reformatted scan has an unusual appearance.
- Uncommon in modern CT.



Z-sensitivity

Z-sensitivity refers to the effective imaged slice width.

Factors affecting z-sensitivity

1. Detector slice thickness

•The wider (in the z-axis) the detector row, the lower the resolution

2. Overlapping samples

•Acquiring the data using overlapping slices can improve Z-sensitivity. This is achieved by using a low spiral pitch i.e. pitch <1.

3. Focal spot

•A fine focal spot improves the z-sensitivity

- Reduced partial volume effect
- Better multi-planar reformatting
- Improved volume rendering e.g. displaying 3D representations of the data (e.g. cardiac imaging, vascular imaging, CT colonography etc)

Importance of slice thickness

1. Noise

•The thinner the slice the better the resolution BUT the worse the noise

2. Partial volume effect

•Thicker slices increase the partial volume effects

3. Isotropic scanning

•Thin slices allow isotropic scanning, i.e. the pixels in the axial and the z-axis are the same size (cubes).

Noise

Even if we image a perfectly uniform object (e.g. a water filled object) there is still a variation in the Hounsfield units about a mean. This is due to noise. Noise degrades the image by degrading low contrast resolution and introducing uncertainty in the Hounsfield units of the images.

We can measure noise in any uniform region of the image e.g. with a water phantom. The standard deviation of the Hounsfield Units in a selected region-of-interest gives the mean noise measurement.

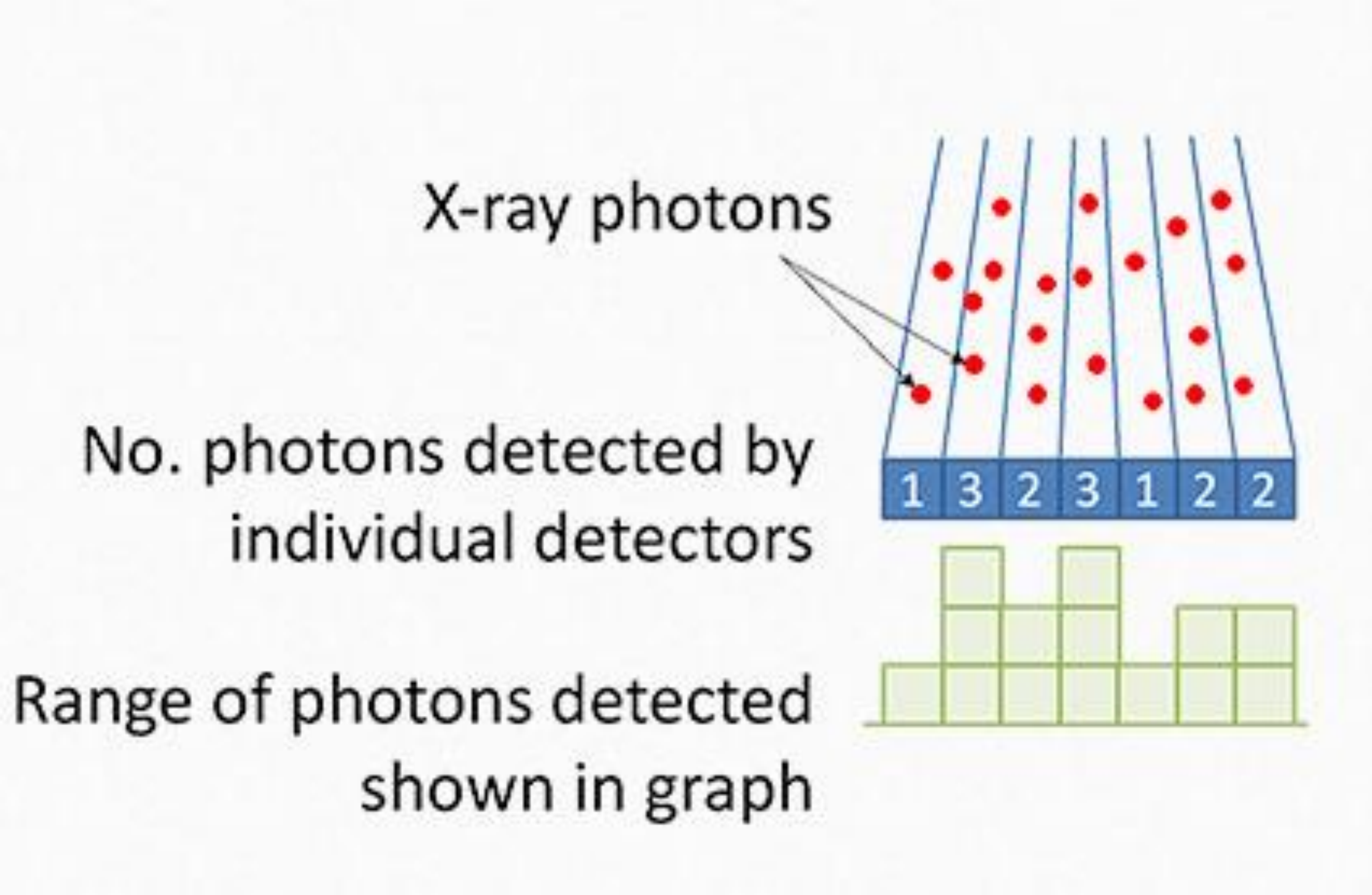
There are three sources of noise:

1.Quantum noise

2.Electronic noise

3.Noise introduced by the reconstruction process e.g. back-projection.

Stochastic noise



This is the dominant source of noise in an image. Photon registration by the detectors is a stochastic process. The number of photons detected will vary randomly about a mean value and that variation is the noise. The noise in the final image is given by:

Noise (standard deviation) $\propto 1/\sqrt{(\text{no. of photons})}$

From this equation we can say that increasing the number of photons reduces the amount of noise and, therefore, anything that increases the number of photons (increases the photon flux) will reduce the noise. If we double the number of photons we will reduce the noise by $\sqrt{2}$ (i.e. increasing the number of photons by a factor of 4 will halve the noise).

Doubling the number of photons can be achieved by:

- Doubling the tube current (mA)
- Doubling the rotation time (s)
- Doubling the slice thickness (mm)

Increasing the tube kilovoltage (kV) also increases the photon flux but it is not directly proportional (output is approximately $\propto kV^2$).

Thank You

References :-

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- Three-dimensional volume rendering of spiral CT data: theory and methods. Radiographics 1999;745-764.
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