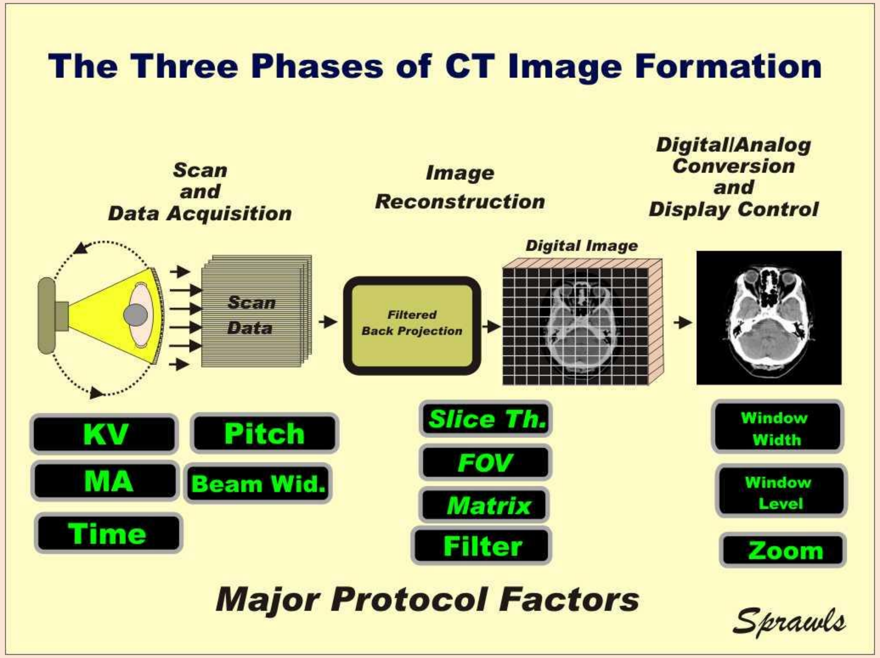
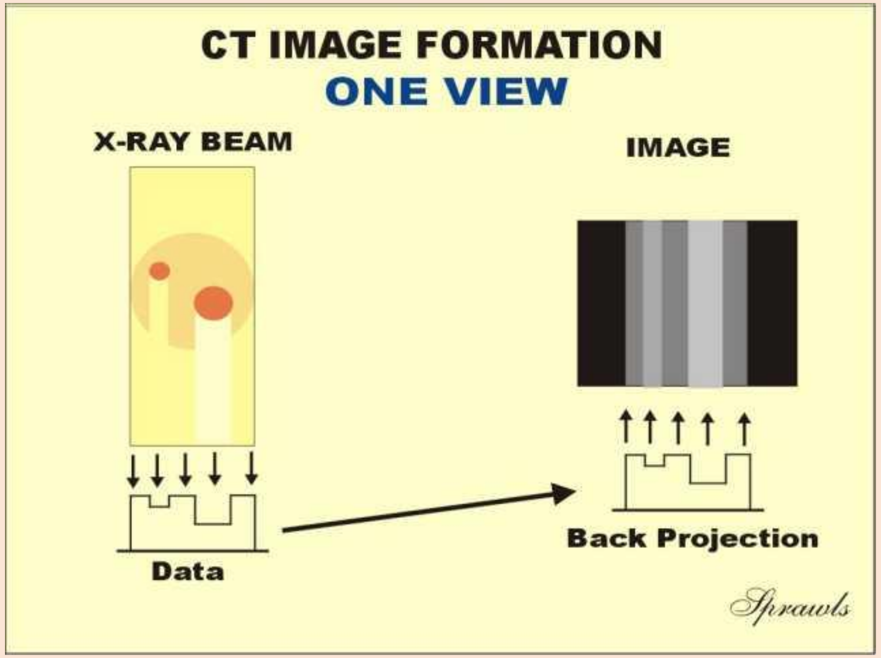
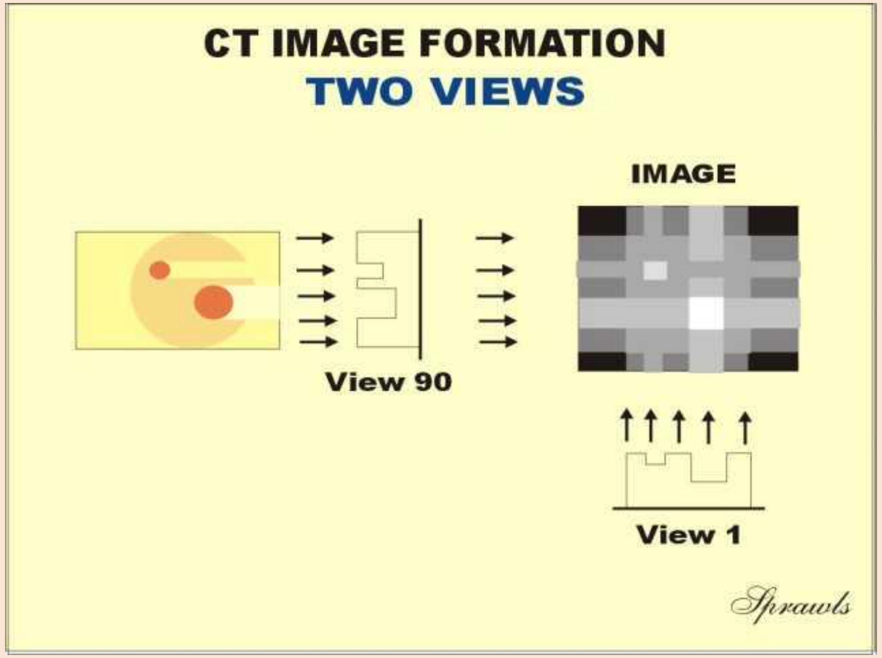
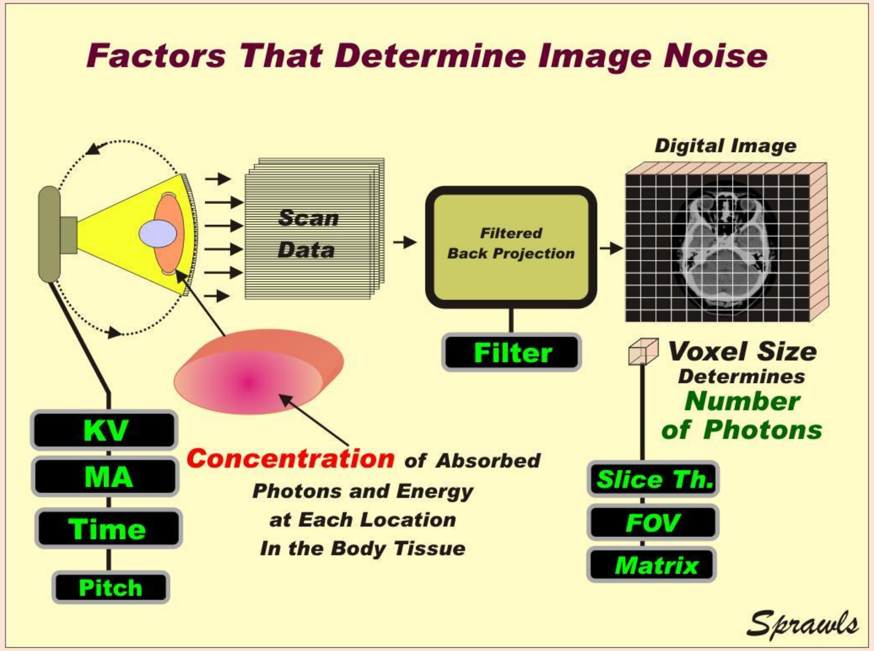
Σημαντικές χρονολογίες στην εξέλιξη της Υπολογιστικής Τομογραφίας

- 1924 - μαθηματική θεωρία τομογραφικής ανακατασκευής δεδομένων (Johann Radon)
- 1930 - κλασσική τομογραφία (A. Vallebona)
- 1963 - θεωρητική βάση της Υ.Τ. (A. McLeod Cormack)
- 1971 - 1ος εμπορικός αξονικός τομογράφος – CT (Sir Godfrey Hounsfield)
- 1974 - 1ος CT 3ης γενεάς
- 1979 - Nobel price (Cormack & Hounsfield)
- 1989 - CT μονής τομής
- 1994 - διτομικός ελικοειδής CT
- 2001 - 16-τομών ελικοειδής CT
- 2007 - 320-τομών ελικοειδής CT

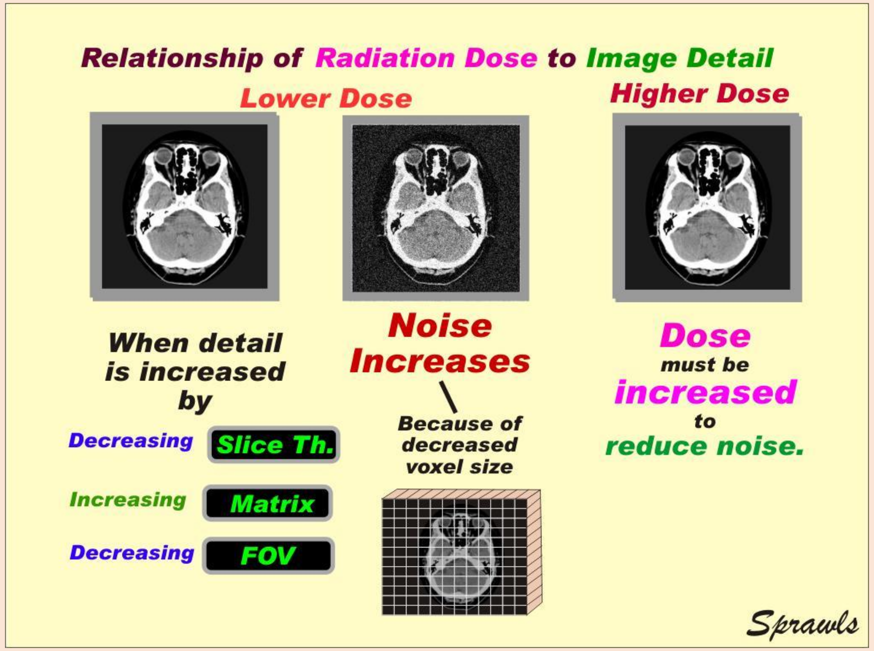

The Three Phases of CT Image Formation
Scan
and
Data Acquisition
Image
Reconstruction
Digital/Analog
Conversion
and
Display Control
Scan
Data
Filtered
Back Projection
Digital Image
KV
Pitch
MA
Beam Wid.
Time
Slice Th.
FOV
Matrix
Filter
Window
Width
Window
Level
Zoom
Major Protocol Factors
Sprawls


CT IMAGE FORMATION
ONE VIEW
X-RAY BEAM
IMAGE
Data
Back Projection
Sprawls

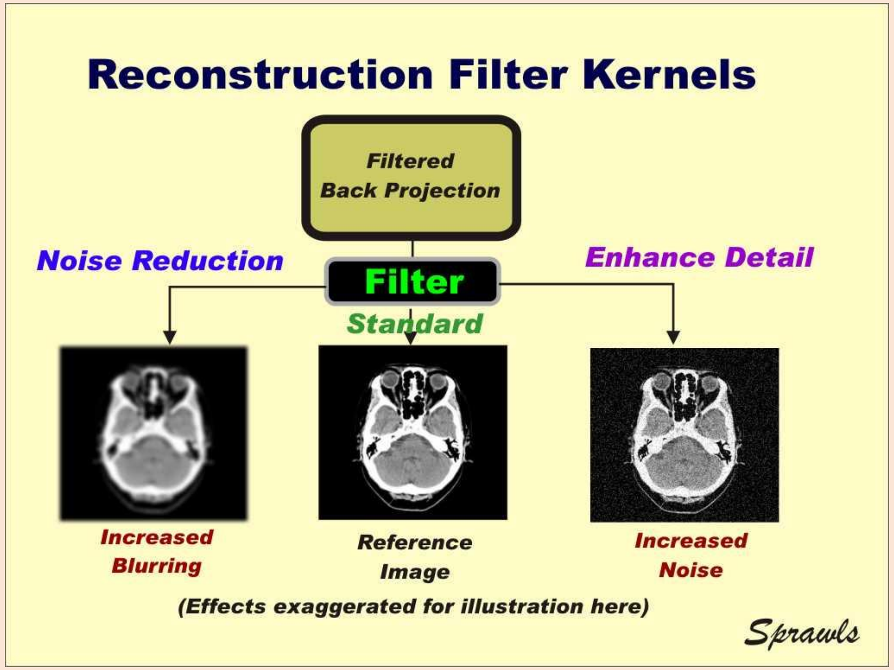

CT IMAGE FORMATION
TWO VIEWS
IMAGE
View 90
View 1
Sprawls



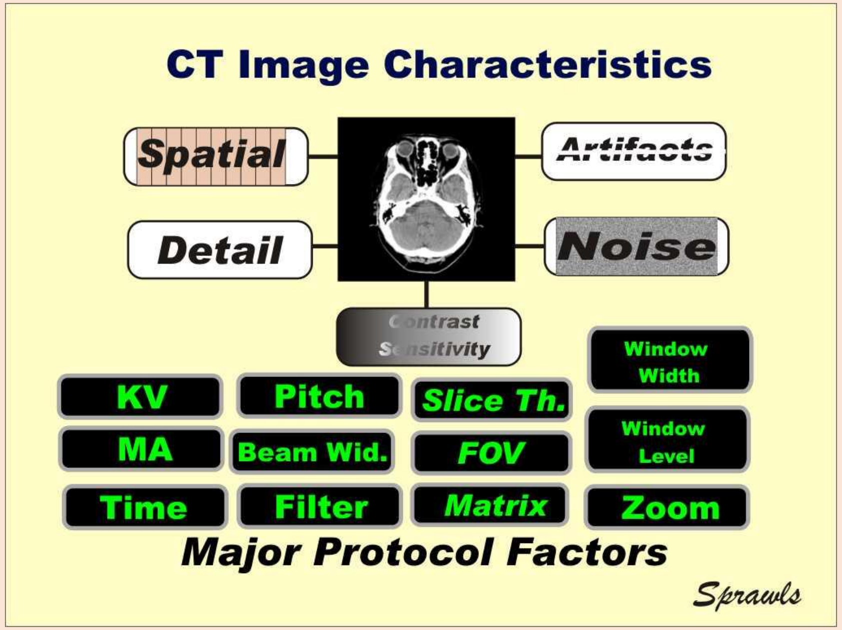
Factors That Determine Image Noise
Scan Data
Filtered Back Projection
Digital Image
Filter
Voxel Size Determines Number of Photons
KV
MA
Time
Pitch
Concentration of Absorbed Photons and Energy at Each Location In the Body Tissue
Slice Th.
FOV
Matrix
Sprawls

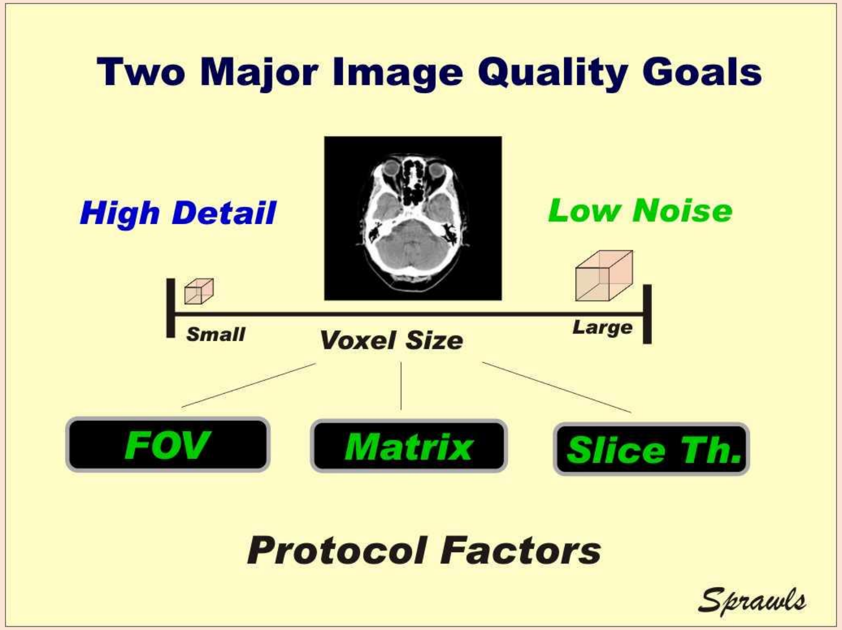


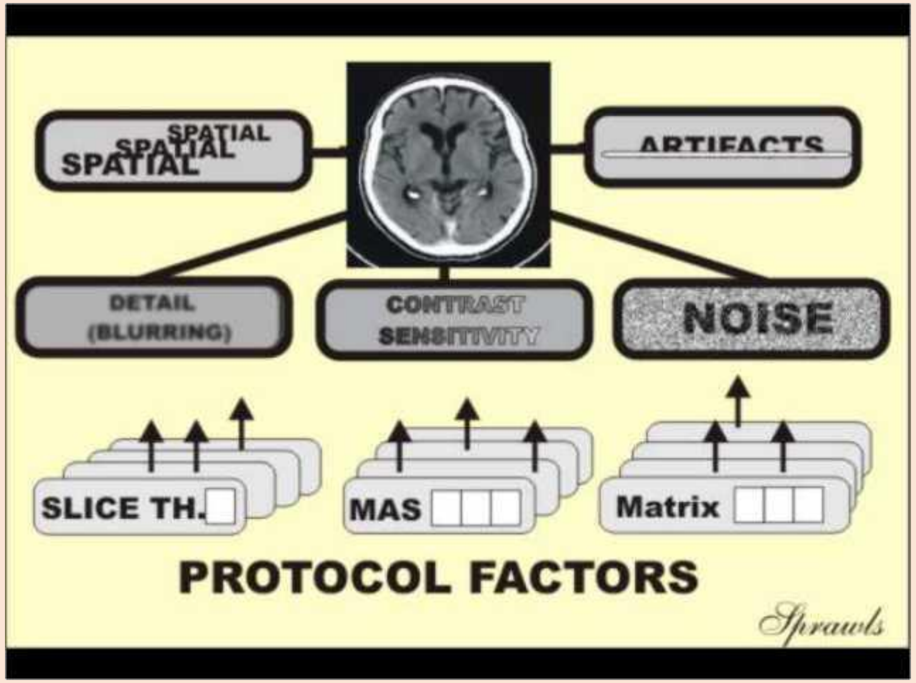
Relationship of Radiation Dose to Image Detail
Lower Dose
Higher Dose
When detail is increased by
Decreasing Slice Th.
Increasing Matrix
Decreasing FOV
Noise Increases
Because of decreased voxel size
Dose must be increased to reduce noise.
Sprawls



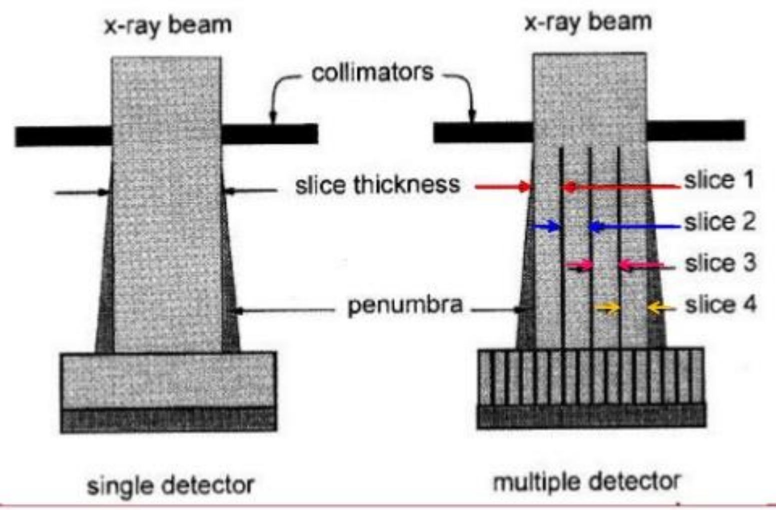
Reconstruction Filter Kernels
Filtered Back Projection
Noise Reduction
Filter
Enhance Detail
Standard
Increased Blurring
Reference Image
Increased Noise
(Effects exaggerated for illustration here)
Sprawls


CT Image Characteristics
Spatial
Detail
Artifacts
Noise
Contrast
Sensitivity
Window
Width
Window
Level
KV
Pitch
Slice Th.
MA
Beam Wid.
FOV
Time
Filter
Matrix
Zoom
Major Protocol Factors
Sprawls


Two Major Image Quality Goals
High Detail
Low Noise
Small
Voxel Size
Large
FOV
Matrix
Slice Th.
Protocol Factors
Sprawls


SPATIAL
ARTIFACTS
DETAIL
(BLURRING)
CONTRAST
SENSITIVITY
NOISE
SLICE TH.
MAS
Matrix
PROTOCOL FACTORS
Sprawls

Slice Thickness: Multiple Detector Array Scanners



Detector Pitch & Collimator Pitch

Pitch is a parameter that comes to play when helical scan protocols are used.

In a helical CT scanner with one detector array, the pitch is determined by the collimator (collimator pitch).

Single and Multiple detector arrays scanner have different definitions.

Detector Pitch & Collimator Pitch

- ❖ In a helical CT scanner with one detector array, the pitch is determined by the collimator (collimator pitch),

$$\text{Collimator pitch} = \frac{\text{table movement (mm) per 360-degree rotation of gantry}}{\text{collimator width (mm) at isocenter}}$$

$$\text{Detector pitch} = \frac{\text{table movement (mm) per 360-degree rotation of gantry}}{\text{detector width (mm)}}$$

Detector Pitch & Collimator Pitch

- ❖ Pitch fundamentally influences radiation dose to the patient, image quality, and scan time.

- ❖ For single detector array scanners, a pitch of 1.0 implies that the number of CT views acquired, when averaged over the long axis of the patient, is comparable to the number acquired with adjoining axial CT.

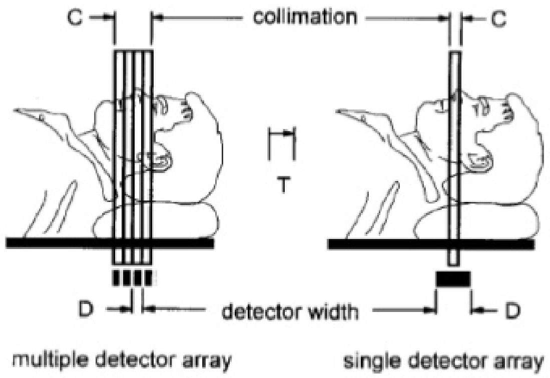
Detector Pitch & Collimator Pitch

- A pitch of less than 1.0 involves over-scanning may result in some slight improvement in image quality and a higher radiation dose to the patient.
- pitches greater than 1.0, and pitches up to 1.5 are commonly used.

Detector Pitch & Collimator Pitch

Collimator Pitch = T / C

Detector width = T / D



For a multiple detector array CT scanner with four detector arrays, a collimator pitch of 1.0 is equal to a detector pitch of 4.0.

Detector Pitch & Collimator Pitch

- Scanners that have multiple detector arrays require a different definition of pitch.

$$\text{Collimator pitch} = \frac{\text{Detector pitch}}{N}$$

N = Number of detectors used

Detector Pitch & Collimator Pitch

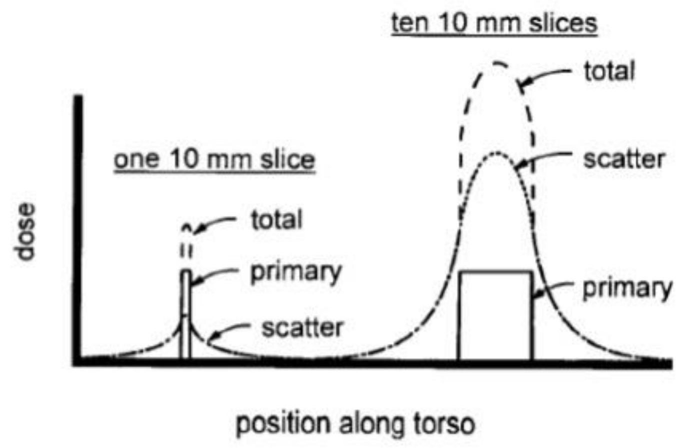
For scanners with four detector arrays, detector pitches running from 3 to 6 are used. A detector pitch of 3 for a four-detector array scanner is equivalent to a collimator pitch of 0.75 (3/4), and a detector pitch of 6 corresponds to a collimator pitch of 1.5 (6/4).

CT Dose Measurement

- Compton scattering is the principal interaction mechanism in CT, so the radiation dose attributable to scattered radiation is considerable, and it can be higher than the radiation dose from the primary beam.

- The multiple scan average dose (MSAD) is the standard for determining radiation dose in CT.

CT Dose Measurement Methods



CT Dose Measurement

- The MSAD is the dose to tissue that includes the dose attributable to scattered radiation emanating from all adjacent slices.

- The MSAD is defined as the average dose, at a particular depth from the surface, resulting from a large series of CT slices.

CT Dose Measurement Methods

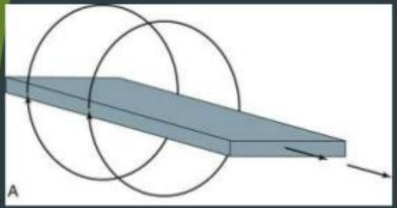
- Same as previous but a correction factor is needed,

$$\text{Dose (helical)} = \text{Dose (axial)} \times \frac{1}{\text{Collimator pitch}}$$

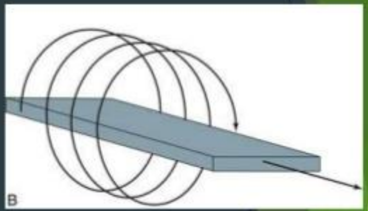
$$\text{Dose (fluoro)} = \text{CTDI dose} \times \frac{\text{Time} \times \text{Current}}{\text{mAs of CTDI measurement}}$$

HELICAL CT

- A technique where by the patient is transported continuously through the gantry while data are acquired continuously during several 360 degrees scans.
- Also k/a spiral or volume CT.



Slice by slice CT



Spiral/helical CT

Advantages of helical CT

- Increased speed a study.
- Exact continuity of images.
- Less slice misregistration.
- Less motion artifact.
- Need for less contrast medium.
- Availability of volumetric data.

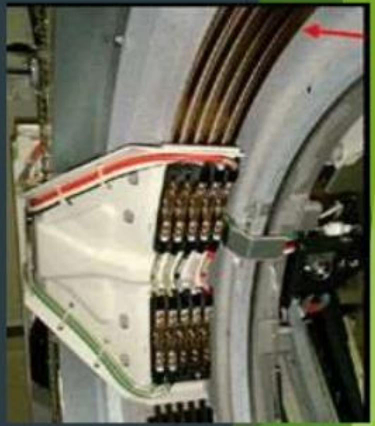
Advances of spiral CT over conventional CT

1. Slip ring device
2. More efficient tube cooling
3. Increased milliampere capability
4. Smoother table movement
5. Software adjustment for table movement
6. Efficient detectors

Slip-ring technology

- Is one of the important parts of spiral CT machines.
- Serves as a connection of the gantry rotating part and the fixed portion of the power line and the signal line.

Slip-ring technology

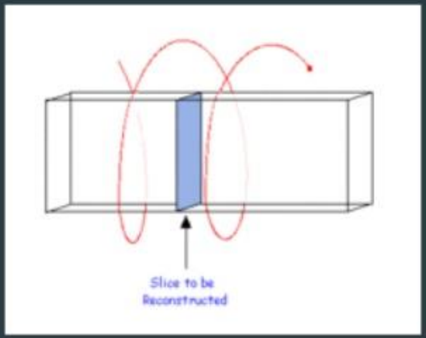


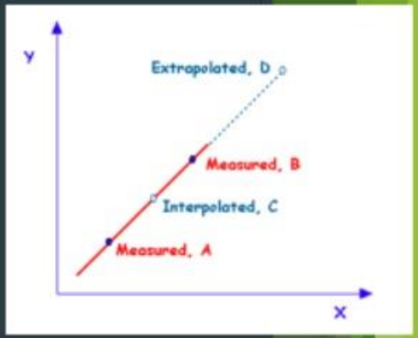
Adjustment in reconstruction algorithm

▶ Interpolation algorithm

- Estimation of a unknown value between known values is called **interpolation**.
- Estimation of a unknown value beyond known values is called **extrapolation**.

Interpolation





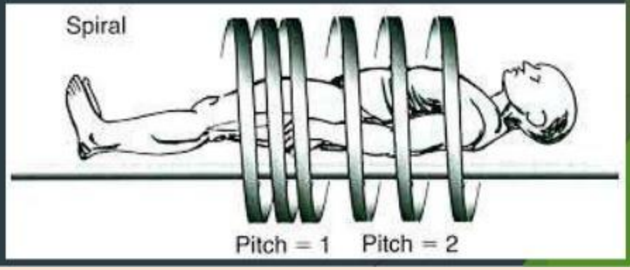
❖ Pitch: relation of table speed to slice thickness

Pitch= Table feed * time per rotation of 360

collimation

Pitch= Table feed , if it is a 1 sec scan.

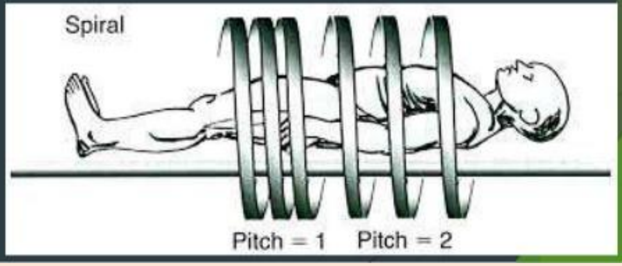
collimation



❖ <u>Pitch:</u> relation of table speed to slice thickness

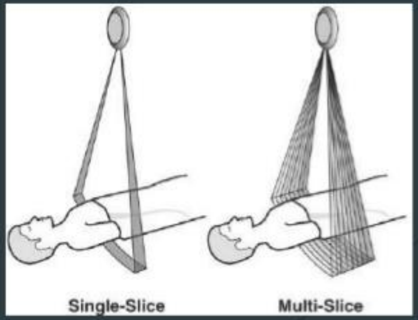
$$\text{Pitch} = \frac{\text{Table feed} * \text{time per rotation of } 360}{\text{collimation}}$$

$$\text{Pitch} = \frac{\text{Table feed}}{\text{collimation}}\text{, if it is a 1 sec scan.}$$



Multislice CT (MSCT) or Multi-detector CT (MDCT)

- Uses multiple parallel detectors instead of a single detector array.
- And hence greater volume of patient is scanned in a single rotation



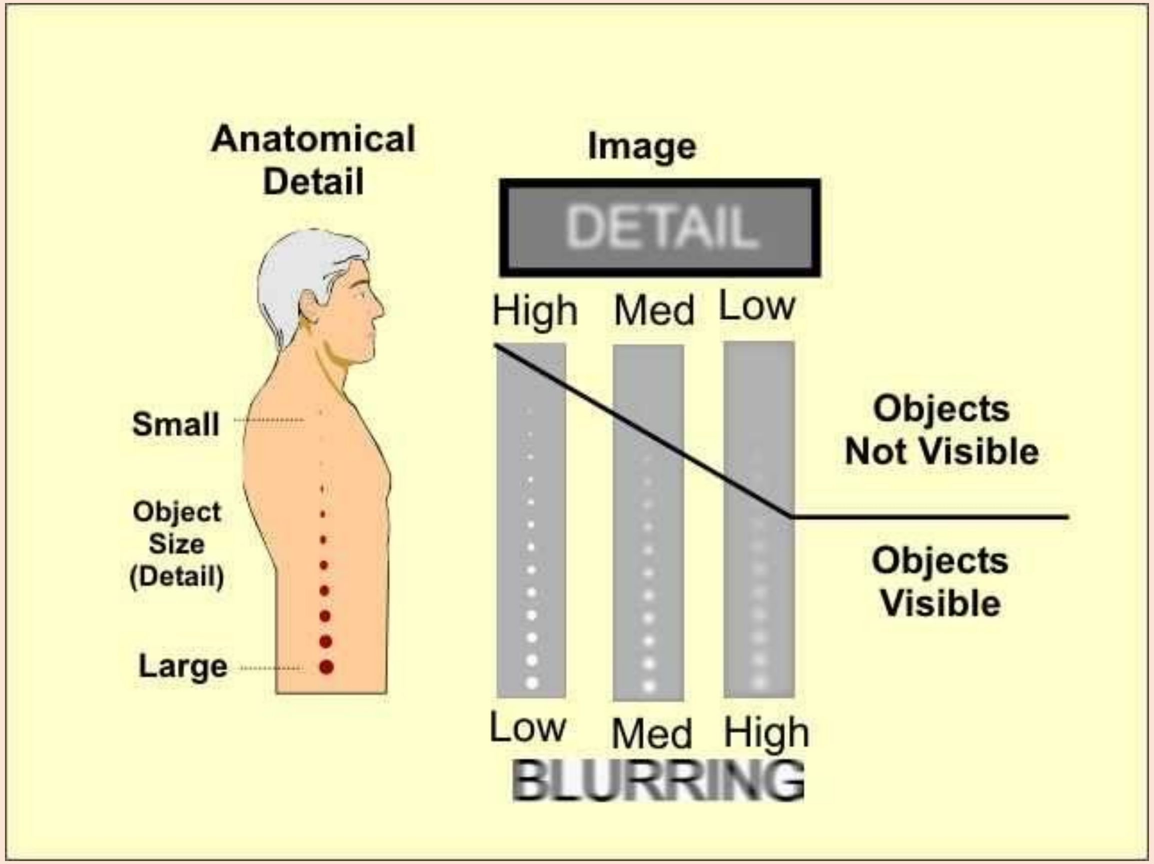
Advantages of MSCT over single slice

- Same acquisition in shorter time.
- Improved z-axis resolution.
- Reduced artifacts.
- Increased coverage per rotation.
- Reduced overall tube loading.
- More information available for the radiologists.
- Possibility of new applications.

Ποιότητα εικόνας CT : Ασάφεια

Η ασάφεια μειώνει την ορατότητα λεπτομερειών (μικρά αντικείμενα και χαρακτηριστικά. Κάθε ιατρική απεικονιστική μέθοδος διαθέτει πηγές ασάφειας που μειώνουν την ανάδειξη μικρών λεπτομερειών και καθορίζουν τους τύπους των διαγνωστικών διαδικασιών στους οποίους μπορούν να χρησιμοποιηθούν.

Για παράδειγμα, η ακτινογραφία που χαρακτηρίζεται από χαμηλά επίπεδα ασάφειας και προσφέρει μεγάλη ανάδειξη λεπτομερειών χρησιμοποιείται για την ανάδειξη μικρών οστικών καταγμάτων. Στην CT υπάρχουν αρκετές πηγές ασάφειας που συνδυαστικά μειώνουν την διακριτική ικανότητα λεπτομερειών.
Το ζήτημα είναι ότι μειώνοντας την ασάφεια, αυξάνουμε τον οπτικό θόρυβο και μπορεί να οδηγήσει σε αυξημένη δόση στον ασθενή.



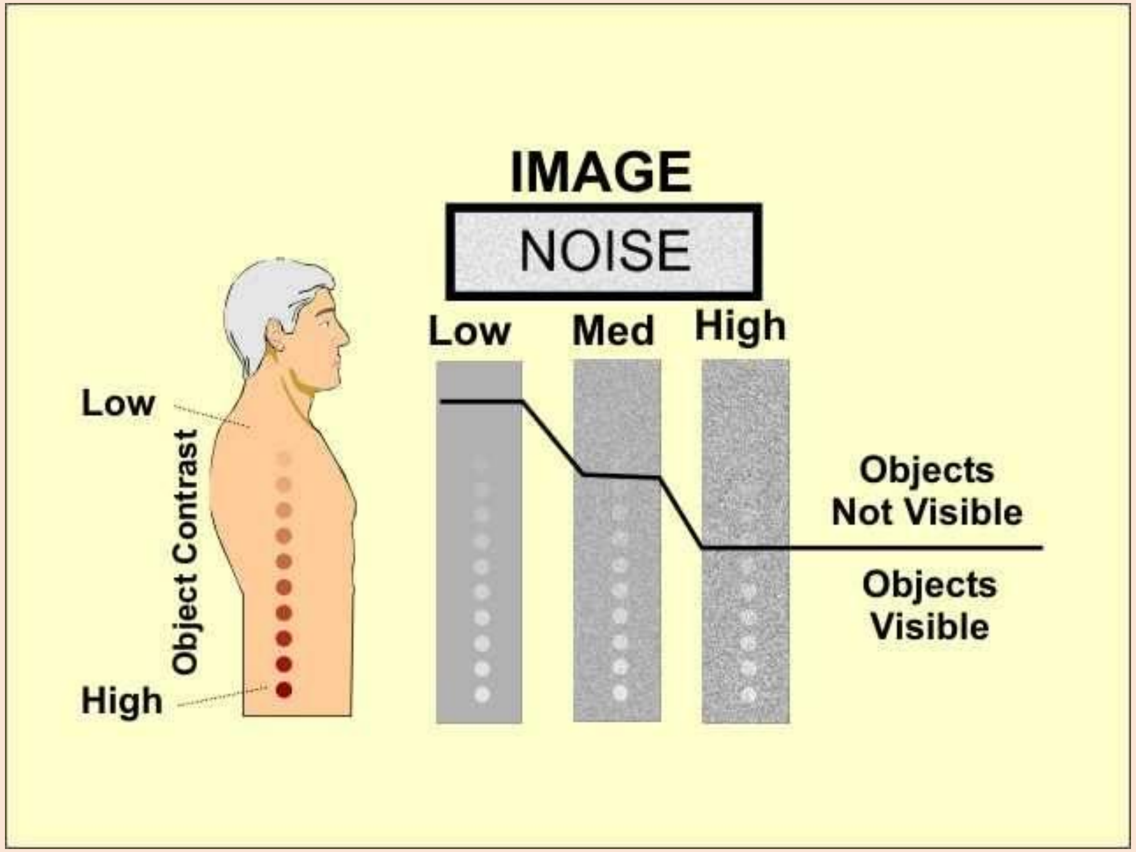
Ποιότητα εικόνας CT : Θόρυβος

Ο θόρυβος είναι ανεπιθύμητο απεικονιστικό χαρακτηριστικό που μειώνει τη διακριτικότητα συγκεκριμένων τύπων δομών.

Ειδικότερα, ο θόρυβος μειώνει την διακριτικότητα δομών χαμηλής αντίθεσης (low contrast objects). Στην CT ο θόρυβος έχει μεγάλη σημασία διότι η CT χρησιμοποιείται για την απεικόνιση χαμηλής αντίθεσης διαφορών μεταξύ ιστών.

Διαφορά μεταξύ Θορύβου και Ασάφειας. Και τα δύο χαρακτηριστικά μειώνουν την απεικόνιση διαφορετικών τύπων δομών. Ο Θόρυβος μειώνει την ορατότητα-απεικόνιση δομών χαμηλής αντίθεσης, ενώ η Ασάφεια μειώνει την απεικόνιση μικρών δομών ή λεπτομερειών.

Ο Θόρυβος της εικόνας μειώνεται τροποποιώντας παράγοντες των πρωτοκόλλων. Βέβαια, η τροποποίηση των παραγόντων αυτών έχει ως πιθανό αποτέλεσμα την αύξηση της ασάφειας ή/και την αύξηση της δόσης στον ασθενή.

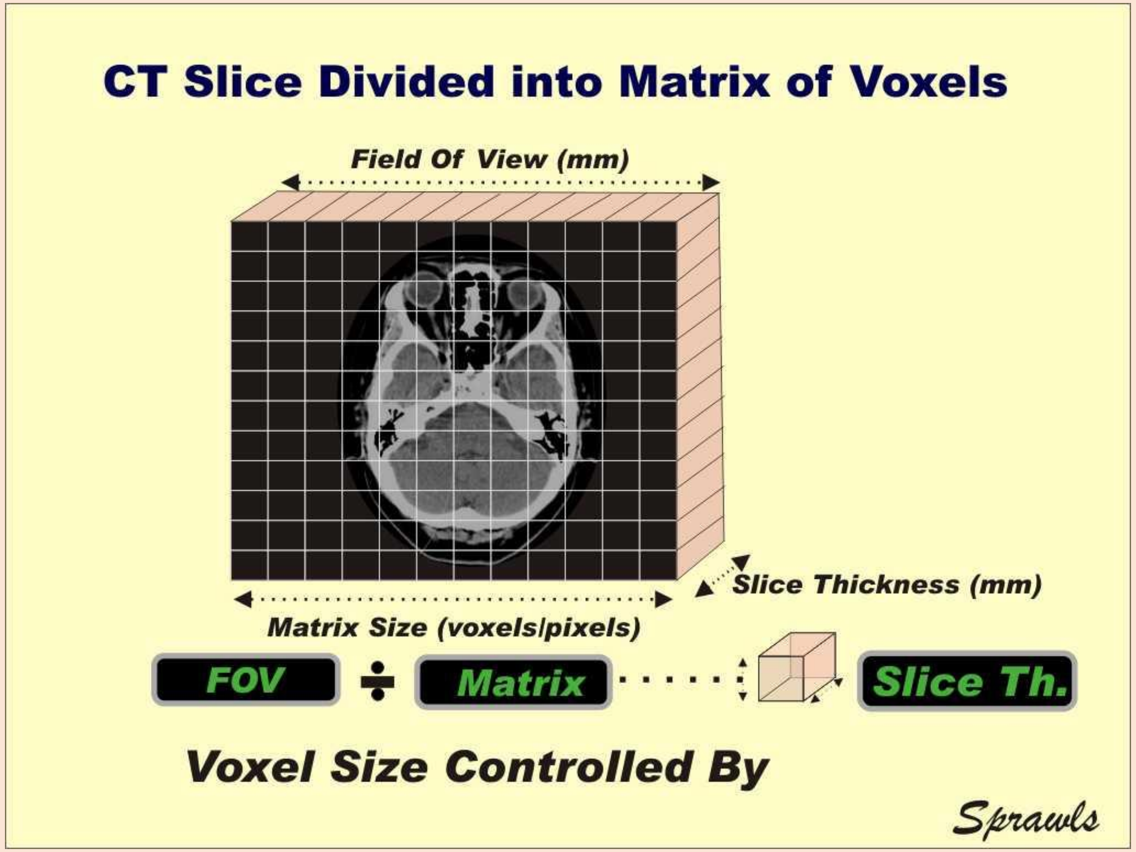


Ποιότητα εικόνας CT : Χωρικά & Γεωμετρικά χαρακτηριστικά

Τα χωρικά και γεωμετρικά χαρακτηριστικά της CT εικόνας συμβάλουν σημαντικά στη βελτιστοποίηση την απεικονιστικών πρωτοκόλλων. Αυτό οφείλεται στο γεγονός ότι η εικόνα αποτελείται από χιλιάδες-εκατομμύρια μικρά εικονοστοιχεία (voxels – volume elements).

Μία τυπική CT εικόνα αποτελεί, συνήθως, εγκάρσια τομή του σώματος. Κατά τη διάρκεια της φάσης ανακατασκευής εικόνας , κάθε τομή διαιρείται σε μία μήτρα voxels (ογκοστοιχείων). Το μέγεθος των voxels έχει σημαντική επίδραση στην ασάφεια και το θόρυβο εικόνας και στη δόση στον εξεταζόμενο.

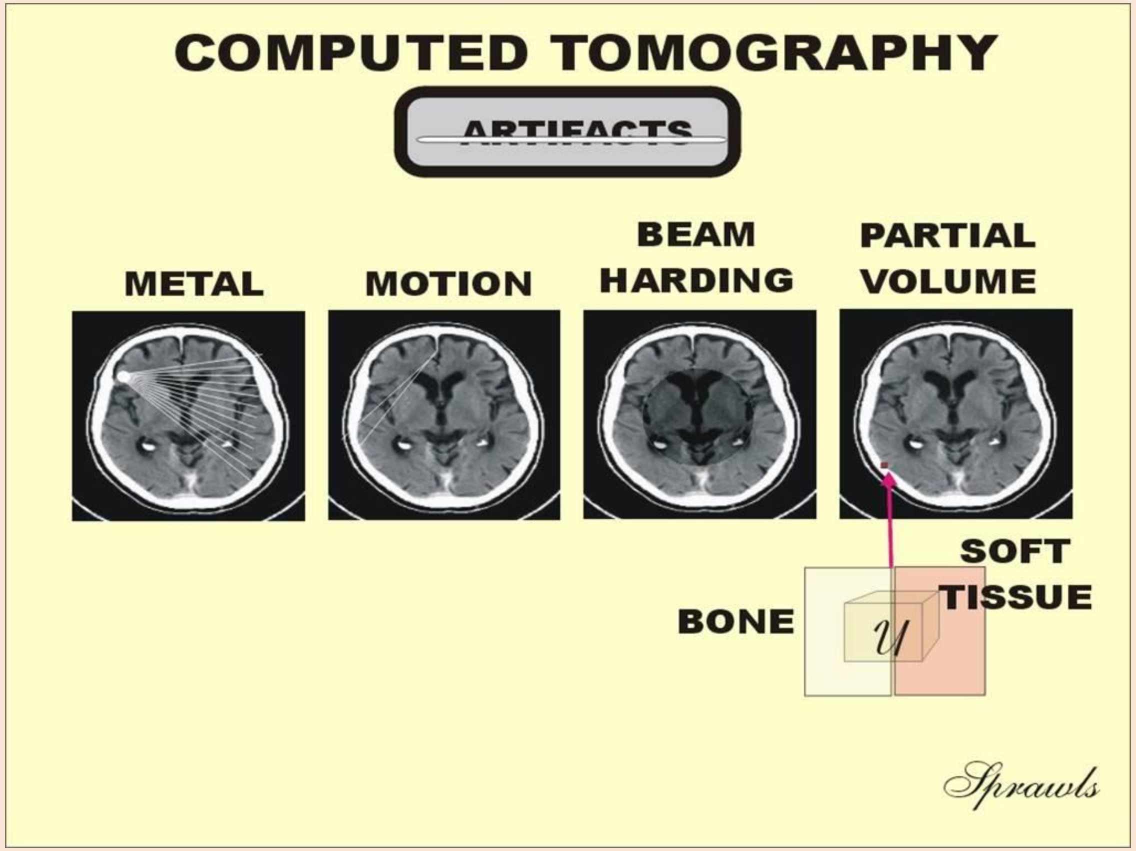
Η διάσταση των voxels καθορίζεται από τρεις παράγοντες : FOV (mm), διαστάσεις μήτρας (voxels/pixels) και πάχος τομής (mm).



Ποιότητα εικόνας CT : Ψευδοεικόνες - Artifacts

Ψευδοεικόνα είναι κάτι που εμφανίζεται σε μία εικόνα-ακτινογραφία το οποίο δεν αποτελεί οπτικοποίηση ενός πραγματικού αντικειμένου ή δομής στο σώμα του εξεταζόμενου.

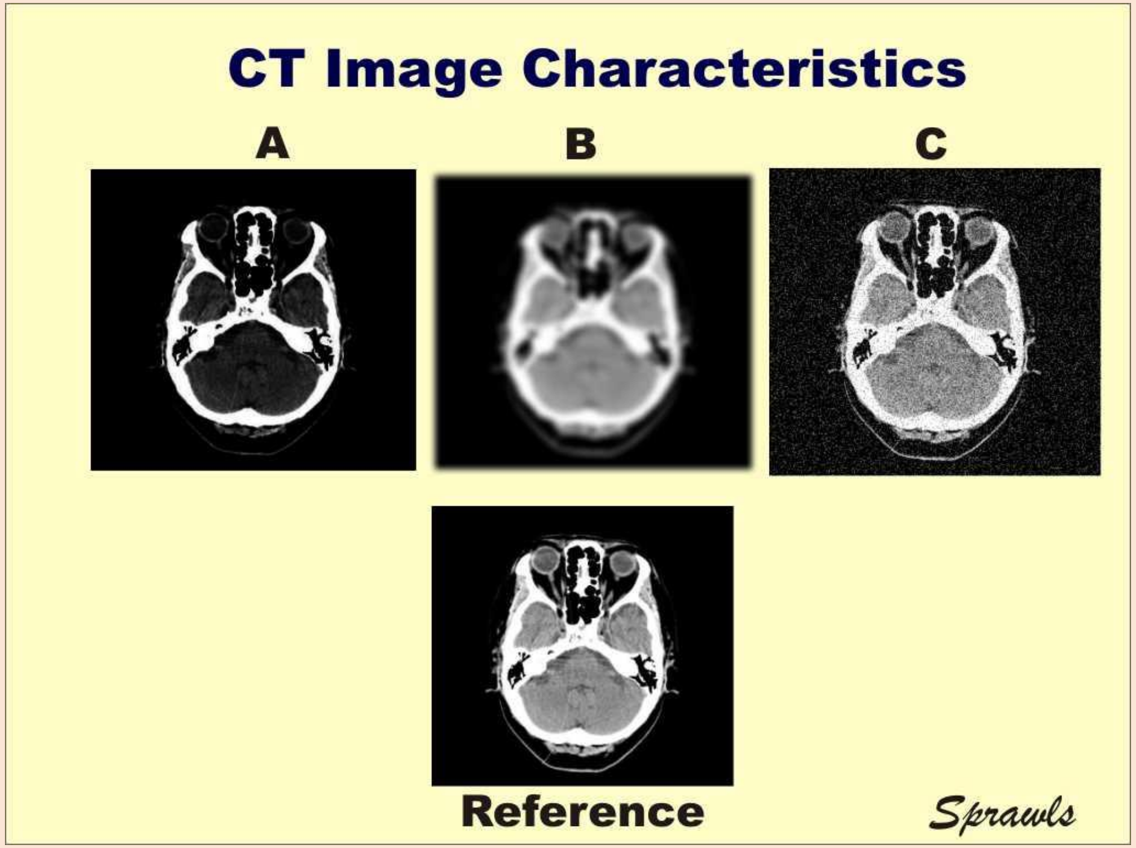
Υπάρχουν αρκετές ψευδοεικόνες, προερχόμενες από ποικιλία συνθηκών, που εμφανίζονται σε μία απεικονιστική διαδικασία. Κάποιες είναι πολύ σαφείς όπως τα streaks και τα ghosts (φαντάσματα), ενώ άλλες είναι λιγότερο εμφανείς και αναγνωρίζονται ως αλλαγές στον τρόπο απεικόνισης συγκεκριμένων ανατομικών δομών.



Ποιότητα εικόνας CT : Χαρακτηριστικά εικόνας

Τα σημαντικότερα χαρακτηριστικά που επηρεάζουν την ποιότητα της εικόνας είναι :

- Θόρυβος εικόνας
- Αντίθεση (Contrast Sensitivity)
- Ασάφεια (Blurring)

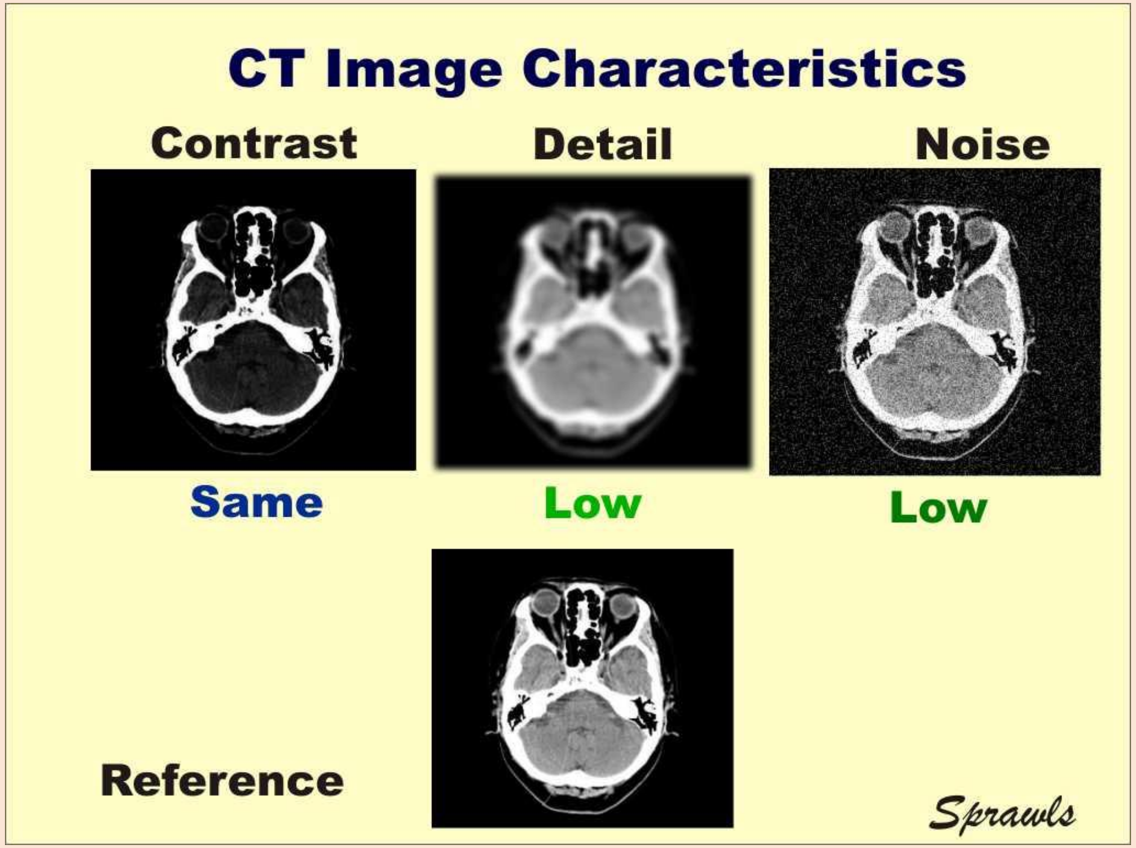


Ποιότητα εικόνας CT : Χαρακτηριστικά εικόνας

Στην πρώτη εικόνα φαίνεται διαφορά στην αντίθεση εικόνας. Τέτοιες διαφορές οφείλονται σε παραμέτρους όπως το επίπεδο και το εύρος παραθύρου (window level, window width).

Στη μεσαία εικόνα, υπάρχει ασάφεια και μειωμένη ανάδειξη λεπτομερειών.

Στην δεξιά εικόνα, ο θόρυβος είναι εμφανής (όπως τα ‘χιόνια’ σε παλαιές τηλεοράσεις).
Η ποιότητα εικόνας σχετίζεται με την απορροφούμενη δόση.

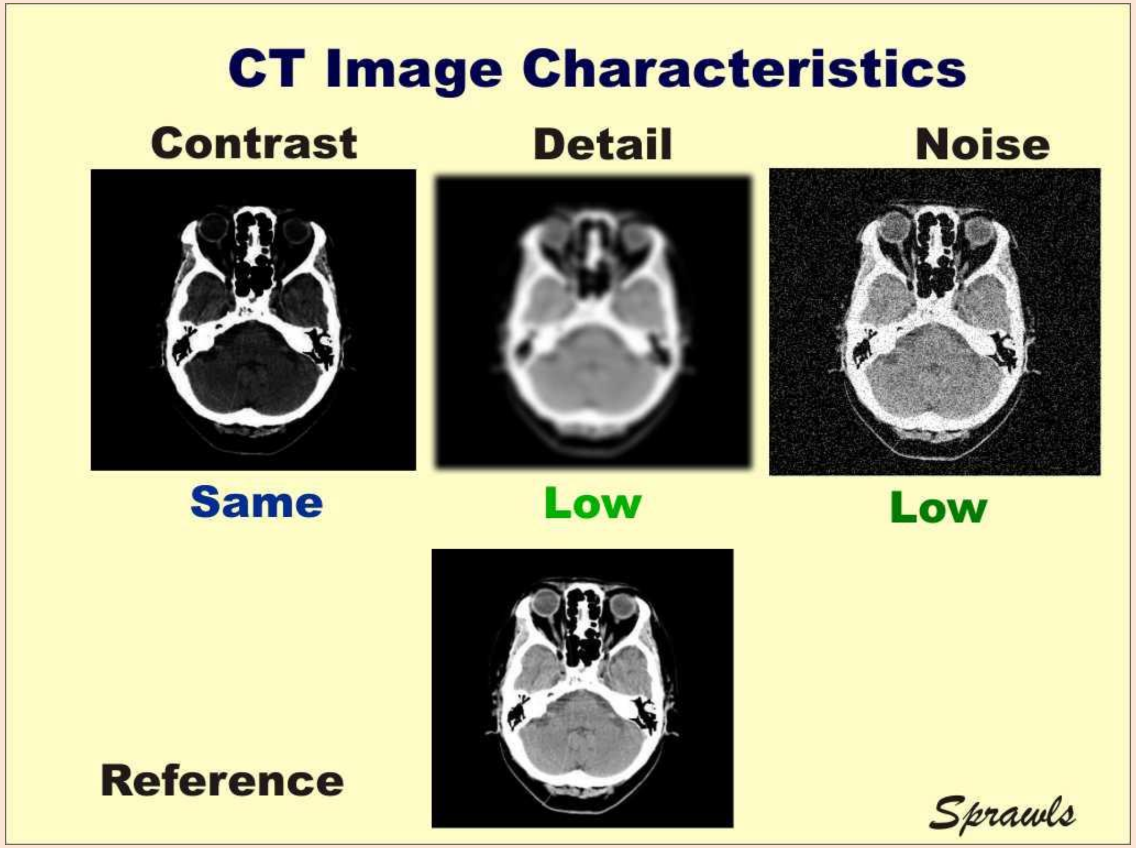


Ποιότητα εικόνας CT : Χαρακτηριστικά εικόνας

Γενικά, στην CT η αντίθεση εικόνας ή η contrast sensitivity μίας διαδικασίας δεν εμφανίζει σημαντική εξάρτηση από τη δόση. Σε αντίθεση με την CT, στην κλασσική ακτινογραφία και στη μαστογραφία, η αύξηση της contrast sensitivity σχετίζεται άμεσα με αύξηση της δόσης.

Η λεπτομέρεια εικόνας (image detail) δεν σχετίζεται άμεσα με τη δόση. Όμως, όταν πραγματοποιούνται αλλαγές στα πρωτόκολλα προκειμένου να αυξηθεί η λεπτομέρεια, αυξάνεται και ο θόρυβος. very strong indirect effect. Συνεπώς, χρειάζεται να αυξήσουμε τη δόση (αύξηση mAs) ώστε να μειωθεί ο θόρυβος. Therefore the dose must be increased to compensate for this.

Ο θόρυβος εικόνας είναι σημαντικός παράγοντας καθορισμού της δόσης στον εξεταζόμενο. Αύξηση ττης δόσης απαιτείται για τη μείωση του θορύβου. Στο διπλανό σχήμα, η μετάβαση από την εικόνα με θόρυβο στην εικόνα αναφοράς απαιτεί αύξηση της χορηγούμενης δόσης.



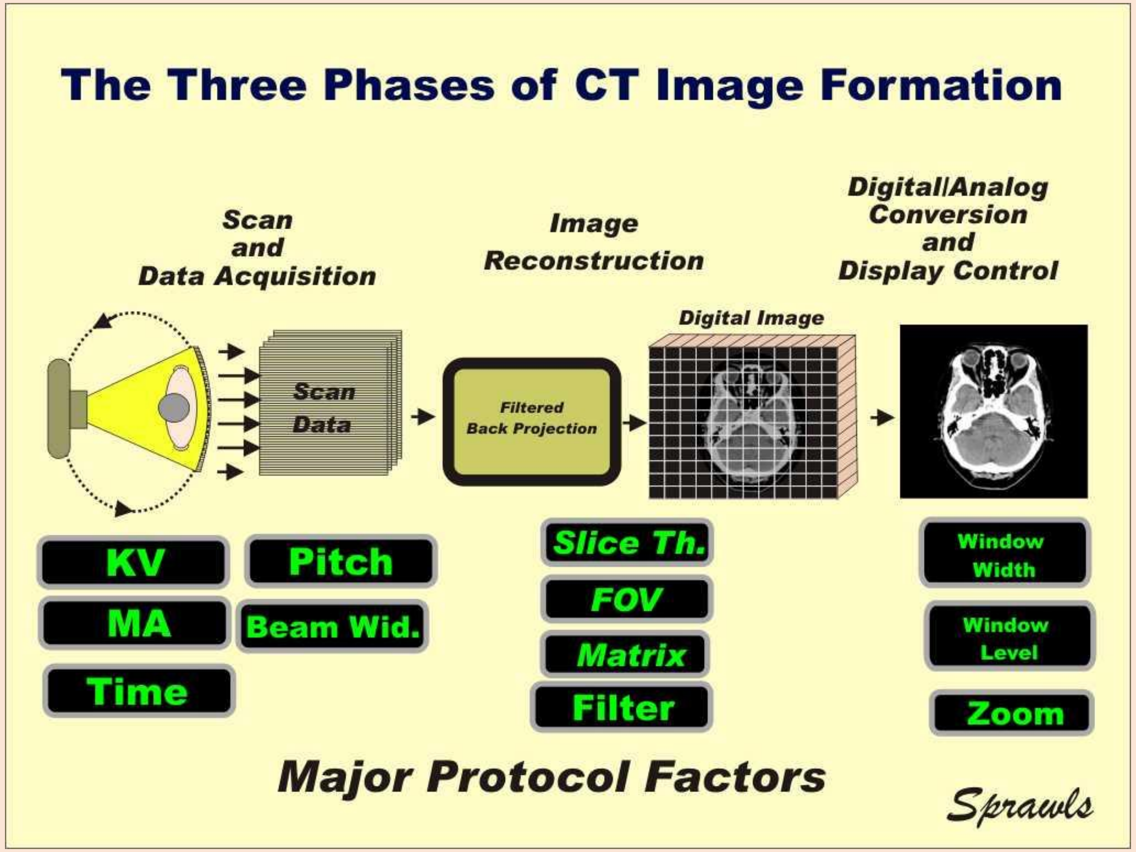
Φάσεις δημιουργίας CT εικόνας

There are adjustable protocol factors associated with each phase that have an effect on image quality and in some cases the dose to the patient. In this illustration we have identified so of the major protocol factors that control image quality.

The first phase is the scanning of the x-ray beam around and along the patient's body and the collection or acquisition of the data. This produces the scan data (but not yet an image) that is stored in the computer memory.

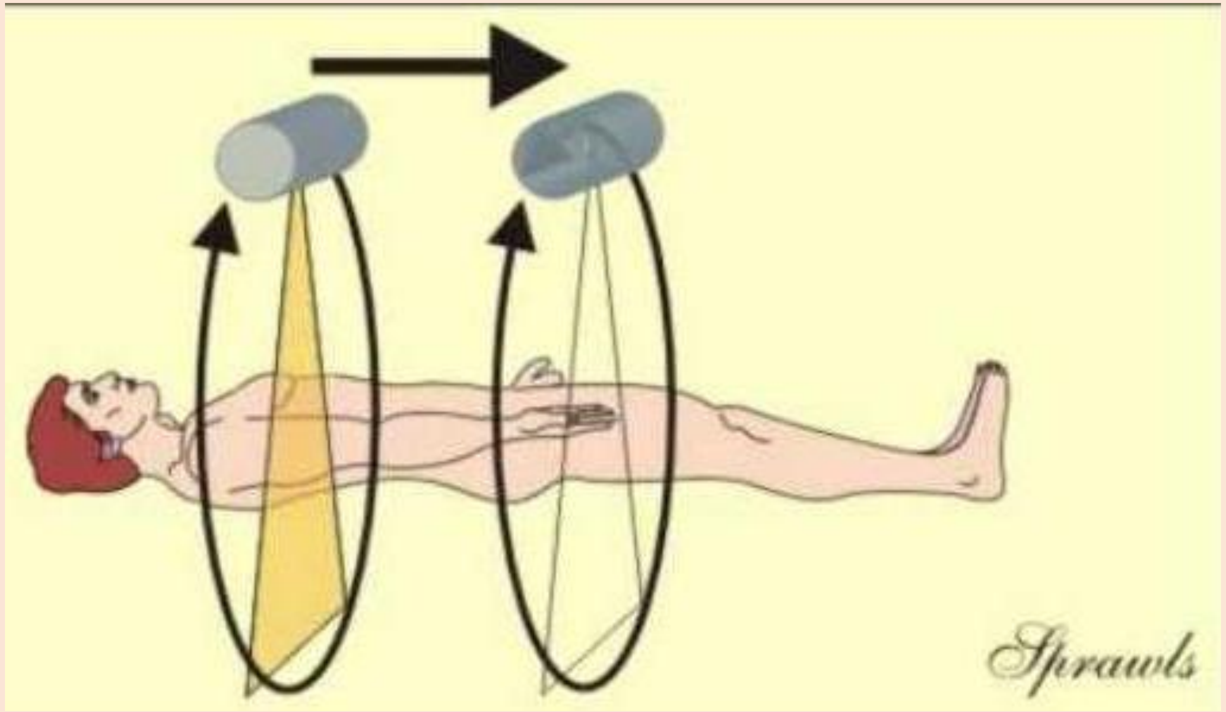
The second phase is the image "reconstruction" from the collected data. This results in a digital image of the individual slices or 3D volumes

The third phase is the conversion of the invisible digital image into a visible (analog) image for display and viewing. There are several adjustable factors that can be used to optimize the displayed image for a specific clinical task.

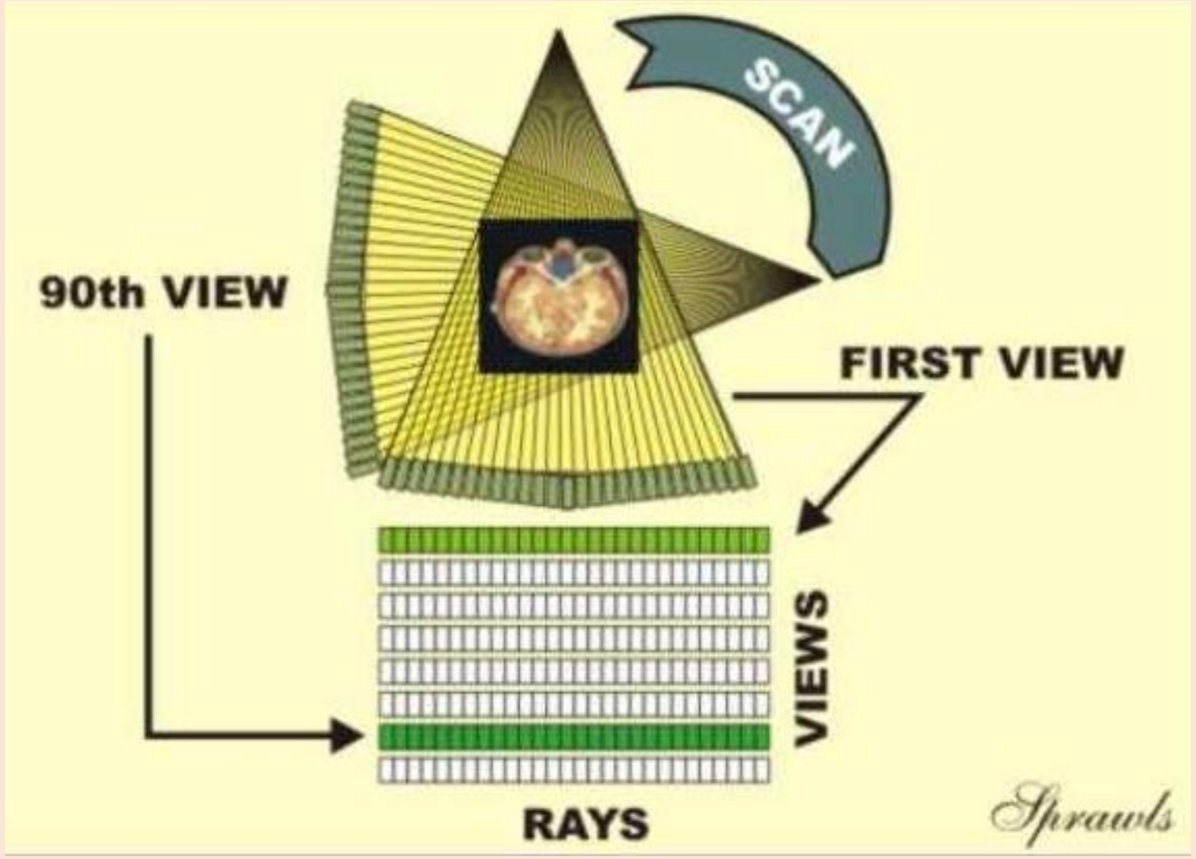


Κίνηση της λυχνίας

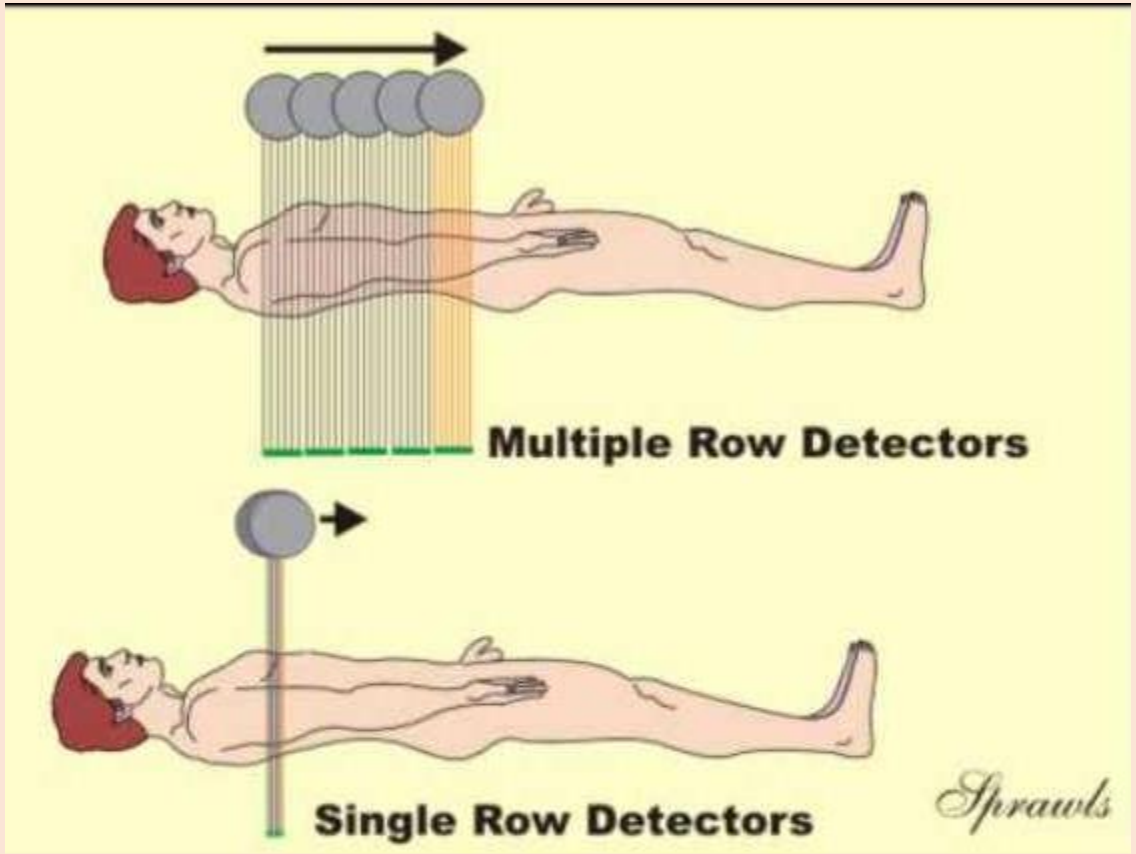
ne is the rotation of the x-ray tube and x-ray beam around the body to produce many different "views" thought the body.
The other is scanning along the length of the patient and is achieved by moving the body thought the scanner while the beam is rotating around it.
It is the combination of these two motions that produce the complete set of data from which the images can be reconstructed.



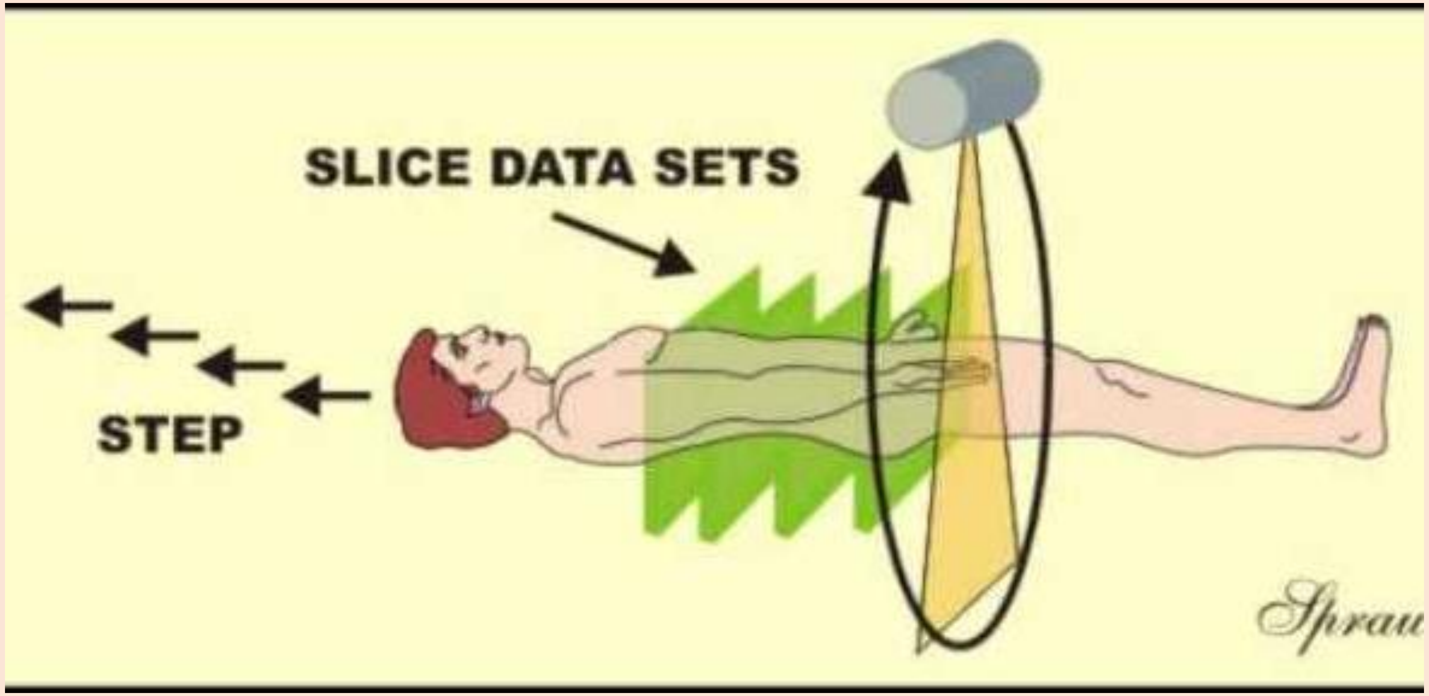
From each x-ray tube position it typically projects a thin, fan-shaped beam through the patient's body. After passing thought the body the beam in intercepted by an array of radiation detectors.
The pathway from each x-rat tube focal spot position to each detector element is designated as a ray. The detector measures the radiation that penetrates the body along each ray and records it as one data point.
As the x-ray tube is scanned around the body it produces views from each position, typically about every one degree of angle. One complete scan around the body produces several hundred views. Data from these many views are required to produce a high-quality image for each slice.



The major advantage of multiple-row detectors is they produce multiple, parallel views in each tube position which increases the speed of collecting scan data along the patient's body.
A design characteristic of a scanner is the number of detector rows (16, 32, 64, etc).
A fan-beam scanner has approximately the same number of rows as number of detector elements in each row.



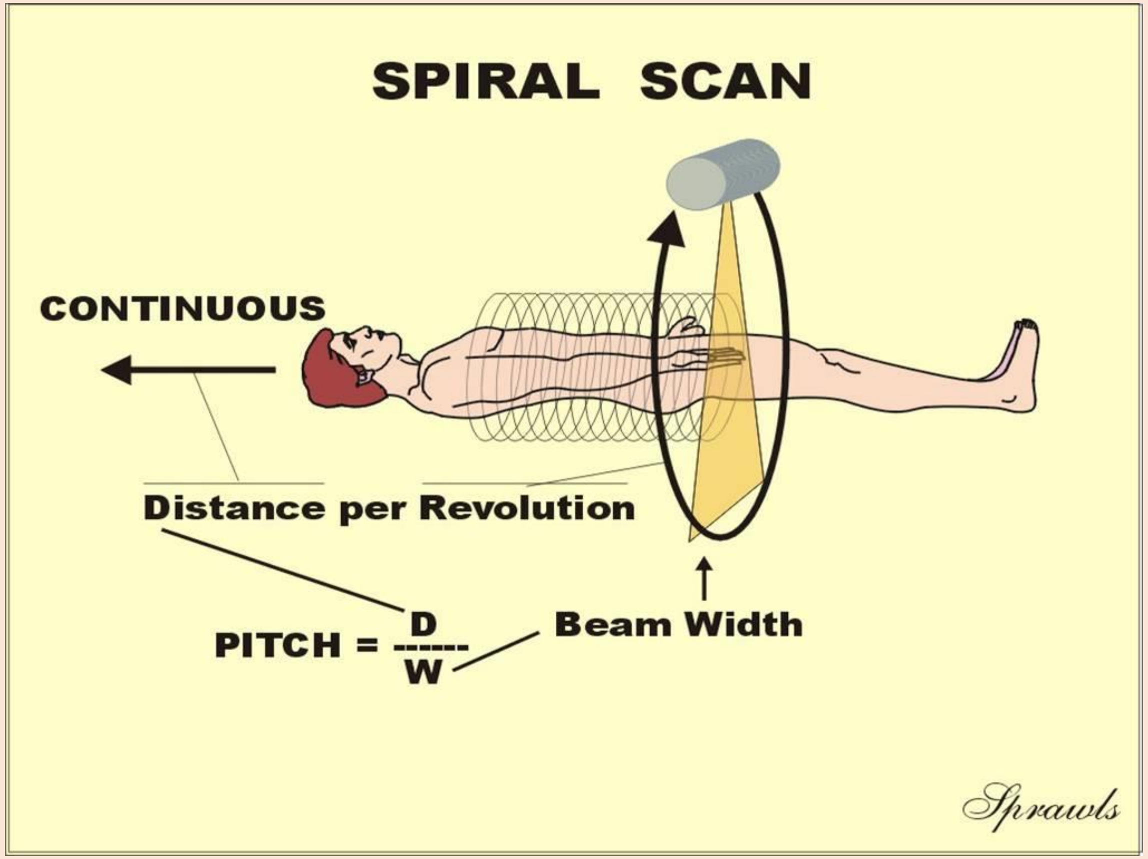
There are two distinct methods of scanning the beam along the length of the body. Each has its characteristics that must be considered.
The simplest, and the only method available for many years is the "scan and step" method illustrated here. With the body not moving the x-ray beam is rotated completely around the body and a data set for a slice is collected. With this method the data set is "locked" to a specific slice determining its thickness and position.
After each beam rotation the body is then moved or "stepped" to the next slice position where it is scanned.
The two major limitations of this method are that it is relatively slow in covering a body section and the slices (position and thickness) are fixed at the time of the scan and cannot be changed later.



Spiral or Helical Scanning

The preferred alternative for many procedures is the helical or spiral scanning method shown here.

With this method there are two continuous motions occurring at the same time. The x-ray tube and beam is rotating around the body continuously and at the same time the body is being moved through the scanner.



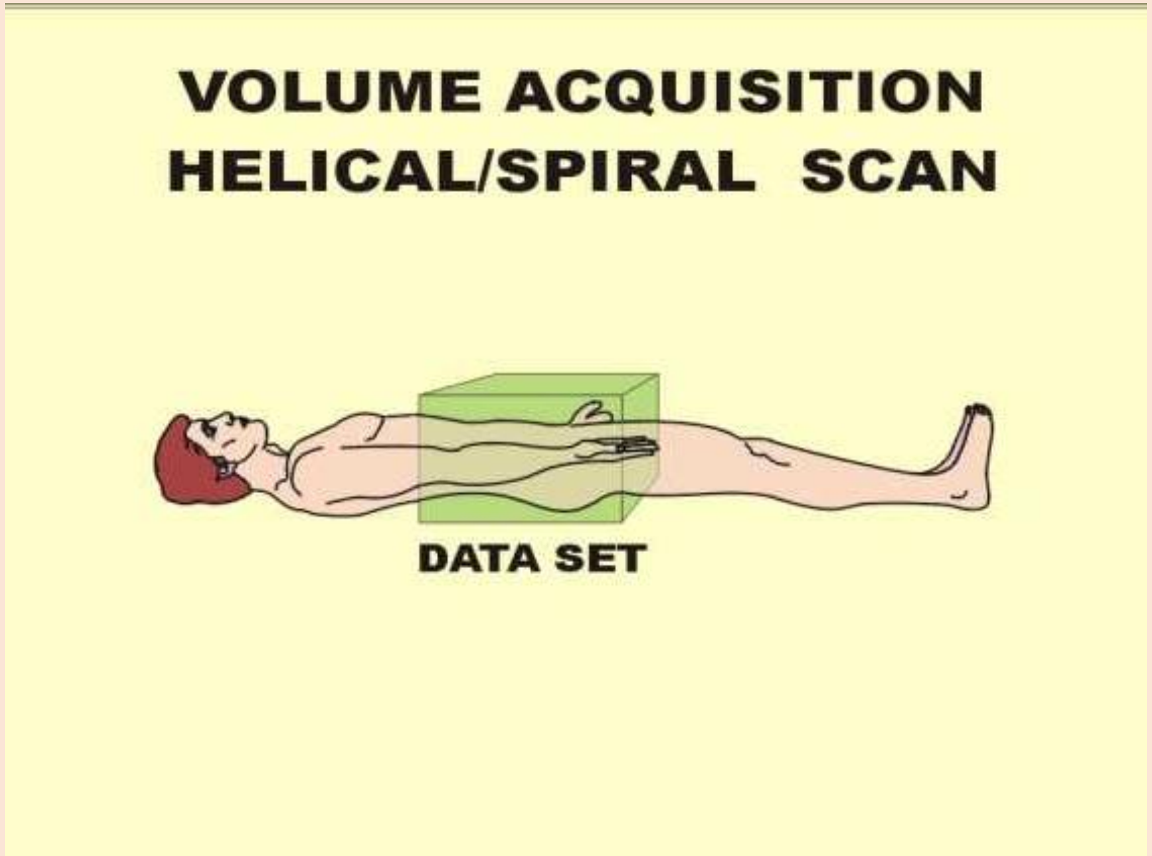
If we can imagine the path of the x-ray beam on the patient's body it would form a spiral or helical pattern. Either name is an appropriate description of this method.

There is one very important adjustable protocol factor associated with this method that can have an effect on both image quality and dose to the patient. That is the Pitch factor which is the distance the body is moved, as a multiple of the beam width, during one rotation of the x-ray beam. For example, if the pitch is set to a value of 2, the body would be moved twice the thickness of the beam during on rotation

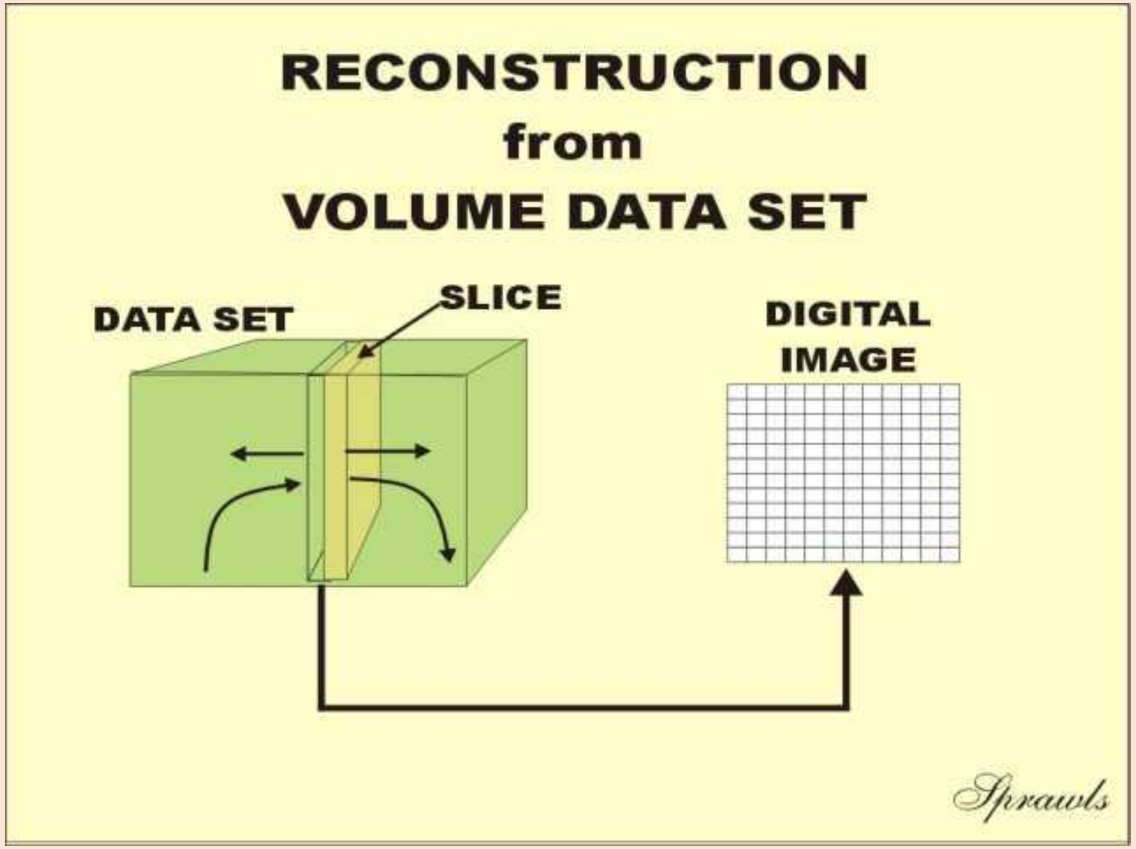
Increasing the pitch value increases the relative speed of moving the body and reduces the time required to cover an anatomical area. We will come back to this in more detail later but in general increasing pitch reduces the dose to the patient but can limit image quality.

A very significant value of spiral/helical scanning is the characteristics of the data set that is produced as illustrated here.

The data set is continuous over the anatomical area scanned and not divided into individual slices as with the scan and step method described above. This offers much flexibility when the images are being reconstructed. With the continuous data set it is possible to reconstruct images of slices anywhere within the data set. These images can be reconstructed for different slice thicknesses and orientations.

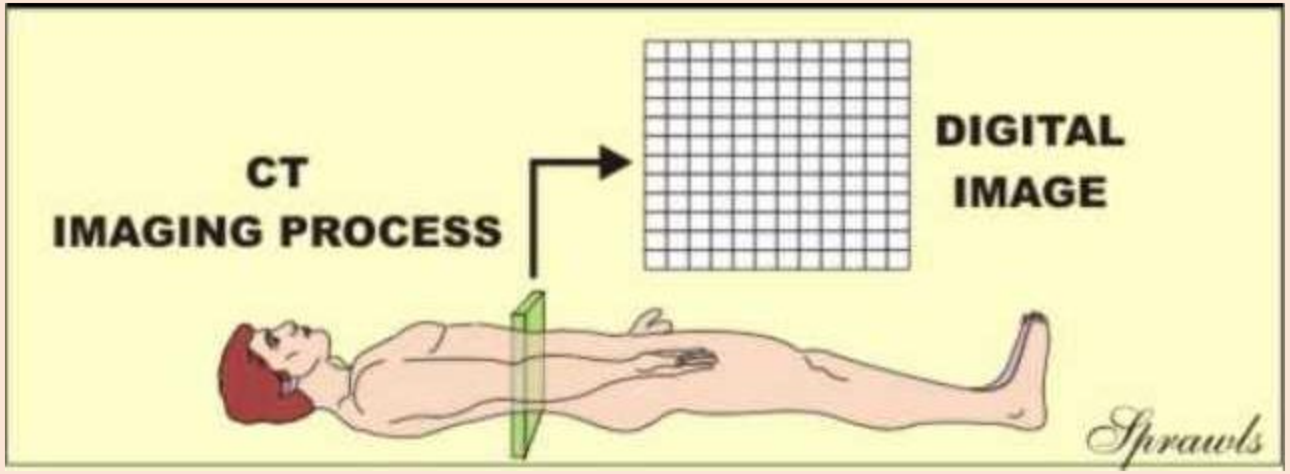


With spiral scanning the slices are determined at the time of reconstruction, not at the time of the scanning and data It is possible to go back and reconstruct images for different slice characteristics without scanning the patient again. Another great value of helical scanning is that the continuous data set can be used to reconstruct 3D or volume (compared to slice) images.



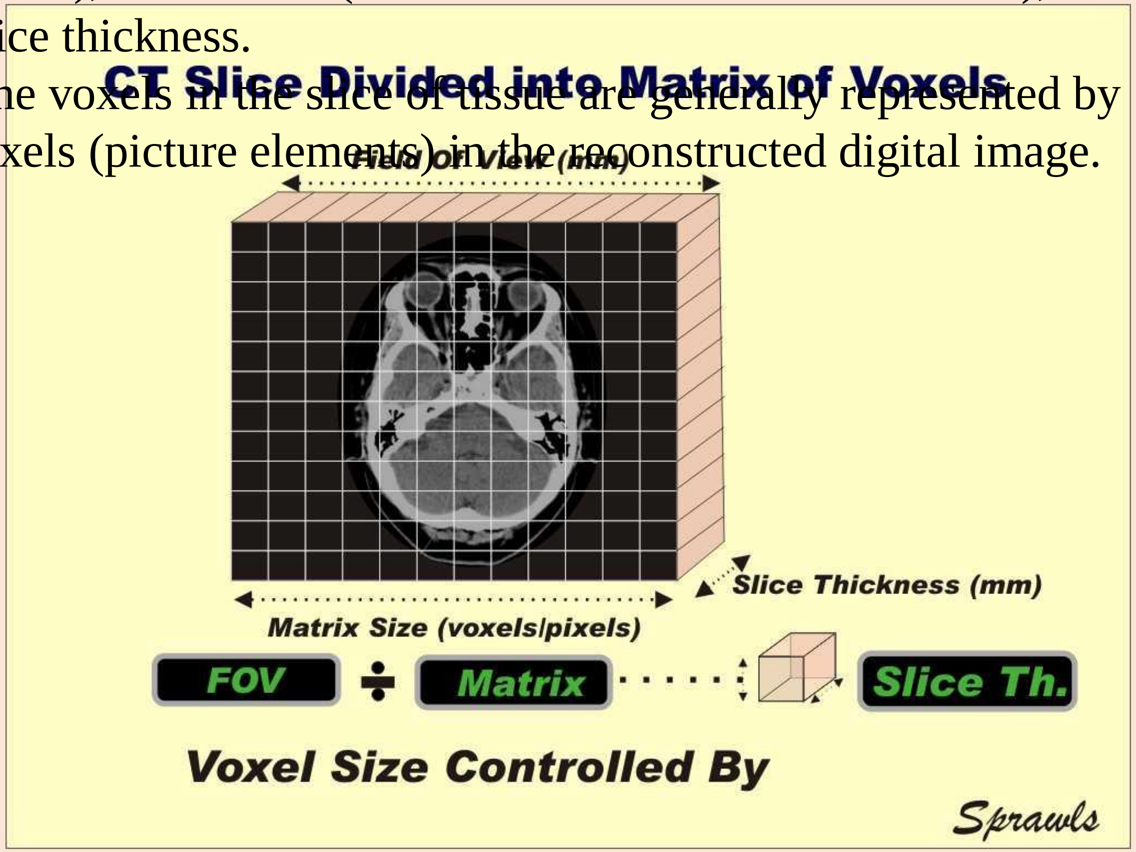
Ανακατασκευή εικόνας - Voxels

CT image reconstruction is a mathematical process for converting the scan data into a digital image of a specific anatomical area. Most images are created with the filtered back-projection method or sometimes with an enhanced process generally known (generically) as iterative reconstruction. It is not necessary for us to go into the mathematical details but focus on sertain characteristics of the reconstruction process that are adjustable and have an effect on image quality and radiation dose.



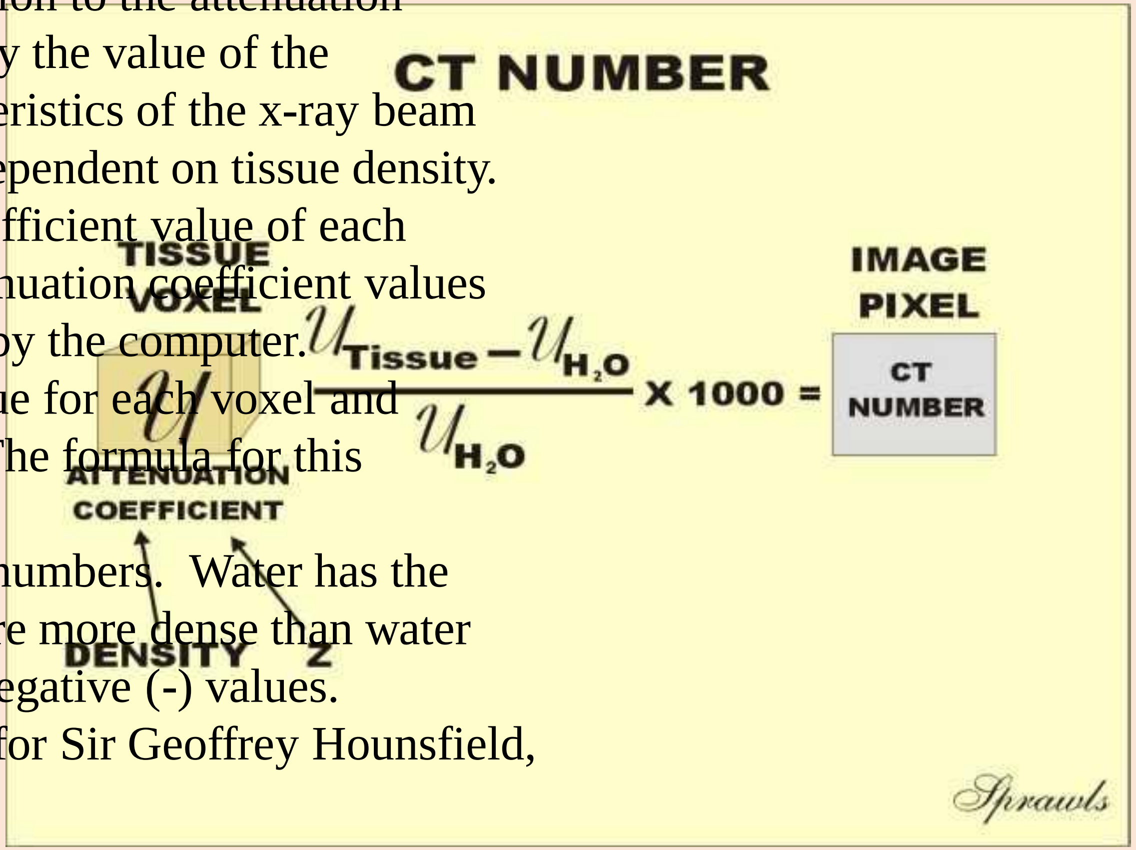
One of the most critical factors in CT image quality and patient dose is the size of the individual voxels. The size is controlled by a combination of three protocol factors: the field of view (FOV), matrix size (number of voxels in each direction), and the slice thickness.

The voxels in the slice of tissue are generally represented by pixels (picture elements) in the reconstructed digital image.



Ανακατασκευή εικόνας – CT Numbers

Let's recall that during the scanning phase the individual rays of the x-ray beam are projected through the patient's body. They are attenuated (absorbed) in proportion to the attenuation properties of the tissue along the path. This property is represented by the value of the attenuation coefficient for the specific tissue. Because of the characteristics of the x-ray beam used in CT (high KV and heavy filtration) the attenuation is highly dependent on tissue density.
During the back projection reconstruction process the attenuation coefficient value of each individual voxel is calculated. However, we never see the actual attenuation coefficient values because there is another step in the mathematical process performed by the computer.
A CT number value is calculated from the attenuation coefficient value for each voxel and becomes the value for the corresponding pixel in the digital image. The formula for this calculation is shown in the illustration.
Water (H_2O) is used as the reference and calibration material for CT numbers. Water has the assigned CT number value of zero. Tissues or other substances that are more dense than water will have positive (+) values and those that are less dense will have negative (-) values.
CT numbers calculated in this matter are in Hounsfield units, named for Sir Geoffrey Hounsfield, the engineer who invented and developed computed tomography.



Παράγοντες που επηρεάζουν την λεπτομέρεια εικόνας

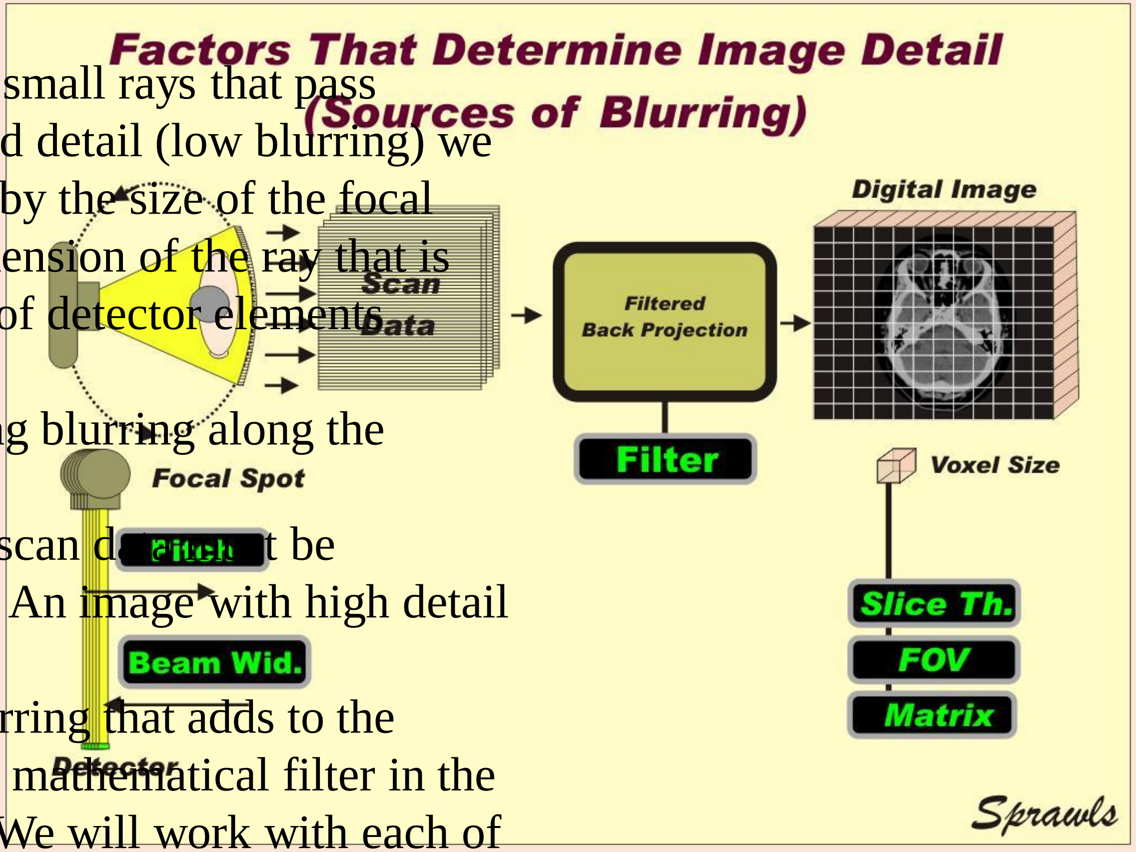
The ultimate detail available in an image is determined by the total effect of the blurring that occurs throughout the image formation process. The most significant blurring occurs in the first two phases, scanning and reconstruction.

We call that in the scanning phase the x-ray beam is divided into many small rays that pass through the body and measure the attenuation along their path. For good detail (low blurring) we must scan with small rays. The size of an individual ray is determined by the size of the focal spot on one end and the size of the detectors on the other end. The dimension of the ray that is usually variable is the beam width which is determined by the number of detector elements selected in that dimension.

Increasing the pitch has the effect of "spreading" the ray and introducing blurring along the length of the body.

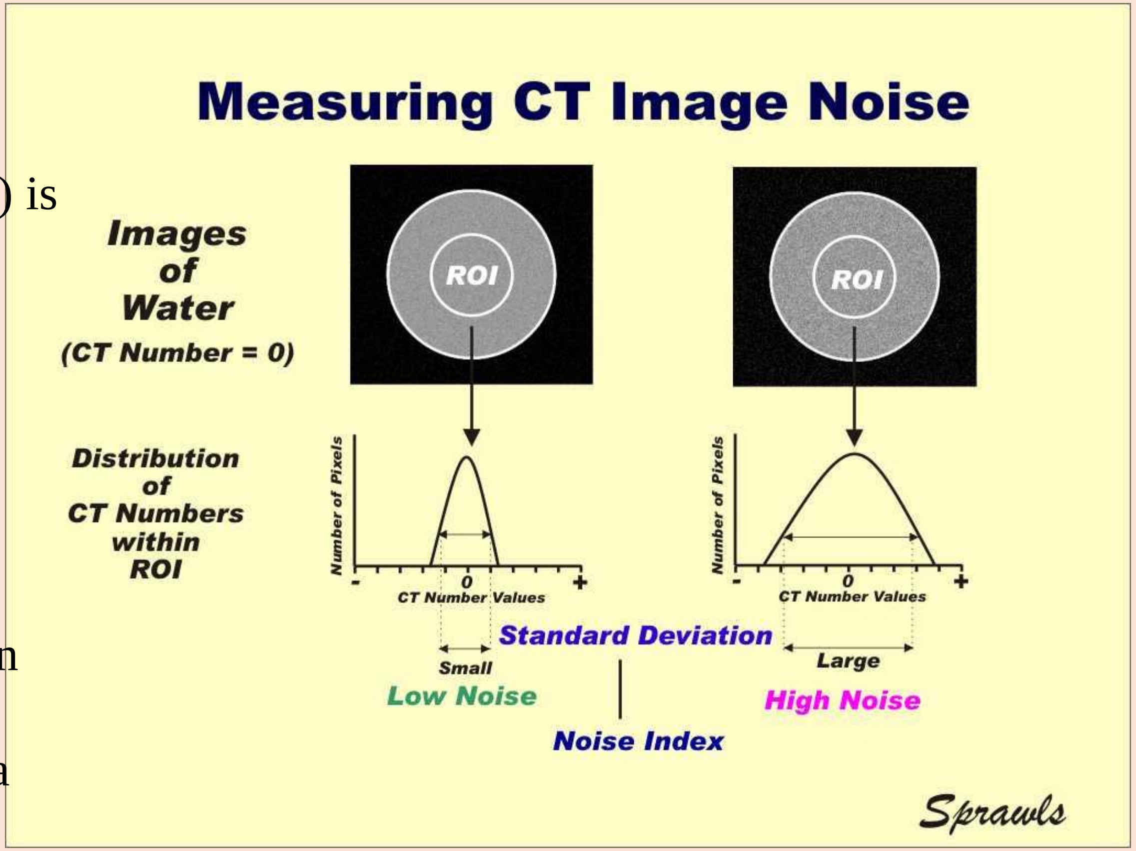
Important point....if an image with high detail is required "high detail" scan data must be collected, generally by scanning with a thin beam and low pitch value. An image with high detail cannot be reconstructed later if the "detail" is not in the scan data.

During the reconstruction phase there are two additional sources of blurring that adds to the blurring in the scan data. One is blurring that might be produced by the mathematical filter in the reconstruction calculation and the other is the size of the tissue voxel. We will work with each of these after we bring noise back "into the picture".

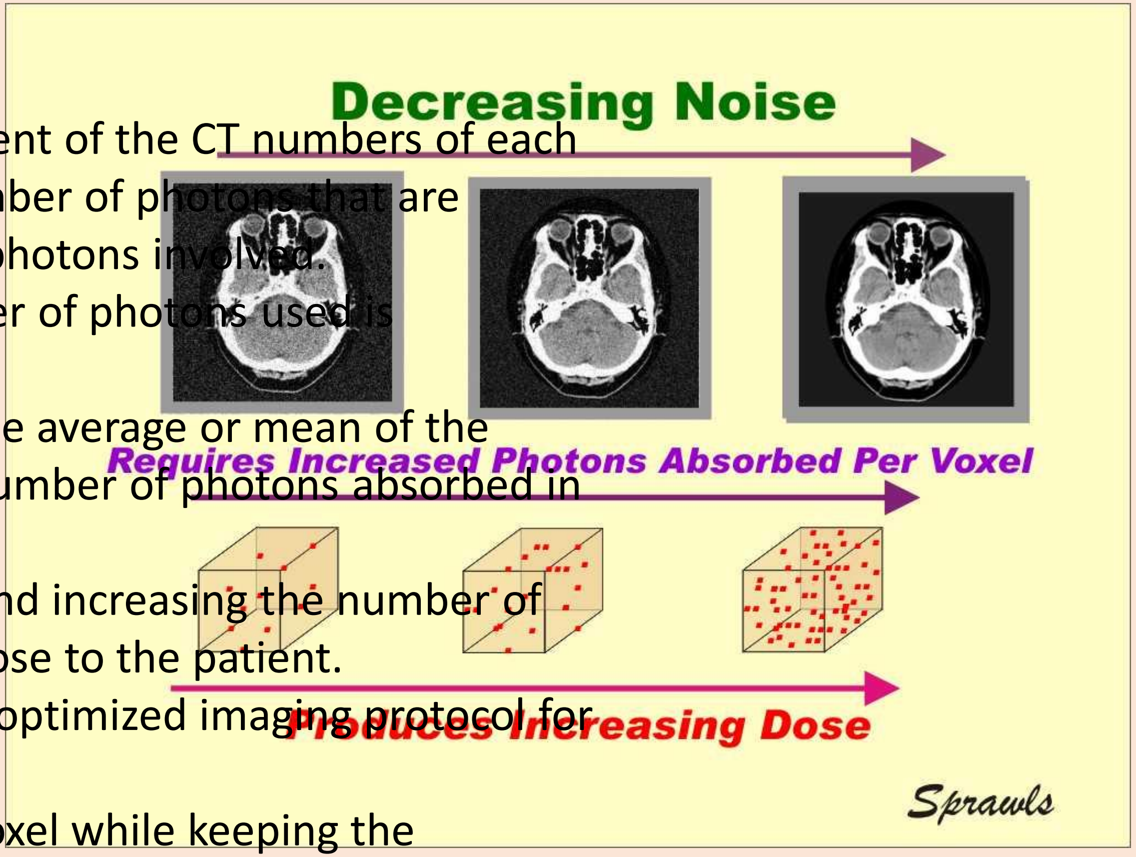


Υπολογισμός θορύβου εικόνας

The measurement is made using the CT system from the viewing console. Using a standard viewing function a region of interest (ROI) is selected within the image.
The next step is to use another standard function of the CT system which measures the statistical standard deviation of the CT numbers (pixel values) within the ROI of the image.
Water has a theoretical CT number value of zero (0) Hounsfield units. That is the value all pixels would have in a perfect image WITHOUT any noise.
The effect of the noise in a CT image is to cause a statistical variation in the actual CT numbers from pixel to pixel. The range of this variation, as measured by the calculated standard deviation (SD), is a measure of the noise. This is often referred to as the noise index.
We see that the image on the right has more visual noise, a greater variation in CT number values, and a larger calculated SD value.



Ρύθμιση θορύβου εικόνας

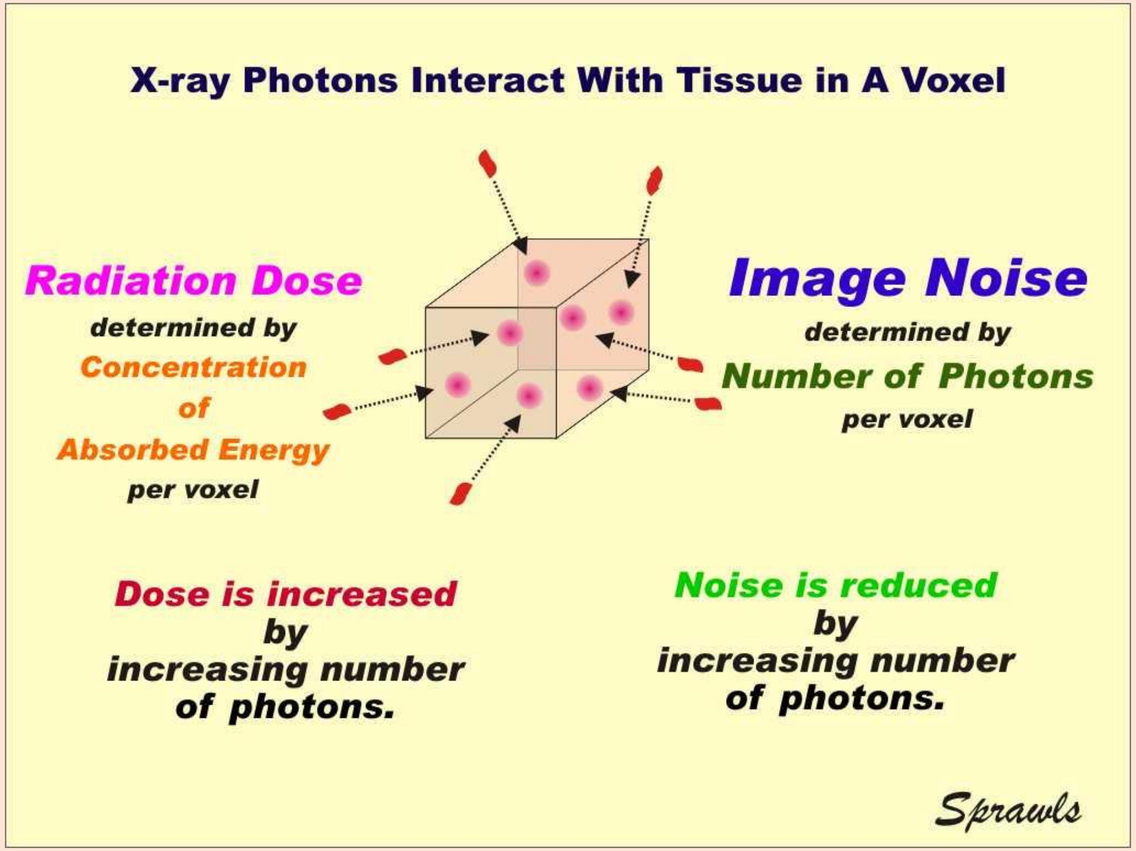


The noise in an image comes from statistical errors in the measurement of the CT numbers of each tissue voxel. The precision of measurements made based on the number of photons that are attenuated or absorbed depends on the average (mean) number of photons involved.
The precision increases (the error and noise decreases) as the number of photons used is increased.
Let us recall that the standard deviation (SD) is inversely related to the average or mean of the number of the photons in each measurement. In this case it is the number of photons absorbed in each voxel.
While we can easily reduce the noise by "turning up" the exposure and increasing the number of photons that has the undesirable effect of increasing the radiation dose to the patient.
That is one of the issues we must address in our effort to provide an optimized imaging protocol for each patient.
The goal is to get a sufficient number of photons absorbed in each voxel while keeping the absorbed energy (dose) as low as possible.

Ρύθμιση θορύβου εικόνας

The important point here is the difference between the number of photons absorbed in each voxel and the concentration of photons absorbed which determines the dose. So the question is how can we increase the number of photons (reduce the noise) without increasing the concentration of photons and increasing the dose.

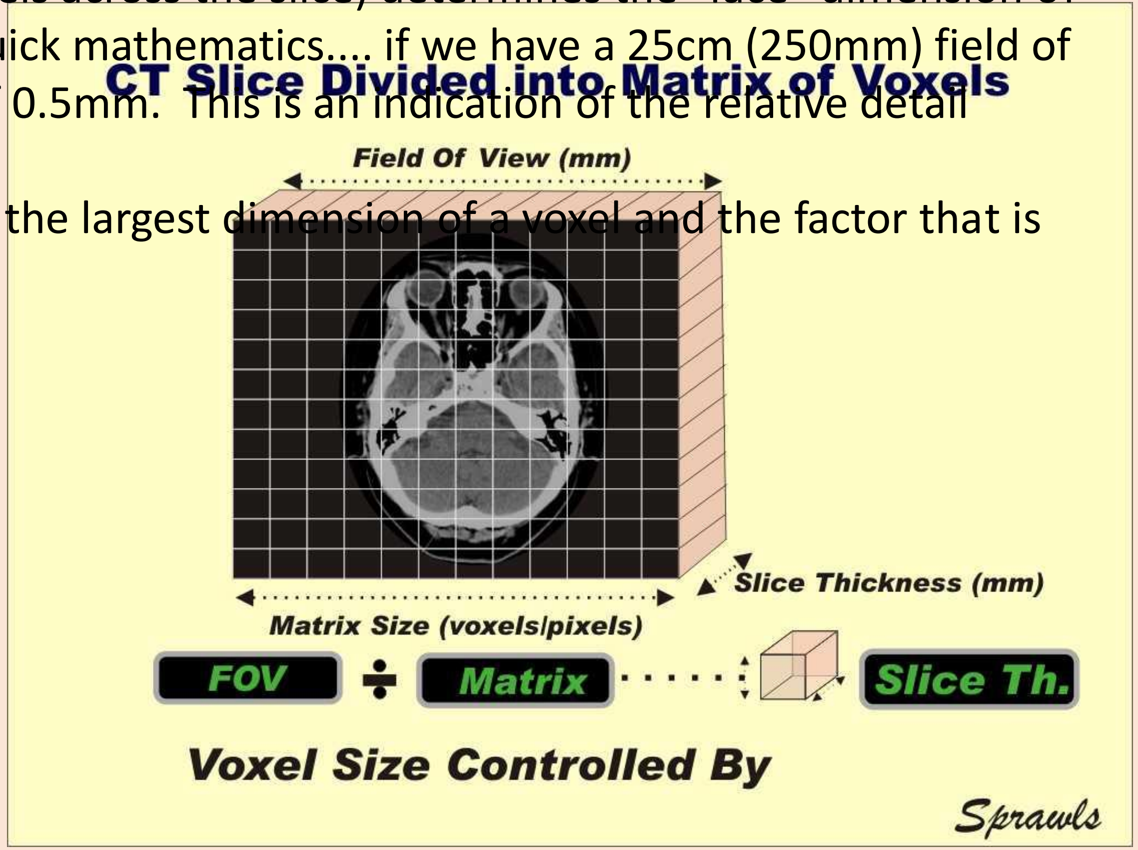
This can be achieved by increasing voxel size. A larger voxel will capture more photons without affecting the dose.

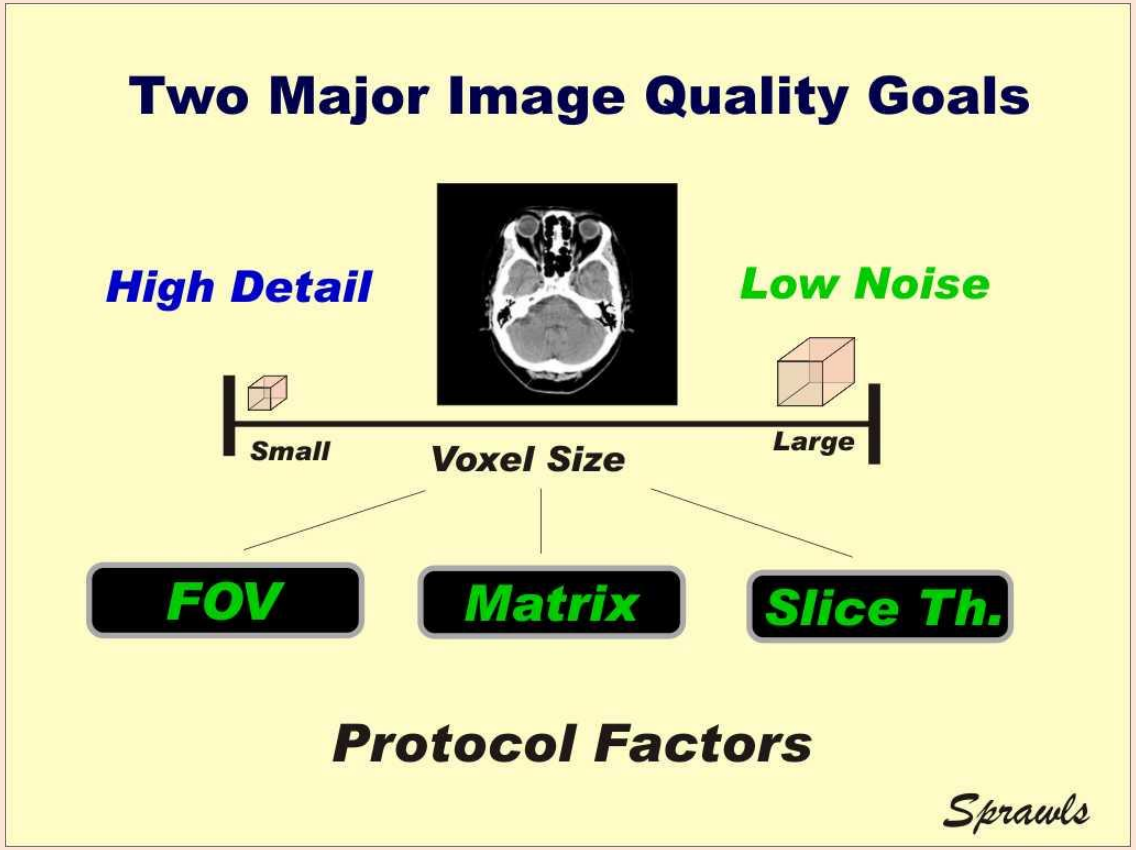


Το μέγεθος του voxel

The ratio of the field of view (FOV) to the matrix size (number of voxels across the slice) determines the "face" dimension of a voxel. This corresponds to relative pixel size in the image. Some quick mathematics.... if we have a 25cm (250mm) field of view and a 512 matrix size that gives voxels with a face dimension of 0.5mm. This is an indication of the relative detail available in CT images.

The other dimension of the voxel is the slice thickness. It is typically the largest dimension of a voxel and the factor that is often varied the most for different clinical procedures.





We need to reconstruct images with small voxels to minimize blurring and get good image detail but increasing voxel size reduces image noise.

That is because larger voxels "capture" more photons for a specific dose and that improves the statistics and reduces the noise.

Παράγοντες που επηρεάζουν το θόρυβο εικόνας

During the scanning phase the same factors we introduced earlier as those that affect dose also have an effect on noise because they control the concentration of photons absorbed in tissue.
Changing any of these factors changes both noise and dose.
Now to the reconstruction phase. As we have just observed, changing the voxel size changes the ratio of number of photons (noise) to the concentration of photons (dose).
The other controlling factor is the mathematical filter that is included in the reconstruction process.