CT Basics: Data Acquisition
Module 3

1. Title Screen
   Welcome to Module 3 of CT Basics – Data Acquisition. This module was written by Chandra Gerrard, A.S., R.T.(R)(CT).

2. License Agreement

3. Objectives:
   After completing this module, you will be able to:
   - Name the methods of acquiring computed tomography images.
   - Explain the functions of the data acquisition system (DAS).
   - Describe the process of data acquisition.
   - Identify the factors that influence data acquisition.
   - List the selectable scan factors used to acquire images.

4. Axial
   There are three major methods of data acquisition in computed tomography: axial, spiral and cine. The choice of scan mode depends on the type of examination required. Axial acquisition refers to the rotation pattern of the x-ray tube around the patient. This method creates an image that is free of superimposition of the anatomy above and below the area of interest.

   Axial imaging is very different than imaging performed in diagnostic radiology. In diagnostic radiology, the patient is positioned and an image is obtained. The diagnostic image offers only one view of the area of interest. To see the anatomy at a different angle, another film must be taken.

5. Axial Scanning
   Axial can refer to the method of scanning, or it can identify the appearance of the projected image. Images acquired solely in an axial format usually are performed as a start-stop acquisition. For example, during a CT scan of the sinuses, the patient lies on the table in the prone position with the neck in hyperextension. The x-ray tube around the patient and acquires an image. The table then moves a little further into the gantry, and the x-ray tube rotates again around the patient, acquiring another image. This process takes place until all the scan is completed. Click on the button to begin the animation.

6. Axial Scanning
   Axial scanning is used when the x-ray tube within the CT scanner can't sustain the heat load necessary for rapid spiral acquisition. Axial acquisition also reduces several artifacts such as partial volume averaging and noise. Images with less noise provide better visualization of very similar shades of gray and have better low-contrast resolution. Noise is decreased because axial imaging does not require the interpolation equations, or the estimation of data, necessary for the display of spiral images.

   However, greatly improved low-contrast resolution is the most significant image quality factor related to axial scanning, which is especially important when imaging structures such as the brain. The thicker scan slices typically associated with axial imaging decrease the amount of noise artifact and increase the viewer’s ability to differentiate between two tissues of similar densities.
7. **Helical/Spiral Technology**

Early single-slice CT scanners contained fewer detectors and needed time for the x-ray tube and detector array to unwind after a single rotation. The scanning ability of the x-ray tube was restricted because of tension in the cables connected to the generators. This holdup was referred to as an interscan delay. Slip-ring technology made helical/spiral scanning possible. The slip rings allow the x-ray tube and the detectors opposite the tube to move continuously in a circular path within the gantry, free of hang-ups.

8. **Helical/Spiral Scanning**

The terms “helical” and “spiral” can be used interchangeably when discussing a CT scanner that acquires images using a corkscrew-like motion. During a helical/spiral acquisition, the x-ray tube and detectors rotate around the patient continuously while the patient table slides through the scanner. Helical scanning usually is performed during a single breath hold. Click on the button to begin the animation.

The multiple rows of detectors in a helical scanner collect a great deal of image data during the scan. The configuration of the detector rows creates multiple images within a single slice of scan data. For example, each rotation of a 4-slice scanner yields 4 images. A 16-slice scanner produces 16 images for a single rotation and so on. As the number of detector rows increase, the length of the breath hold for the patient as well as the scan time decreases.

9. **Helical/Spiral Scanning**

A helical scan mode is used when larger areas of anatomy need to be covered in a short period of time. Helical/spiral scanning acquires images at a faster rate than the time required for slice-by-slice scanning. Helical scanning also creates raw data that can be reconstructed in any desired plane after the exam is completed. The detail of the exam can be increased or decreased based on how compressed or tight the corkscrew pattern is around the patient. In other words, if the number of times the x-ray tube and detectors rotate around the patient increases, the percentage of overlap of the slices also increases. Overlapping improves the detail in an image, but also results in increased radiation dose to the patient.

If the scanning pattern is compressed and the slices overlap, the x-ray beam penetrates the same area of anatomy more than once. The percentage of overlap equals the amount of anatomy re-irradiated for each slice. For example, let’s say the slices overlap during the scanning process. The same anatomy is irradiated again by the percentage of the overlap. The overlap increases spatial resolution but also increases patient dose.

On the other hand, the patient couch can move faster during the scan to allow the anatomy to be imaged quickly. This degrades the spatial resolution within the image, but at the same time reduces the radiation dose to the patient. The dose decreases because the slices don’t overlap and there’s unscanned anatomy. However, this situation increases an artifact called partial volume averaging. Partial volume averaging occurs when the computer calculates what the Hounsfield number should be based on the pixel values in the adjacent tissues. This estimate leads to misdiagnosis because the actual tissues may have different Hounsfield numbers than the assumed values the computer generates.

10. **Cine**

Cine is a way of viewing CT images in a movie-like format and is typically performed retrospectively. The images acquired for cine mode are usually obtained by overlapping thin image sets. These thin data sets increase spatial resolution, but the patient receives more radiation. Cine mode uses computer software to compile the entire axial image data set, allowing the viewer to scroll through the
images in different formats. Formats include coronal, sagittal or even 3-dimensional presentations. Use the slider bar to scroll through the images.

11. Practice Question

12. Practice Question

13. Data Acquisition Process

At its core, data acquisition in a computed tomography scanner consists of collecting image data, processing the data based on how the x-ray beam was attenuated and displaying the image data in a way that makes sense to the observer. This chain of events begins when x-rays from the radiographic tube are attenuated by the patient and absorbed by detectors. The difference between the power of incoming and outgoing x-rays is what makes CT possible. The initial raw data are nonsensical until the computer algorithms give it meaning.

14. Lambert-Beer Law

When Sir Godfrey Hounsfield first experimented with CT technology, he used a gamma radiation source that produced x-rays with equal energies. This type of x-ray beam is referred to as a homogenous beam. A homogenous beam means that the raw data can be processed using the Lambert-Beer law. According to the Lambert-Beer law, for every equal amount of tissue irradiated, an equal amount of x-ray photons will be absorbed. The Lambert-Beer law is based on the energy of the x-ray photons when they strike the detectors, the amount of x-ray photons to initially leave the x-ray source, the thickness of the anatomy of interest and the expected x-ray absorption.

For the x-ray beam to fully pass through the tissue of interest, the scan time must be adjusted, because every 1 inch of tissue irradiated absorbs a known, unchanged number of x-ray photons. If a constant number of x-ray photons are absorbed for every 1 inch of anatomy, enough x-ray photons must strike the detectors when the beam finally does pass completely through the patient.

Modern CT scanners do not use the Lambert-Beer law because the source is now an x-ray tube that creates a heterogeneous beam of different x-ray strengths. The mathematical equations used by current CT scanners focus on the linear attenuation coefficient. The linear attenuation coefficient calculates the total number of x-ray photons after the photons have hit the detectors and the scan is completed.

15. CT/Hounsfield Numbers

The algorithms used to determine attenuation and tissue type are far too complex for this module; however, the results of the data analysis using these algorithms are extremely important. Specific numbers are assigned to individual pixels within the image data set based on the type of material imaged. These numbers are referred to as CT, or Hounsfield, numbers.

CT/Hounsfield numbers give the CT scan a desired appearance. The range of CT/Hounsfield numbers is referred to as the window width, or the total amount of CT/Hounsfield numbers applied to any given examination. Window width is typically characterized as either narrow or wide.

A narrow window width is associated with fewer changes in the shades of gray within the scan. This is especially important when imaging anatomy with similar densities like the brain. Gray and white matter found in the brain are different from one another by only a few CT/Hounsfield numbers. A wide window width is used when an examination needs many shades of gray to adequately show the changes in different tissues.

16. Data Acquisition System
When an x-ray beam travels through a patient, the beam is absorbed, deflected or simply passes through with no interaction. Some x-rays eventually hit the detector array opposite the x-ray tube. When the x-rays hit the detectors, light is produced and converted into electrical energy. Regardless of the method used to acquire data, whether spiral, axial or cine, the data acquisition system, or DAS, must convert the raw electrical energy into functional data that can be used to create an image. The data acquisition system is made up of several components including an analog-to-digital converter, a digital-to-analog converter and 2 types of amplifiers.

17. Analog-to-Digital Converter
Analog refers to a signal that is a continuous stream of information. Digital refers to a signal that uses selected portions of information rather than the entire stream. The analog-to-digital converter, or ADC, takes the continuous stream of analog data and selects bits and pieces to create a new digital stream of data. The digital stream contains more than enough information; however, there is a smaller amount of data for the computer to sort through. An ADC runs the analog signal through several different elements before the signal is converted to a digital one. These units include a sampler, a quantizer and a coder.

A sampler is a unit that takes bits and pieces of a continuous signal. The pieces of the signal are then reconstructed to get the desired information. A quantizer is the piece of equipment that collects the small pieces of information and gives usable data to the coder. The coder takes the information from the quantizer and assigns the binary units that become the digital signal.

18. Digital-to-Analog Converter (DAC)
The digital-to-analog converter, or DAC, changes the data provided by the analog-to-digital converter back to a usable analog signal. The data are converted because many monitors use an analog signal. The quality of resolution provided by the DAC is directly related to the number of bits, or information, provided by the ADC. When a system uses both an analog-to-digital converter and a digital-to-analog converter, the two are joined together in what is known as a digital signal processor.

The DAC ensures the data are free from sampling errors. The DAC also takes the data to a smaller continuous stream of information. Ensuring the data are free from artifacts and contain enough usable data is known as preprocessing. The preprocessed data available to the computer are called reformatted raw data.

19. Amplifier/Log Amplifier
After the x-ray beam passes through the patient and hits the detector elements, the x-ray photons are converted to electrical energy. This electrical energy is so weak that it needs to be increased significantly. An amplifier is an device that increases the electrical signal produced by the x-ray interactions. After the electrical signal has been increased, it is sent to the log amplifier.

The role of the log amplifier is similar to that of the initial amplifier in that the log amplifier takes the increased electrical signal and performs a series of mathematical equations to calculate the amount of attenuation that occurred during data acquisition.

The calculation, or logarithmic conversion, is based on how much tissue was irradiated and how many x-ray photons actually hit the detector elements. The calculated values represent the thickness and the intensity at which the images were acquired. The logarithmic conversion changes when the thickness of tissue changes.

20. Transmitted Beam Measurement
The data acquisition system is responsible for measuring the transmitted beam. The attenuated x-ray beam is evaluated by 2 sets of detectors. The detectors measure both the primary and secondary
radiation beam. The primary x-ray beam, or incoming x-ray beam, is the stream of photons that leaves the x-ray tube but has yet to reach the patient. The secondary x-ray beam is the beam that has been attenuated or changed after interacting with the tissues of the patient.

21. Binary Data Encoding

The data acquisition system assigns a brightness or intensity number to the incoming electrical energy. These numbers then are converted to binary numbers — that is, a string of zeros and ones called the binary code. The computer reads and corrects any errors in the code, which then is available for use. The prefix “bi-” in the word “binary” is a clue about what type of number system is being used. In the binary system, two numbers, 0 and 1, represent the code. Numbers are assigned a binary code based on factors of 2. In other words, if you start the number assignments at 1, you then multiply 1 by 2 to equal 2; then 2 by 2, which equals 4; 4 is multiplied by 2, which equals 8, and so on.

22. Fourier Transform

Attenuation describes what happens to the incoming x-ray beam as it passes through the patient. Some of the x-rays are completely absorbed by the patient and some are deflected from their desired course. The rest of the beam passes through the patient with no interaction. Attenuation is a complex process affected by the strength of the x-ray beam, the thickness of the anatomy and the probability that a particular x-ray will be absorbed by a given tissue such as bone or soft tissue. After the attenuated x-ray beam strikes the detector array, a logarithmic conversion is performed to calculate just how many x-rays actually hit the detectors. This mathematical operation is known as a Fourier transform.

The Fourier transform calculates the length of time, or attenuation time, it takes for the x-rays to pass through the patient and strike the detectors. This attenuation-to-detector time is referred to as a phase. Each phase, or segment, of the patient’s scan is then assigned a value indicating how much power, or intensity, the segment produced. The intensity scale is referred to as amplitude. Next, the attenuated values are run through a series of equations. The equations divide the values into different subdivisions and each subdivided segment is assigned a different value. The new values are called Fourier coefficients.

Fourier coefficients are what make the construction of a CT image a reality. The new attenuation calculations, or Fourier coefficients, are then recalculated. These new calculations are called linear attenuation coefficients. The calculation uses the total average value of each detector to display the attenuation of the x-ray beam. After the segments of attenuated x-ray data are calculated and assigned new values, the data are sent by the log amplifier to the analog-to-digital converter.

23. Data Transmission to the Computer

Remember from a previous discussion that the analog-to-digital converter takes the analog signal and transforms it to a digital signal. In other words, it digitizes the information. Before the new digital information can be displayed, however, several corrections or preprocessing techniques must take place.

In addition, different types of filters must be applied to the data. Because the digital information contains only parts of the actual scan data, there can be sampling errors, as well as potential artifacts. The computer software ensures that the images are free of artifacts. Let’s look at two problems caused by incomplete data: aliasing artifacts and ring artifacts.

24. Aliasing Artifacts

Aliasing artifacts appear as unwanted streaks in the image. The artifact occurs when the sampler does not collect enough bits of the signal for the analog-to-digital converter. Insufficient sample
collection leads to streaks in the image because there is simply not enough information to fill in the gaps. The streaks occur when the computer adds information in the place of the missing data.

25. Ring Artifacts
   Additional preprocessing ensures that the data were calculated correctly. For example, ring artifacts appear as a circular pattern in the image and indicate that data from a detector element are missing. The computer software checks for faulty detector elements by searching the data for uniform detector readings. If detectors do not measure the attenuated beam or provide less information than the surrounding detectors, ring artifacts can occur. To correct this issue, the detector element must be repaired.

26. Convolution, Backprojection and Iterative Algorithm
   After correcting the initial scan data, the computer software creates a reformatted raw data set, which then goes through a process known as convolution. In convolution, the reformatted raw data are assigned different filters, which are used to enhance the desired anatomy of interest. Essentially, this step is where the process of backprojection used to occur.
   
   During convolution or backprojection, the reformatted raw data are copied to create what looks like the original raw data. Backprojection doesn’t create an exact duplicate, but makes an educated guess as to what the data should look like. Backprojection combines the incoming raw data with what a similar scan would look like, and creates a data set that represents the anticipated outcome.
   
   The backprojection process was initially difficult, so an alternative method, called an iterative algorithm, was created. An iterative algorithm uses what should be the likely outcome of the scan. The computer’s anticipated scan is compared with the actual scan, and the two are averaged together. This process is repeated again and again until the scan data and anticipated data are the same, or nearly the same. Backprojection and iterative algorithms are not used in CT today because of numerous limitations such lack of image sharpness and the length of time for processing.

27. Analytic Reconstruction Algorithm
   Modern CT scanning uses an analytic reconstruction algorithm to work with the data set. During this process, the Fourier transform once again divides the incoming scan data into individual segments, which are assigned numbers, or coefficients. The Fourier transform determines how the scan data are reflected back to the scanner. This data “copy” is called a filtered backprojection. The filtered backprojection then uses convolution to assign filters and remove certain types of artifacts. The convolution process within filtered backprojection is very important to how the scan data will appear to the viewer.

28. Kernel
   The kernel, sometimes referred to as a convolution algorithm, is the part of the process in which the computer adds a filter to the scan data. Unlike those found in diagnostic radiology, these filters are not made of plastic or aluminum; they are software algorithms that enhance certain types of anatomy based on the viewer’s requirements.
   
   Examples of these filters include the standard convolution algorithm, a smoothing filter and edge enhancement. These filters add detail to bone and soft tissues or help balance areas of similar densities. After filters are added to the incoming scan data, the newly-created data are put through the traditional backprojection method. The data are essentially copied into another set of scan data that is now free of artifacts.

29. Interpolation
The final step before the viewer can see the scan is called interpolation. Interpolation is a mathematical process that enables the spirally acquired images to appear as individual slices. In other words, the images are “flattened” to look as though each slice were acquired separately. The image data are then sent to the computer and separated out into a viewable format.

30. Digital Acquisition System Performance
So far we’ve looked at the computer algorithms that affect the data acquisition system, but several other factors influence how the DAS performs. These include the radiographic tube, detectors, filters and collimators. Let’s look at each of these individual components.

31. Radiographic Tube
Early CT scanners used an x-ray tube that had a stationary anode and was cooled with oil. The parallel beam geometry of first generation scanners produced a very narrow x-ray beam. During the scanning process, the x-ray tube and detectors moved 1 degree, created an exposure, and then moved another degree, created another exposure, and so on, until all the images were acquired. This scanning method allowed the anode to cool off between exposures. The x-ray tube remained stationary in second generation scanners even though the beam geometry changed to a small fan beam.

32. Radiographic Tube
Third generation scanners use a wide, fan beam geometry that is similar to the fan beam used in second generation scanners. The most significant change in x-ray tube construction for third generation scanners was the addition of a rotating anode. The rotating anode allows continuous x-ray production while withstanding the increased heat load to the anode.

The x-ray tubes in third generation scanners must provide uniform x-ray production, dissipate heat quickly and be cost effective. The cost effectiveness of an x-ray tube depends on its tube life, as well as the durability its parts, such as the bearings and rotor assembly.

33. Practice Question
34. Practice Question
35. Practice Question
36. Detectors
Detectors are small devices located opposite the x-ray tube. Situated in an array of small squares, the detectors are placed in rows, with the number of rows determining the number of slices the CT scanner can acquire. In fact, the addition of several detector rows changed the single-slice CT scanner to a multislice scanner. The detectors are responsible for receiving the attenuated x-ray beam and recording each x-ray encounter as a measurable incident. Detectors contain either gas or scintillation crystals.

37. Gas Ionization Detectors
Gas ionization detectors are made up of individual gas cells. Each cell is filled with a stable gas that produces little afterglow, or additional light, following an x-ray photon interaction. Xenon gas typically is used in these detectors.

The detectors themselves are constructed from tungsten plates. The plates separate the cells from each other and attract the ions produced from x-ray interaction. When the x-rays strike the detector, they ionize the gas, creating both positive and negative ions. As the ions drift to the tungsten
plates, a signal is produced. The signal intensity is proportional to the number of x-rays that hit the detectors. The data acquisition system detects the electrical signal and amplifies the signal for the image production phase.

38. Scintillation Detectors

Most modern CT scanners use rows of detector elements covered with scintillation crystals. A variety of crystals are available, but regardless of the type, the crystal must be able to produce light quickly and then return to its original state. Scintillation detectors are preferable to gas ionization detectors because they produce light efficiently.

When the x-ray photons strike the scintillation detectors, the crystals release small amounts of light. Because the light isn’t a strong enough signal, it must go through a photomultiplier tube, or photodiode. The photomultiplier is located just behind the crystals on the detectors. When the crystals release light, the light strikes the photocathode, which turns the light energy back into electrons. The electrons then are focused toward an arrangement of dynodes, which in turn release their own signals. These signals are multiplied over and over again to strengthen the signal. The amount of signal at the end of this process is equivalent to the amount of light given off by the crystals. The electric signals then travel to the data acquisition system, which processes the signals for computer display.

39. Detector Properties

The reliability of detectors depends on several properties: efficiency, response time, the dynamic range, reproducibility and stability. Let’s look at each of these characteristics.

Efficiency can be described as the ability of the detector to assimilate the incoming x-ray photons. The detector must be able to capture any x-ray photons attenuated from the patient and then absorb those photons to allow new incoming x-rays to hit the detectors. The detectors also must be able to quickly convert the incoming x-ray photons to light.

Response time is the amount of time necessary for the crystals to produce light, to have the light picked up by the photomultiplier and then for the light to totally dissipate from the crystals. The crystals must completely rid themselves of light from each and every x-ray interaction.

The dynamic range refers to the range of measurements, or light readings, within the same scan. In other words, the dynamic range is the ability of the detector to “see” measurements from one million light encounters to just one light encounter. The most common dynamic range in today’s CT scanners is one million to one.

Reproducibility means that a sequence of events must be performed correctly and identically time after time. A detector converts light into an electric signal in milliseconds. Each light signal picked up by the photomultiplier comes from crystals on the detectors. As soon as the photomultiplier picks up the light signal, the crystals must be free of any light, or afterglow, and be ready to produce the exact same intensity of light from the next x-ray photon interaction. This process must occur millions of times and in exactly the same way to produce CT scans with identical outcomes.

The stability of a detector refers to how precisely each x-ray photon is read, how fast the light from the crystals dissipates and how the detector works in relation to the detectors around it. If the detectors are not stable in any given parameter, the CT images may contain artifacts.

40. Detectors and Data Acquisition

The detector array, or rows of detector elements, defines the slice thickness of the scan. The more rows of detectors in a scanner, the thinner the slices. Multislice scanners can perform faster examinations because the multiple detector rows can collect more information for each rotation of the x-ray tube.
41. Filters
The x-ray beam from the tube is filtered through a specially-designed material that removes low-energy, or long wavelength, photons and ensures beam uniformity at the detectors. Removing the long wavelength photons reduces patient dose because the patient’s skin absorbs these photons. In addition, the average energy of the x-ray beam is increased, resulting in a “harder” beam.

42. Filters and Data Acquisition
Beam hardening occurs when the filter removes many of the weaker x-rays from the beam. Let’s look at the following example. Imagine that the x-ray beam is a series of numbers and each x-ray is an individual number. Calculate the average for the series by adding the numbers and dividing by the number of x-rays. In this case, the sum of the series is 184, divided by 18 equals an average of 10.22.

Using a filter removes the weaker x-rays, so let’s remove all the numbers below 10 in our example. The average of the remaining numbers is 11.17, or slightly higher than our original average. In the example, removing the smaller numbers from the series produces a higher average; in a CT scanner, filtering low-energy photons results in higher average beam intensity.

43. Collimators
Collimators are small metal masks that ensure uniform slice thickness. CT scanners contain two types of collimators: prepatient collimators, located on the tube side, and postpatient collimators, located at the detector side.

44. Prepatient Collimators
Located just outside the x-ray tube, the prepatient collimators must be aligned exactly with the postpatient collimators to ensure that slice thicknesses are identical. The width of the prepatient collimators determine slice thickness. The CT technologist selects the slice thickness based on the desired detail for the scan. If a really narrow width, or thin slice, is selected, the amount of time necessary to cover the anatomy of interest increases.

However, using smaller slice thicknesses means that more data are collected during a scan, thus giving the analog-to-digital converter more samples of information. This result occurs because thin slices are distributed over fewer detectors. Each detector is saturated with more total x-ray photons and now collects much more data for the scan. Increased detail comes at the expense of an increased scan time and increased patient dose.

45. Postpatient Collimators
The postpatient collimators are located just above the detector array. They maintain the correct slice thickness and ensure uniformity of the x-ray beam at the detectors. Postpatient collimators do not affect patient dose because the x-ray beam has already passed through the patient. Another important function of postpatient collimators is to remove scatter radiation emerging from the patient.

46. Practice Question

47. Practice Question

48. Selectable Scan Factors
The CT technologist can select several scan factors during data acquisition. Adjusting any one of these factors can have an impact on the resulting image. Some factors only affect the manipulation of the image data, while others permanently change the image display data.
49. **Scan Field of View (SFOV)**

The scan field of view, or SFOV, is the region or area the CT technologist typically selects during scanning. It is especially important for the quality of certain exams. A matrix, or checkerboard grid, is usually positioned over the selected scan field of view. The scan field of view is responsible for the spatial resolution, or fine details, within an image. When the scan field of view is selected and tightly collimated to the area of interest, pixel size is affected.

Today’s CT scanners typically use matrices of either 512 x 512 pixels or 1024 x 1024 pixels. Matrix size never changes, but the pixel size can vary based on the selected scan field of view. Higher spatial resolution requires many more pixels to provide the desired details.

The images on this page show the same anatomy with two different scan fields of view. The image on the left displays a large scan field of view and the image on the right shows a small scan field of view.

50. **Display Field of View (DFOV)**

The display field of view, or DFOV, is the size of an area shown after the patient has been scanned. The display field of view does not contribute to patient dose because it is selected after the scan has been performed. The display field of view also is not able to recreate, or gain, any new scan data; therefore, it’s important for the CT technologist to select the appropriate-sized scan field of view to yield the best spatial resolution for the scan.

Pixel sizes can vary greatly, from smaller than 1 mm to 10 mm in size. To calculate the pixel size of the scan, the display field of view is divided by the size of the matrix. Let’s look at an example using a 512 x 512 matrix and a display field of view of 30.2 cm. First, the DFOV needs to be converted from centimeters to millimeters. Moving the decimal point 1 space to the right changes 30.2 cm to 302 mm. Only the first number of the matrix is used in the equation. So for our example, a 302 mm DFOV is divided by a matrix of 512, yielding a pixel size of 0.59 mm.

51. **Matrix**

A matrix is essentially a checkerboard pattern laid over an image before the scan is performed. Remember that the typical matrix format is either 512 x 512 mm or 1024 x 1024 mm. To determine how many matrix cells or pixels are available, multiply the length of the matrix by its width. So a 512 x 512 matrix would have 262,144 matrix cells, or pixels.

A 1024 x 1024 matrix improves spatial resolution because there are more matrix cells that can house more pixels. The matrix is usually a preset parameter in a CT scanner and cannot be changed. It is an important consideration when purchasing equipment, however, especially if spatial resolution is important.

52. **Slice Thickness**

The slice thickness is targeted in the scan field of view. Remember that the slice thickness is also formed by the collimators and that both the prepatient and postpatient collimators should be aligned exactly to ensure a uniform slice thickness during the entire scan. The prepatient collimators and detector array are actually bigger than the intended slice thickness because there is magnification from the x-ray tube and at the detector array. Magnification occurs in these places because the “true” slice thickness is at the center point.

The pitch of an exam directly affects the slice thickness. Pitch is the width of the helical/spiral path around the patient. When a pitch of 1.0 is used, the slices are next to each other, with no overlap or gaps between them. An equation, called the pitch equation, is used to determine the pitch.
Slice thickness affects how the details of an exam will appear. To increase spatial resolution, the slice thickness needs to be very thin. The CT/Hounsfield numbers assigned to the pixels may change based on the slice thickness.

53. Spacing/Reconstruction Interval
The scan time varies depending on the slice thickness within the scan. Very thin slices actually overlap, with the amount of overlap based on the pitch used. The pitch is determined by how fast the table moves through the gantry. If the table slows down, the scan time increases, and the patient remains in the x-ray beam for a longer period of time. In addition, the slices typically overlap, which improves spatial resolution.

The spacing interval is the term used to describe how much distance should be between each slice. If an examination needs greater detail, such as in computed tomography angiography, the pitch needs to be adjusted to less than 1.0. Using a pitch of less than 1.0 causes each slice to overlap the adjacent slice.

Setting a pitch of 0.75 means that 25% of each slice overlaps the slice next to it. The overlap creates better spatial resolution because that portion of the scan is rescanned during each rotation of the x-ray tube and detector array. Conversely, using a pitch of greater than 1.0 means there would be gaps between each slice. For a pitch of 1.2, there would be a 20% gap between each adjacent slice.

The reconstruction interval, also termed reconstruction increment, or RI, is specific to helical/spiral scanning and is set to ensure the best resolution for the examination. The RI is similar to pitch in that image slices overlap; however, the RI is more of a preset parameter, or preset within a scan protocol for any given exam.

54. mAs and kVp
The term mAs, or milliampere seconds, describes the total amount of x-rays created for a given exam. This factor affects several scan parameters including noise and contrast resolution. Longer scan times are required at lower milliamperage to compensate for the reduced number of electrons generated. If scan time is not increased, image noise may occur when too few electrons are delivered to the anode focal track.

To reduce noise, mAs must be increased. To preserve spatial resolution, the time in seconds must be increased so that more data is acquired to produce the image. The kVp, or kilovolt peak, is responsible for the penetrating power of the x-ray beam and can influence several different scan parameters, including beam hardening, scatter, and the consistency of the CT/Hounsfield numbers, contrast resolution and noise.

Beam hardening occurs when the x-ray beam passes through a very dense area and the tissues absorb the lower-energy, or lower kVp, photons. This action then increases the average power of the remaining photons. An x-ray beam with too much beam hardening produces broad streaks on the image and affects the CT/Hounsfield numbers that are calculated from the image data.

The center of an image affected by beam hardening has decreased CT/Hounsfield number assignments based on x-ray beam changes during attenuation.

The kVp directly influences scatter. A lower kVp setting reduces scatter, and when the selected kVp increases, scatter increases. However, increased kVp also means that thicker body parts can be scanned. A correct kVp setting is crucial to demonstrate subtle changes of gray. If the selected kVp can’t adequately penetrate the anatomy of interest, the resulting images are grainy and subtle changes are more difficult to see.

55. Kernel
A kernel is a selectable scan parameter that applies a series of mathematical algorithms to the incoming scanned data. These algorithms give the data a desired appearance. The kernel is preset using a histogram, which is a curved line showing what the scan should resemble. This page shows an example of a bone histogram and a soft tissue histogram.

56. Bypass Filter
Different types of filters correct the incoming data. Sometimes referred to as high-pass filters, they include edge enhancement filters, the standard convolution algorithm and a smoothing filter. An edge enhancement filter is applied to scans needing a sharper, crisper image. This filter is used for orthopedic examinations and scans of dense, bony structures.

The standard convolution algorithm is used for scans needing soft tissue adjustments. This filter adds definition to the slight changes in densities within soft tissues. Both the edge enhancement filter and standard convolution algorithm are used for scans needing increased spatial resolution.

The smoothing filter is applied to scans that display very low contrast resolution. This filter highlights areas with small changes in the shades of gray. It’s used for areas such as the abdomen and pelvis, or areas of similar densities.

57. Scan Time and Rotational Arc
CT scan time is based on the amount of anatomy to be scanned, the slice thickness, the milliamperage selected, the generation of CT scanner and the number of detectors in the scanner. Longer scan times are required to cover the anatomy of interest when slices overlap. Overlap occurs with thinner slices and when the pitch is less than 1.0, so scan time can vary depending on slice thickness. Click on the buttons to see the variation in scanning times.

When a lower mA is used, such as for exams requiring increased spatial resolution, the scan time must be increased to allow the appropriate amount of x-ray photons to pass through the patient. Modifications and improvements to CT scanners have affected scan times significantly. Changes in the way the x-ray tube and detectors rotate around the patient and the addition of more detectors have decreased scan times.

For example, scan times for first generation scanners were quite long because the x-ray tube and detectors had a limited rotational arc and there was only a single detector. Second-generation scanners had shorter scan times because of the development of the fan beam and the addition of more detectors. Third-generation scanners use a bigger fan beam x-ray geometry, and the x-ray tube and detectors rotate continuously around the patient.

58. Rotation Time
The movement of the x-ray tube and detectors affects the rotation time. In first and second generation scanners, the rotation time was affected by the interscan delay, which was the amount of time needed for the high-tension cables to unwind between x-ray exposures. Slip-ring technology in subsequent generations of scanners allowed the x-ray tube and detectors to rotate continuously, eliminating interscan delay.

59. Region of Interest (ROI)
A region of interest, or ROI, is a tool used for quality control purposes. The ROI is placed within the scan field of view of a water phantom image. It measures the noise level of the scanner and the CT/Hounsfield number for water. An ROI also can measure different densities within an image to help the radiologist interpret the scan.

60. Magnification
Magnification enlarges the image and typically is used for close-up views of the anatomy. Magnification, when used as a tool, does not add or subtract image information because it is a postprocessing technique.

The image may be magnified if the patient is placed above or below the correct z-axis. This phenomenon occurs because the scan field of view is located at the center of the fan beam, midway between the x-ray tube and the detector array. This center point is sometimes referred to as being isocentered.

61. Focal Spot Size and Tube Geometry
The focal spot size is determined by the filament selected in the cathode and by the target angle of the anode. With regard to focal spot size, the tube geometry is related to the target angle of the anode. The milliamperage determines the filament. If a lower mA is selected, the CT scanner uses a smaller filament size. The small filament produces a thinner x-ray beam, which increases spatial resolution. When a higher mA is selected, the larger filament, located at the cathode, is directed to the larger target angle of the anode.

62. Dose Modulation
Automatic milliamperage, or auto mA, is used to reduce patient dose. Sometimes referred to as dose modulation, auto mA is similar to the automatic exposure control used in diagnostic x-ray. Auto mA determines the length of exposure needed to fully penetrate the patient’s anatomy. As soon as the appropriate amount of x-ray photons strike the detectors, the x-ray exposure turns off automatically.

Auto mA usually occurs during the topogram, or scout, image. The topogram is significant because the scanner includes metallic items or areas of increased density in the mA assigned to the scan.

63. Pitch
An ideal pitch is 1.0 because each slice in the scan is positioned next to one another, with no overlap. When the pitch is greater than 1.0, the scan time decreases, and the patient dose also is lower. However, the examination contains less scan data, and there are gaps between image slices. Increased pitch results in partial volume averaging, an artifact that leads to degraded image contrast.

Partial volume averaging occurs when the data acquisition system is forced to assign CT/Hounsfield numbers to missing information based on the data around the gap. Small differences in densities are not as apparent because the assigned CT/Hounsfield numbers were properly calculated based on actual data. Click on the button “Pitch equals 1” and let the animation play through until the end. Then click on the button labeled “Pitch equals 2.”

64. Practice Question

65. Practice Question

66. Conclusion
This concludes CT Basics Module 3 – Data Acquisition. You should now be able to
- Name the methods of acquiring computed tomography images.
- Explain the functions of the data acquisition system (DAS).
- Describe the process of data acquisition.
- Identify the factors that influence data acquisition.
- List the selectable scan factors used to acquire images.
67. Development Team

68. Final Slide