2. CT for Technologists
Image Formation, Reconstruction, and Artifacts

CT for Technologists is a training program designed to meet the needs of radiologic technologists entering or working in the field of computed tomography (CT). This series is designed to augment classroom instruction and on-site training for radiologic technology students and professionals planning to take the review board examinations, as well as to provide a review for those looking to refresh their knowledge base in CT imaging.

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This material will be reviewed for continued accuracy and relevance. Please go to www.icpme.us for up-to-date information regarding current expiration dates.

Note: Terms in **bold** can be found in the glossary.

OVERVIEW

The skill of the technologist is the single most important factor in obtaining good quality diagnostic images. A successful CT examination is the culmination of many factors under the direct control of the technologist.

CT for Technologists:2・Image Formation, Reconstruction, and Artifacts introduces the learner to the concepts of the image process, scanning methods, the digital image, image quality, image reconstruction, quality assurance tests, and artifacts. Many factors contribute to image quality, and a thorough understanding of the interactions among these factors allows the technologist to acquire high-quality, diagnostic images.

EDUCATIONAL OBJECTIVES

After completing this material, the reader will be able to:

- List the various types of CT scanning
- Discuss the interactions of parameters, reconstruction, and evolving technologies of image formation
- Describe the details that contribute to the digital image and to image quality
- Discuss image reconstruction and postprocessing techniques
- Identify the different types and sources of CT artifacts by their characteristic patterns
- Reduce or eliminate the appearance of artifacts through continual quality assurance and assessment of technical parameters, including phantom testing

EDUCATIONAL CREDIT

This program has been approved by the American Society of Radiologic Technologists (ASRT) for 2.75 hours of ARRT Category A continuing education credit.

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Victoria Phoenix, BS, has no conflicts to report.
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2. CT for Technologists

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OVERVIEW OF THE IMAGING PROCESS

The total computerized tomography (CT) imaging process involves the production of x-rays, the mechanics of the tube motion and table travel, and the role of the computer for image reconstruction, presentation, and storage **(Figure 1)**.

Figure 1. **CT scanner.**

Courtesy of Siemens Medical Systems

Image Formation

Image formation refers to processes in which the information taken from the scan is converted into "raw" digital data. These raw data may then be converted to a numeric "map" representing the final image. Image formation begins once x-rays have been produced and directed towards the patient on the table.

The image formation process includes:

- Multiple projections are taken at 90° angles to the tube for a full 360° per slice as the table moves through the **gantry**.
- A narrow x-ray beam passing through the patient is attenuated by the patient's body, with varying degrees of intensity.
- The attenuated x-rays strike the **detector**s in the gantry and collect the x-rays that pass through the patient.
- The x-rays are sampled, amplified, and converted into **digital signal**s by the **analog-to-digital converter** (ADC) as raw data (also called projections).
- An **algorithm** is applied to accomplish the desired look.

Image Reconstruction

Image reconstruction refers to the application of mathematical algorithms to the acquired raw data to create an image.

The image reconstruction process includes:

- The computer takes the raw data and applies a mathematical formula called **filtering** to a **back-projection** of the image.
- This filtering (the application of reconstruction algorithms) of the raw data is also used to sharpen the edges of the attenuated data so that it is free of streaking and star **artifacts**.
- The filtered back-projected image is displayed on an image grid (or **image matrix**) on a CT monitor.
- The image is now in a form where it can be permanently archived or printed.

SCANNING METHODS

Three single-source scanning methods are most commonly employed in CT imaging: **localizer scans**, conventional CT scans, and spiral/helical CT scans. Each is used to achieve a different objective.

KEY TERMS
localizer scanning
scout
axial CT/conventional CT
spiral/helical scanning
dual-source CT
dual-energy CT

Localizer Scanning

The localizer scan — also called a **scout**, topogram, or reference image — produces an image that looks very much like a plain x-ray and is performed as part of every CT examination. These images are acquired as the table continuously moves under the stationary tube. *Anterior/posterior* (AP) localizer scans are obtained by continuously moving the table under the tube located above the patient (12:00 position); a *lateral* localizer is obtained by moving the table with the tube 90° (9:00 position) to the patient. The length of the scan depends upon the scan time and speed of the table as it passes through the **aperture**. The fixed design of the detectors limits resolution but improves **contrast** due to the reduction in scatter radiation. This is the simplest of CT scanning methods because the single-projection technology does not require complex reconstruction algorithms.

Axial Scanning

Axial or **conventional CT** scanning is characterized by a start-and-stop rotation of the tube around a stationary patient table. After each 360° rotation of the tube is completed, the table moves incrementally according to specific protocols to obtain the next slice. This process is repeated until collection of the desired number of scans or slices is achieved. These scans can be performed singly, in multiples, or in batches and are sometimes referred to as –step-and-shoot” scanning. Axial scans are almost never performed in general body CT but are still done for chest CT because of the high-resolution imaging. Axial CT is often found to be superior for brain imaging, and **spiral/helical scanning** is finding its place in braining imaging, as well **(Figure 2)**.

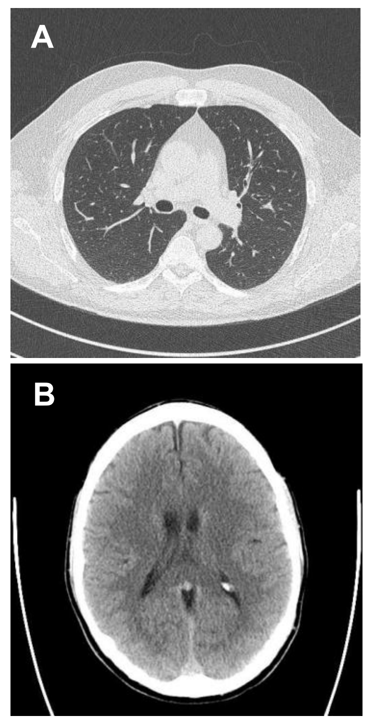


Figure 2. Axial single-source images. (A) Chest. (B) Brain.

Spiral/Helical Scanning

Spiral/helical scanning, also known as **volume acquisition**, refers to the continuous tube rotation while the patient moves under the beam. If the beam could be traced along the patient, the path would look like a coil **(Figure 3)**. Spiral scanners require **slip ring** technology, which allows the gantry to turn continuously around the patient without interruption as the patient table slides through the aperture. The heat generated from the continuous rotation of the scanning tube requires large-capacity heat units for heat tolerance and cooling.

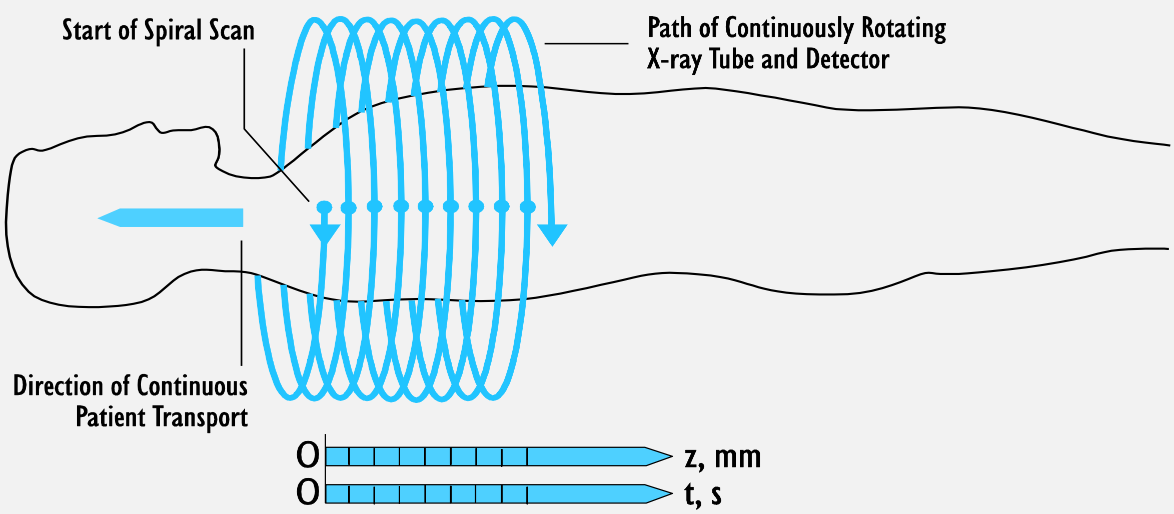


Figure 3. **Spiral/helical data acquisition showing the path of the scanner.**
Courtesy of Siemens Medical Systems

Spiral/helical scanning offers several advantages over conventional CT **(Figure 4)**:

- Overall period of acquisition is faster
- Faster scan time allows a set of images to be taken over an entire body region in a single breathhold
- Breathholds are shorter
- Larger volumes of information are acquired
- Ability to reconstruct overlapping thin slices significantly improves 3D reconstructions
- Less iodinated contrast material is needed for scans that require contrast

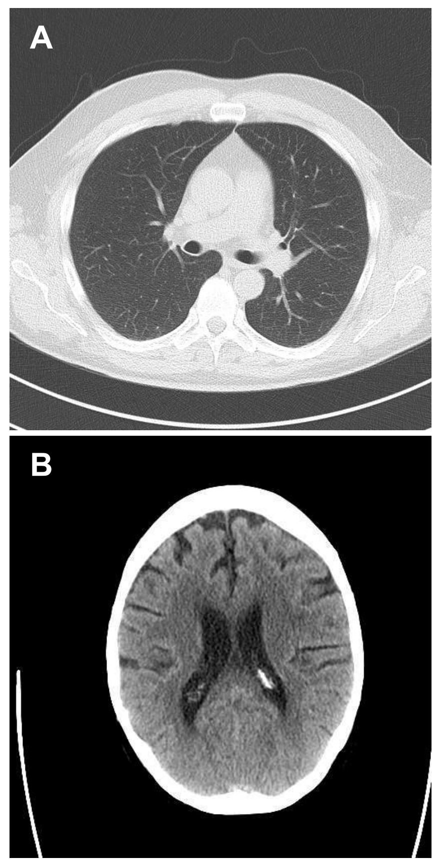
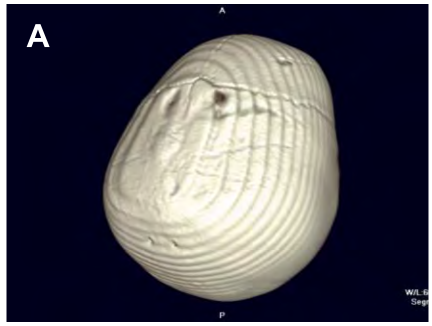


Figure 4. Spiral/helical image. (A) Chest. (B) Brain.

The continuous scanning process allows the image to be initiated from any point in the scanned tissue volume, but the data from helical scans must be interpolated for an accurate reconstruction to occur. **Interpolation** is the process used to predict and calculate the image value between slices based on previously acquired data. With the advent of multi-slice scanners, the need for interpolation is less of an issue because of the greater number of detectors.

The slice thickness and spacing prescribed to the images allow for true **isotropic** resolution, which means uniformity in all directions, for example, for a 0.5x0.5x0.5mm **pixel** size. Isotropic resolution is ideal for imaging 3D structures because **spatial resolution** is identical in all dimensions; in other words, spatial resolution will appear the same no matter from which plane the image was acquired. Large numbers of images can be produced; the stair-step artifacts that can occur when slices are not contiguous are virtually eliminated, resulting in well-defined edges **(Figure 5)**.



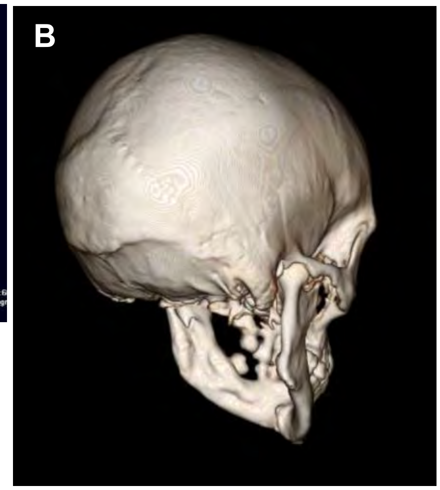


Figure 5. Surface shaded skull.
(A) With stair-step artifact.
(B) With no stair-step artifact.

Dual-source CT

Dual-source CT (DSCT) was introduced in 2006. DSCT utilizes two tubes and two detectors rotating in the same rotational plane at the same time **(Figure 6)**. The use of two detectors helps achieve ultrafast motion-free imaging of structures while applying low dose parameters and maintaining diagnostic imaging quality. DSCT is especially useful in cardiac CT because it:

- Acquires a volume of data through the entire cardiac cycle.
- Can scan at any heart rate.
- Reconstructs multiphase series for 4D viewing.
- Measures left ventricular function.
- Scans in sub-second time.
- Potentially decreases radiation dose to approximately 1 mSv.

Figure 6. Dual-source CT scanner.
Courtesy of Siemens Medical Systems

Bariatric patients

DSCT is useful for the scanning of obese patients. The 200kW generator provides increased power, resulting in the tubes increasing the mAs readings and thereby allowing the proper penetration. A wider gantry better accommodates the bariatric patient.

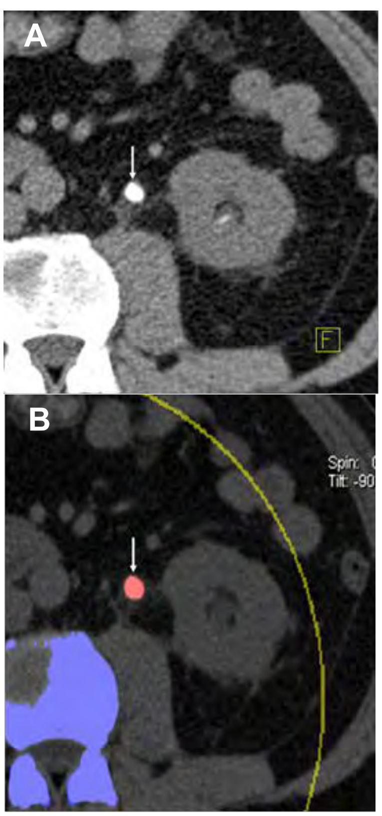
Pediatric patients

DSCT is also helpful for scanning pediatric patients. The use of two tubes increases **pitch** (P) by a factor of three. This increase in pitch results in a lower radiation dose to the patient and decreases scan time to about 0.5, sec eliminating the need for anesthesia. Keep in mind that patients ages 1-2 years will still need to be restrained in order to acquire quality diagnostic scans.

Dual-energy CT

Dual-energy CT (DECT) is an extension of dual-source CT. Each of the two tubes has its own tube voltage, providing tissue differentiation by **absorption** of the energies. DECT enables differentiation of materials based on their distinct absorption of **photon** energy (fat, soft tissue, calcium, or iodine) **(Figure 7)**. Dual-energy pre-filters high kV x-rays, eliminating overlap and allowing each energy source to be neutral from the other. Tube voltage (kVp) can be inter-changeable, dependent on patient size. Advantages of dual-energy over single-source CT are:

- Better bone subtraction from soft tissue
- Greater differentiation of materials
- “Virtual” non-contrast



THE DIGITAL IMAGE

The CT scanning procedure reads information as **analog signals** (also known as continuous waveform) that are then converted to a digital image by taking the original analog signals and converting them into **digital signals**. Although this may seem like a repetitive process, the computer could not be employed without the analog-to-digital conversion (ADC). The digital output of raw data in binary code (0s and 1s) cannot create a recognizable image. This should not be confused with the pixel values recorded in the final image from which the gray scale is created. The actual image is created when these raw data are converted back into analog signals by the **digital-to-analog converter** (DAC) for display or permanent storage. Using digital data also allows for significant manipulation of the image to improve its quality. These data can be manipulated by algorithms that enhance certain features of the image, tailoring the final image to answer the clinical question.

Figure 7. Images with and without use of dual-energy CT. (A) Kidney before applying dual-energy CT. (B) Dual-energy helps identify the composition of the kidney stone. Red represents a uric acid stone; blue represents calcification (here, part of the spine). The type of material revealed will direct the type of treatment required.

Pixel

A pixel or ―picture element" is defined as a two-dimensional imaginary unit or cell of information within the image matrix. A pixel represents a small square formed by the rows and columns of the image grid. The size of the pixel corresponds to the scan diameter; a larger diameter is associated with a larger pixel. Each pixel is assigned a gray value in the form of a **Hounsfield Unit** (HU) or **CT number**. The computer assigns these numbers based on calculations taken from the **filtered back projection** that directly corresponds to the **attenuation** coefficient of the object scanned (for more information on the **linear attenuation coefficient** and image reconstruction, see pages 9 and 21). Assignment of numbers involves comparing the original number to a material such as water (with a value of 0) for reference and multiplying it by a magnifying constant. All tissues are assigned a value of +1 or -1. The magnifying constant is used to express the accuracy of the scanner. For example, a scanner with a magnifying constant of 1000 would assign a value of +1000 to a dense tissue such as bone. In terms of display, higher HU values correspond with lighter shades of gray, while lower HU values correspond with darker shades. Thus, bone appears as bright white, while air appears dark.

Calculating pixel size

Pixel size is determined by converting the **display field of view** (DFOV) size from centimeters (cm) to millimeters (mm) and then dividing by the matrix size.

Formula:

$$\textbf{pixel size (mm}^2\textbf{)} = \frac{\textbf{(DVOF) (mm}^2\textbf{)}}{\textbf{matrix size}}$$

Example:

**Calculate pixel size using a 40 cm DFOV
and a matrix size of 512x512**

Answer:

$$\textbf{pixel size (mm}^2\textbf{)} = \frac{\textbf{(40cm) (10mm/cm)}}{\textbf{512}} = \frac{\textbf{(40mm)}}{\textbf{512}} = \textbf{0.781mm}^2$$

Voxel

A **voxel** or “volume element” is a three-dimensional representation of the tissue volume. The tissue volume is represented by a given pixel and the slice thickness.

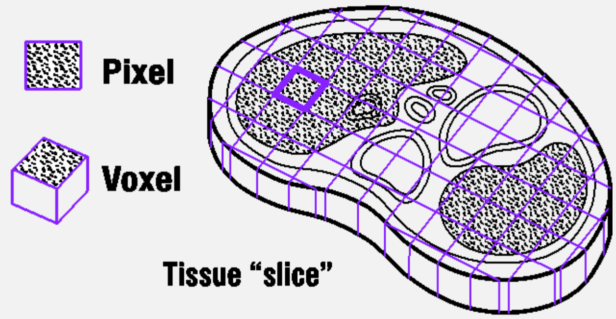


Figure 8. **Each section of body tissue is imaged as a grid of cubes, or voxels.**

Calculating voxel size

A voxel is determined by multiplying the pixel size by the scan slice thickness **(Figure 8)**.

Formula:

$$\textbf{voxel size (mm}^2\textbf{)} = \frac{\text{(DVOF) (mm}^2\text{) (slice thickness mm)}}{\text{matrix size}}$$

Example:

Calculate voxel size using a 40 cm DFOV,
a slice thickness of 3 mm,
and a matrix size of 512x512

Answer:

$$\textbf{voxel size (mm}^2\textbf{)} = \text{(40cm) (10mm/cm)} = \frac{\text{(400mm) (3mm)}}{512} = 2.34\text{mm}^2$$

Image Matrix

The image matrix is comprised of thousands of pixels divided into very small regions in varying shades of gray and represents the number of pixel elements from which the final image is created. The matrix size is determined by multiplying the number of pixels in the length by the breadth of the image grid. Most images are displayed on a 512x512 matrix for a total of 262,144 pixels. The first commercial CT scanners displayed images with an 80x80 matrix. Many scanners use 256x256 displays for “instant images” for monitoring the progression of the study, allowing the technologist to determine if the prescribed anatomic region has been completely covered. Final reconstructions to 512x512 are delayed to a later point in the study, usually as the scanned patient gets off the table and the next patient gets on the table. Very high-resolution scanners have image grids of 1024x1024 or greater **(Figures 9, 10)**.

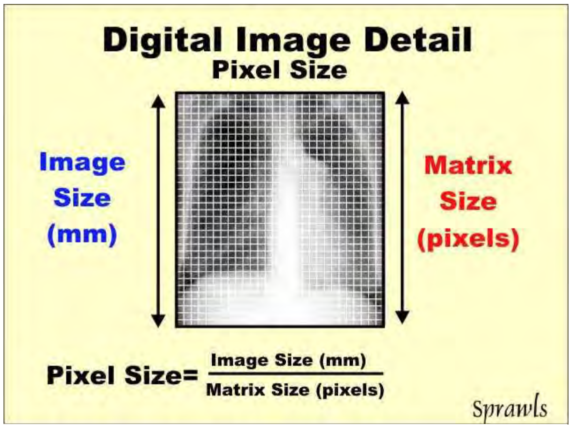


Figure 9. **Image matrix representation.**

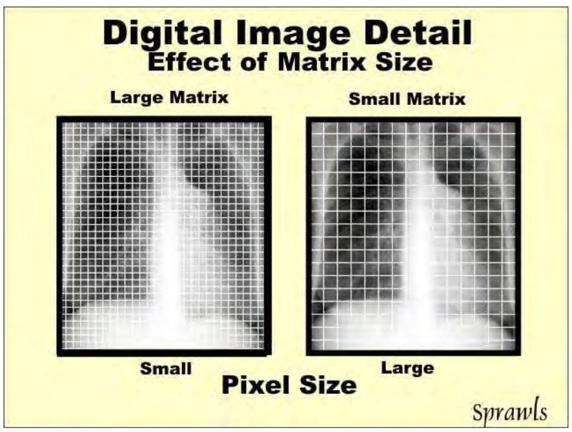


Figure 10. **(Left) Larger matrix shows higher resolution with increased noise. (Right) Smaller matrix shows lower resolution with less noise.**

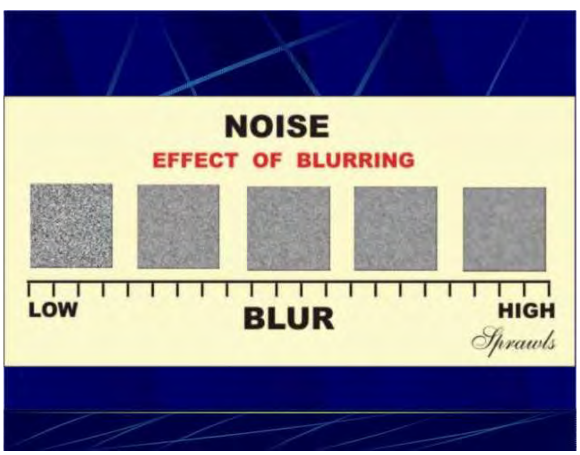


Figure 11. **Effect of noise on blurring.**
Courtesy of Sprawls Educational Foundation

Matrix size generally represents a compromise between spatial resolution and **noise**, which is a blotchy or spotty image. A small matrix is comprised of larger pixels, which decreases noise but also decreases resolution. Conversely, a larger matrix increases the resolution of the CT image by using smaller pixels, but the amount of noise is also increased **(Figure 11)**. When using a larger matrix to increase resolution, adjustments to other parameters — such as increasing tube current and patient radiation dose — must be made to counter the increase in noise.

Linear Attenuation Coefficient

The linear attenuation coefficient represents the rate at which x-rays are attenuated, or diminished, after passing through a patient. Attenuation occurs via two mechanisms: scatter and absorption. The amount of attenuation depends on four variables: atomic number, density of the tissue, thickness of the tissue, and the level of photon energy. Graphing the equation demonstrates that the number of photons detected decreases exponentially as tissue thickness increases. The type of beam determines the energy level of the photons detected: a **mono-energetic beam** produces photons having the same energy, while a **poly-energetic beam** produces photons of varying energies. Scanners that do not employ tin filters throw poly-energetic beams. A type of artifact known as **beam hardening** often occurs when a poly-energetic beam is used. In this case, the lower energy photons attenuate while the higher energy photons remain, increasing the overall energy level of the beam. Scanners now employ correction filters that improve the image while reducing the beam hardening artifact. This improvement is seen especially in the posterior fossa of the brain and in the ribcage, where beam hardening can cause streaking into the anatomy (for more information on beam hardening artifacts, see page 35).

Hounsfield Units

As mentioned earlier, the information contained in a single pixel is assigned a numerical value called a Hounsfield Unit or CT Number **(Table 1)**. The HU value corresponds to a gray scale value. Since the perfect value of water is 0, this is the value to which the system is calibrated. The atoms with the lightest weight (air) are assigned the lowest HU value (-1000), and the densest materials scanned (bone) are given the highest HU value (+1000 or greater). The HU value determines the gray level of the voxel element. Water is assigned an intermediate shade of gray, air is assigned black, and bone is assigned white. As tissues become denser, their HU values increase and they will appear lighter to white. Thus, fatty tissues have negative numbers in the range of –100HU, and soft tissues have positive numbers in the range of 30HU. As the HU value increases, the ~~–~~whiteness" of the voxel increases **(Figure 12)**.

Tissue	HU Unit
Air	-1000
Lungs	-150–400
Fat	-100
Water	0
Tumors	+5–35
Blood	+13–18
Cerebrospinal Fluid	+15
Brain–gray matter	+20–40
Brain–white matter	+36–46
Muscle, aortic muscle	+35–50
Soft tissue (liver, kidney, pancreas, spleen)	+40–70
Blood, coagulated	+55–75
Bone	+100–1000
Petrous bone	+3000

Table 1. Hounsfield Units (CT Numbers) Assigned Various Tissues

Adapted from RR Carlton RR and AM Adler. Principles of Radiographic Imaging. Albany, NY: Delmar Publications, 1996:634.

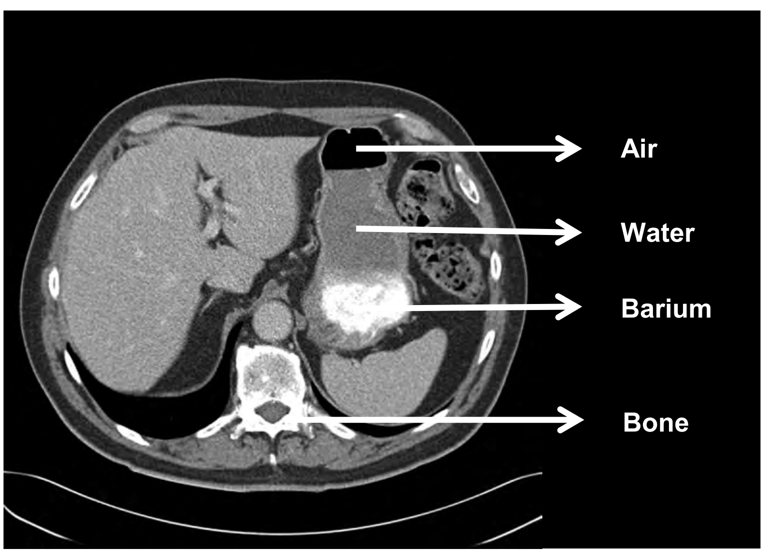


Figure 12. CT scan showing bone and barium (+HU value), water (neutral HU value), and air (-HU value) in the stomach.

Milliampere

A **milliampere** or mA is the tube current that flows down the x-ray tube from the **cathode** filament to the **anode**. When the current strikes the anode, x-ray photons are produced. The **mAs** (milliampere x scan time) is a scanning parameter used to reduce image noise. A high mAs produces a higher number of x-ray photons, thereby reducing the noise in a given image without affecting image contrast, that is, the greater the number of photons, the lower the level of noise. Increasing mAs reduces scan times, which in turn decreases the chance of motion artifacts. The primary disadvantage to increasing mAs is that it exposes the patient to a higher radiation dose, and therefore careful consideration should be given to selecting the appropriate mAs level.

Kilovoltage Peak

Tube voltage from the cathode to the anode is expressed in **kilovolts** (kV). The energy of electrons is expressed as keV, which in turn dictates the maximum energy potential of x-ray photons. The maximum tube voltage is known as **kilovoltage peak** (kVp). The ability of x-rays to penetrate the patient is enhanced by increases in the energy level of the photons. Increasing kVp produces moderate increases in photon energy levels, although mAs and kVp are independent of each other. The typical level of kVp is 120.

Changes to kVp also affect image contrast, noise, and patient radiation dose. Here, a 15% rule applies: by increasing the kVp by 15%, the mAs can be decreased by 50%, resulting in increased noise and a decreased radiation dose to the patient. Likewise, when decreasing the kVp by 15%, the mAs can be increased by 50%, resulting in less noise but higher patient radiation dose.

Z-Axis Resolution

Z-axis resolution is probably the single most important CT parameter for understanding the formation of the CT image. Z-axis resolution refers to the ability to accurately characterize the HU value within a given voxel and is directly related to slice thickness.

Z-axis resolution is often confused with spatial resolution. As discussed later, spatial resolution is a parameter that is purchased with the scanner, whereas z-axis resolution is under the direct control of the technologist who can select the slice thickness *prospectively*. With multi-slice CT, the image data can be *retrospectively* determined and multiple combinations of slice thickness and intervals can be created due to the larger detectors and narrower collimation, providing greater information that can be acquired in a single tube rotation.

Figure 13. Isocentering of wrist in x-, y-, and z-axes.

Isocentering

The use of **isocentering** directly impacts both image quality and amount of radiation dose. Filters, hardware, and software are designed so that structures at the focal point are scrutinized. Remember that the diameter of the **scan field of view** (SFOV) is vendor-dependent. The region of interest should be centered and not fall outside the SFOV **(Figure 13)**. Radiating too superior or too inferior from the center exposes the patient to more radiation than necessary and produces inferior image quality. Certain clinical conditions, for example, kyphotic or lordotic spine, may prevent appropriate centering. In these cases, the technologist should be sensitive to the abilities of the patient while attempting to obtain the best image possible.

Slice Thickness

Slice thickness is a critical factor — more so in single-slice than in multi-slice CT — in determining image quality and affects a number of other variables, including noise, resolution, and patient radiation dose, as well as coverage of the area of interest.

The first determinate of slice thickness is the anatomy being scanned and is specified in most imaging protocols. Although thicker slices provide faster coverage and require less scan time at a lower total radiation dose to the patient, thinner slices provide crisper images of complex anatomy with greater reliability of HU values by observing fewer partial volume artifacts, resulting in better quantitative analysis in the pixels.

Partial volume effect

The major reason thin slices improve the appearance of the image is the reduction of the **partial volume effect**, also called **partial volume averaging**. The understanding of volume averaging is imperative. Each CT voxel represents the attenuation properties of that specific volume of material. If that volume is comprised of a number of different substances, that is, when dissimilar objects occupy the same voxel, the resulting HU value becomes an *average* of properties of those substances. This can occur at the edges of organs or when small objects are present in the anatomic region, for example, tiny nodules along the margins of tumors or thin strands of tissue within masses. The inability to distinguish between these structures could ultimately lead to an incorrect diagnosis.

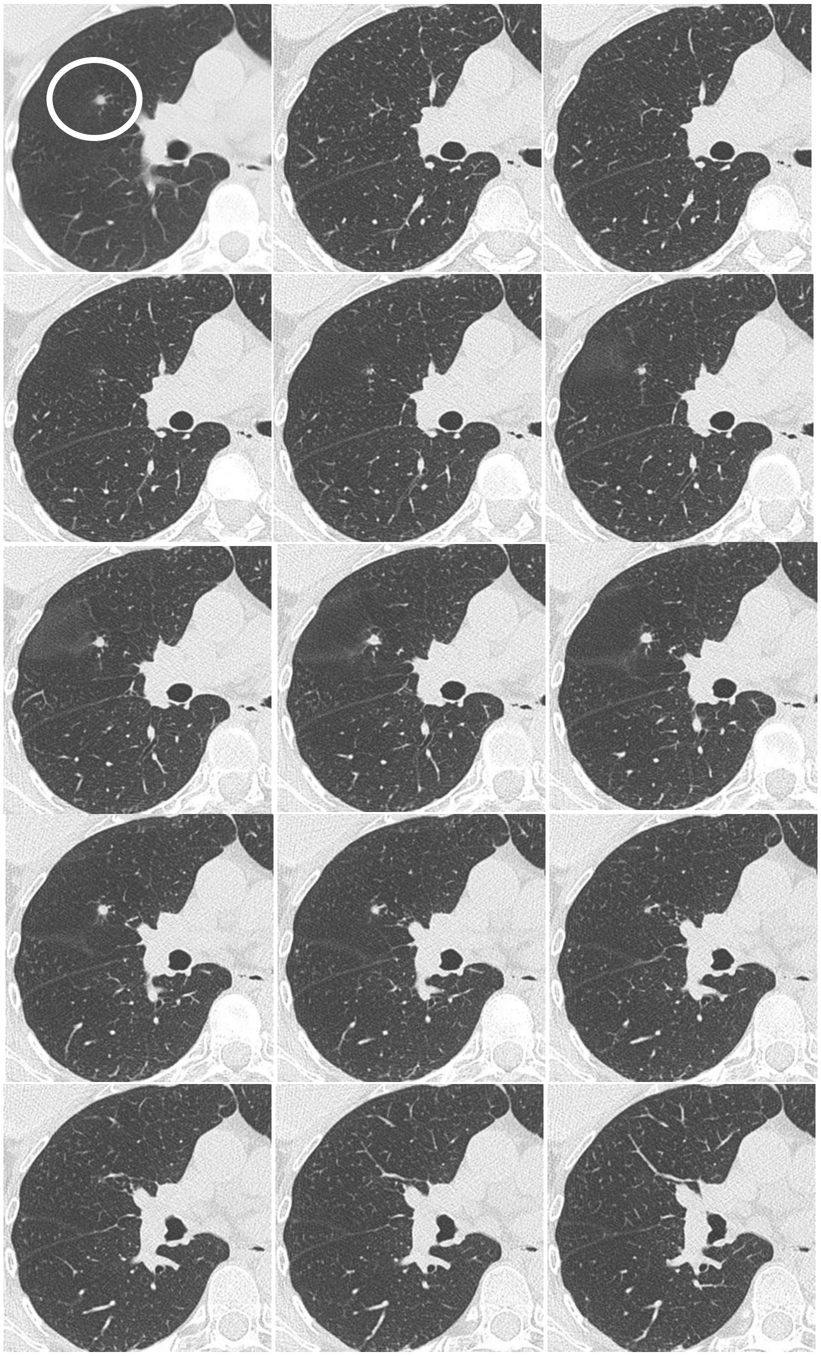


Figure 14. **Partial volume averaging of the chest. Using a 5x5mm reconstruction, the white opacity in the anterior aspect of the chest could be a vessel or a nodule. Adjusting the slice thickness and spacing reveal the area of concern is a nodule and not a vessel.**

Another example of volume averaging is when a small mass with a low HU value is surrounded by higher attenuation of the normal organ. This type of effect is seen when the mass is small and a higher HU value may appear in the voxels in which the mass is displayed, artificially raising the attenuation. For example, a cyst containing water (0HU) can be confused with a metastasis (100HU) only because the voxel is occupied by both HU values of 0 and 100. Because only one HU value number can be assigned to the final image within a voxel, the computer will assign this (and surrounding voxels) an HU value of 50HU (the average), thereby masking the true value of the lesion **(Figure 14)**.

Finally, if two objects with similar HU values are adjacent to one another, volume averaging may result in the total inability to see the pathological mass. Volume averaging is totally dependent on z-axis resolution controlled by slice thickness. However, thinner slices create somewhat noisier images because less radiation passes through the **collimator**. To compensate, the technologist must increase the mAS, which results in both increased tube heating and patient dose. Thorough understanding of the benefits and trade-offs of volume averaging is necessary for deciding on how to optimize the image.

Slice thickness with spiral/helical CT

In the conventional “step-and-shoot” mode, the z-axis thickness of the slice is equal to the collimator width. With spiral/helical motion single-source CT, however, the slices are actually not quite as sharp because of the moving patient under the rotating tube. Therefore a 3mm slice acquired by a spiral CT may have the same z-axis resolution as a 4mm slice if the patient is rapidly moving through the scanner, as might happen during a CT angiogram. This phenomenon is called **slice broadening** and is directly related to a measurement of system performance called pitch. In newer multi-slice scanners, this phenomenon is less of an issue because of the greater number of detectors and greater z-axis coverage, allowing for fewer slice-broadening issues.

Pitch

Pitch (P) refers to the distance the patient table travels in the time it takes the tube to complete one 360° rotation divided by the slice width.

Single-slice pitch calculation

Pitch can be calculated by this simple formula:

$$P = (\text{table travel/rotation}) \div \text{collimation of single slice}$$

Pitch is expressed as a ratio but because the second number in the ratio is always “1,” the pitch is most often referred to by the first digit. At a pitch ratio of 1:1 (pitch 1), the table moves in increments equal to the slice thickness during one tube rotation. If the slice thickness is 5mm and rotation time is 1 second, then the patient is moving 5mm/second. Therefore if the area to be covered is 20cm (200mm), it would require a 40-second breathhold. Increasing the pitch to 2:1 (pitch 2) means that the patient moves 10mm/1-second tube rotation. Here, it would require a 20-second breathhold to cover the same distance. Remember that at higher pitch, there is increased slice broadening, so the benefit of the thin slice may be sacrificed for speed of travel. The 2:1 pitch ratio covers the scan area in half the time but with lessened resolution as a result of the partial volume imaging effect. With a pitch ratio below 1:1, oversampling is likely to occur, with an overlapping of data. This improves image resolution but increases the patient’s radiation dose.

Multi-slice pitch calculation

The concept of pitch is more complex in multi-slice CT because the slice width can be, for example, four times that of the collimator. Nevertheless, the relationship is the same and is expressed as:

$$P = (\text{table feed/rotation}) \div \text{nominal slice width}$$

In this relationship, the term **nominal slice width** refers to the **thin element** within the detector array configuration. Therefore at a pitch of 6 and using a detector configuration of 4x5mm (nominal slice width = 5mm), the table feed per rotation will be 30mm. In this way, a 20-cm distance (200mm) can be scanned in about 6.5 seconds. Compare this speed to the 40-second breathhold required in single-slice. If an operator wished to cover the same length with a nominal slice width of 1mm (4x1mm detector configuration), the table feed would be 6mm and the time would be 33 seconds. Most multi-slice CT units scan with a rotation time of less than 1 second. In the same example, if the rotation time were 0.5 seconds, the total scan time would be 16 seconds.

An understanding of the pitch relationship helps us understand the power of multi-slice CT. For instance, in order to accomplish a vascular study of the lower extremities, the technologist must scan a length close to 1,000mm. Additionally, thin sections (≤2.5mm) are required to accurately display the aortic branches. Using a pitch of 6, a nominal slice of 2.5mm, and a 0.5 second rotation time, the table will move at 15mm/second for a total time of 33 seconds. This length of time allows for capture of the presence of iodinated contrast within the arterial branches of the lower extremities.

Patient Dose

As with all modalities that utilize x-rays, it is important to acquire the image with the lowest amount of radiation exposure to the patient. The main parameters used to control patient dose are the product of tube current (mA) and the exposure time in seconds (mAs). Patient dose can be decreased by shortening the scan time and/or decreasing the tube current. The technologist is challenged to determine the lowest possible dose at the shortest possible time to create an image that is diagnostically acceptable.

Another factor that must be considered in dose is x and y isocentering of patients in the center of the gantry. A patient who is above isocenter and closer to the tube will receive more mAs; if below isocenter, the patient will receive less mAs and be ―under" dosed, increasing the noise in the images. Optimal radiation dose is at isocenter.

The **CT Dose Index** (CTDI), the most commonly used parameter for estimating and minimizing patient radiation dose in CT, may vary across the field of view. For the body **phantom**, the CTDI is typically a factor of two greater at the surface than at the phantom center. Because image noise is decreased when radiation is increased, a CT image can appear fine. Without the visual cues of over-exposure that one sees in plain films, the operator must be vigilant about adjusting the scan parameters to avoid excessive radiation dose.[1]

KEY TERMS
phantom
contrast resolution
spatial resolution
temporal resolution
noise
linearity
window width
window level

IMAGE QUALITY

Image quality is of paramount importance and heavily influenced by the ability of the technologist to control and adjust several variables. The five major parameters that affect image quality are:

- **Contrast resolution**
- Spatial resolution
- **Temporal resolution**
- Noise
- **Linearity**

The ultimate quality of the final image can vary significantly depending upon how each of these variables is adjusted. It is important to realize that even a minor change of any one parameter may affect all of the others.

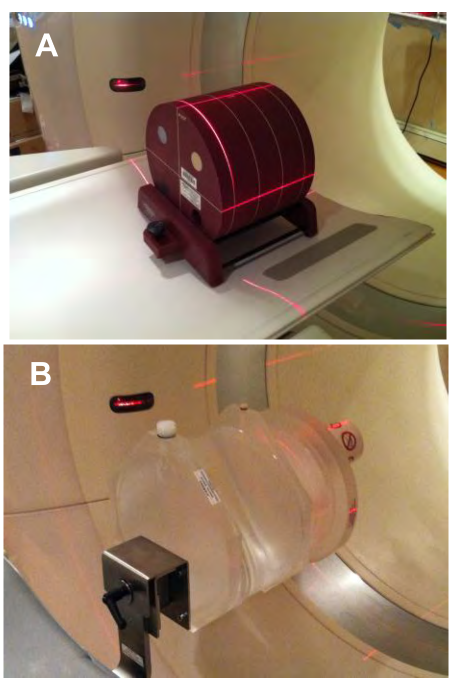


Figure 15. Phantoms. (A) ACR vendor-neutral phantom. (B) Vendor-specific phantom.

Phantoms

Phantoms are objects specifically designed to calibrate CT scanners for optimal performance. The phantom objects are generally created from plastic to provide simulations of the real objects to be scanned in terms of density and shape and are used to measure different performance parameters, such as contrast resolution, noise, slice thickness, and the production of artifacts. Since the HU value of everything in the phantom is known beforehand, scanning this image and comparing it to the value of water demonstrates exactly how closely the scanner is able to recreate a similar image **(Figure 15)**. Commonly used phantoms include a Plexiglas® phantom designed by the American Association of Physicists in Medicine (AAPM), as well as many provided by the manufacturers of CT units (for more information on quality assurance, see page 32).

Contrast Resolution

The beauty of CT imaging is that a wide range of contrasts can be displayed, far surpassing the detail of plain x-rays. Because of the use of digital images, CT can be optimized to recognize a difference as small as 4HU between tissues. Remember that the Hounsfield Scale was created to represent the attenuation of all known tissues. The scale represents densities ranging from -1000 (air) to +1,000 (bone) or more; steel has a HU value of approximately +4,000! Remember that, the critical element about the Hounsfield Scale is that it is centered on water. Water is assigned a value of zero (0) HU and is assigned a certain level of "grayness." This HU value also contains information regarding the type of tissue being evaluated. For example, a mass composed of pixels all measuring 0HU can be confidently diagnosed as a water-filled cyst. Unfortunately, computer monitors currently only recognize 256 shades of gray and the human eye can only distinguish about 20 shades. Therefore, for display, the *amount* of gray or white in a water pixel is fixed by the operator-chosen control called **window level**, and the *spread* of white to gray centered on the level-determined water pixel is controlled by the operator-chosen control called **window width**. HU values are assigned a value on the gray scale so the computer can come close to displaying the image generated by the raw data. It is critical for anyone operating a CT machine to appreciate that **windowing** or changing the level and/or window of a CT scan does *not* change the actual measured value of the HU within a pixel; the only change is the gray tone or color with which that pixel is displayed.

Window width

The size of the window width (WW) determines the range of HU the monitor will allow to be shown. The pixel values for pure white (+1000) and pure black (-1000) remain constant, but as the window width is increased or decreased, the values between them are redistributed. Greater window width allows for the display of a greater range of HU values, meaning there will be less contrast between similar tissues. The use of wide windows (1600HU or more) is generally reserved for viewing high-contrast tissues, such as the lungs. A narrow window width redistributes fewer HU values within the same range of pixel values, for greater contrast. Narrow window width (180-200HU) should be chosen when differentiating tissues of similar densities, like the liver **(Figures 16, 17, 18)**.

Window level

Window level (WL) designates the center HU value of the window width the monitor will display. For example, with a WW of 500, setting the WL at 0 will display the range of HU values from -250 to +250, with 0 in the middle. At this window level, HU values outside of the range will be compressed so that all values below -250 will appear black, and all values above +250 will appear white. Likewise, setting the WW to 500 and the WL at 250 will redistribute the HU values to recognize only numbers from 0 to +500 **(Figures 16, 17, 18)**.

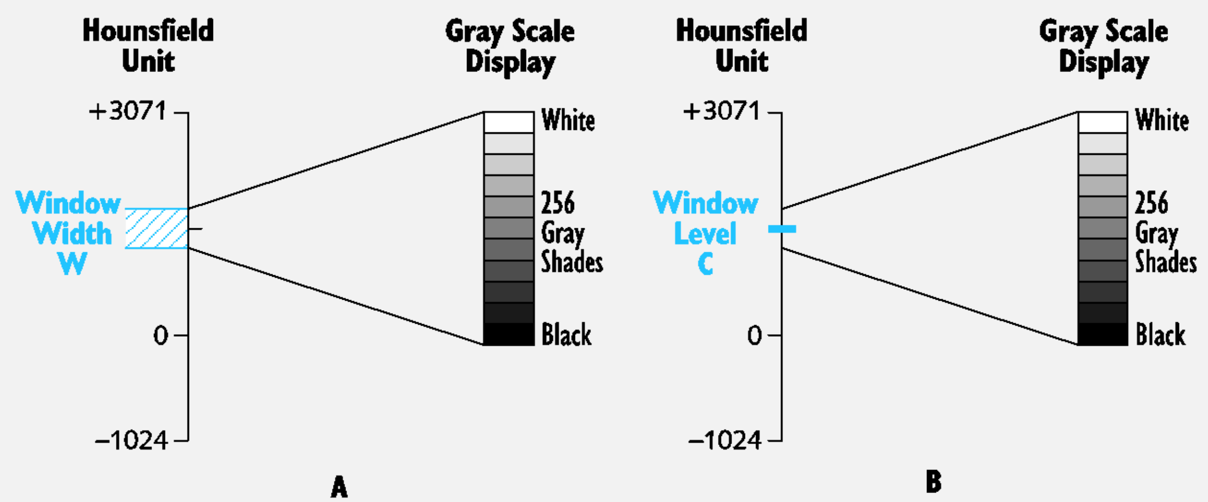


Figure 16. Windowing. The effects of (A) extending window width and (B) window level on CT image contrast parameters.

The technologist should select both the window width and window level to optimize the viewing of specific tissues. Although window level is generally set for the HU value of the object of interest, most CT scanners have preset WW / WL designations for specific types of CT studies (mediastinal, abdominal, lung, and brain) that can quickly display a WW / WL close to the optimized image from which the technologist can fine-tune the best display.

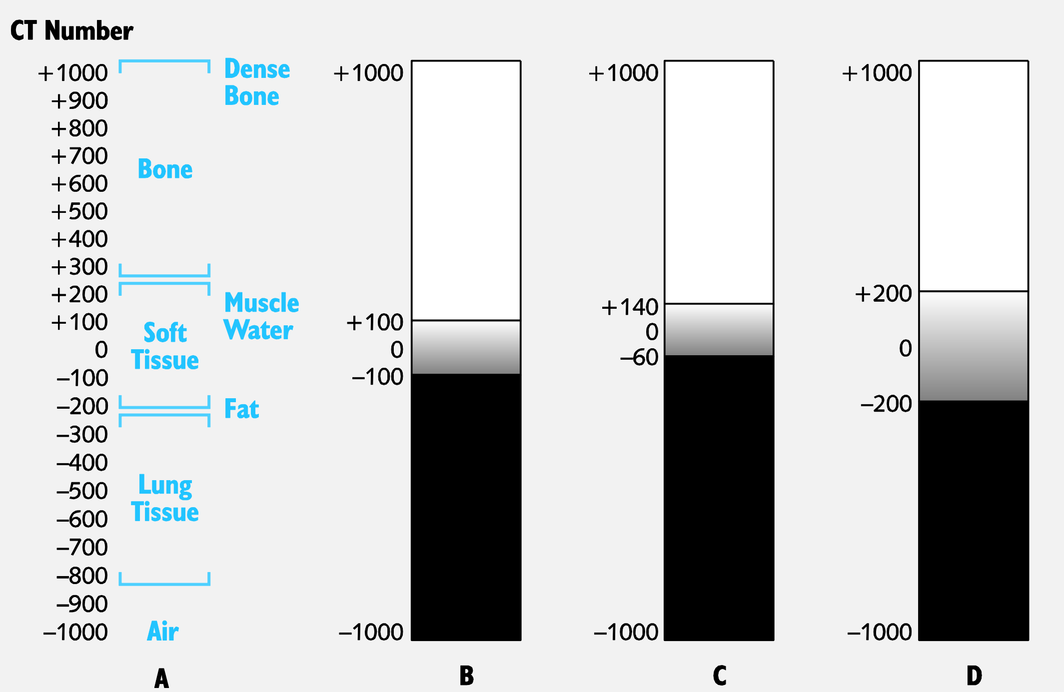


Figure 17. Differences in the range of HU values (CT numbers) recognized with the gray scale when window width and window level are altered.

Spatial Resolution

Spatial resolution defines how much detail is captured in an image and is dependent on the matrix size acquired. CT image resolution depends on the interrelationship of three factors: pixel, voxel, and matrix size. General resolution capability for high-contrast images is approximately 1.5 times the pixel size. Blurring of sharp edges in images is a limitation of CT scanning, resulting from several factors that negatively impact spatial resolution, including increases in pixel size and scanning of lower-contrast objects. Factors that directly affect image contrast will also indirectly impact spatial resolution. A good example of this is scatter radiation. Other factors may include focal spot size, aperture size, focal-spot-to-patient distance, patient-to-detector distance, and slice thickness. Newer CT scanners have an inherently higher spatial resolution in the x-y plane.

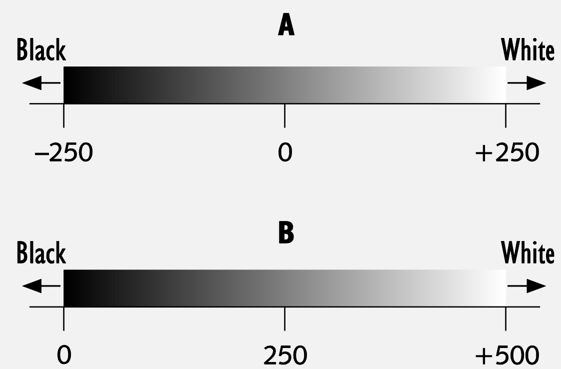


Figure 18. Comparison of the gray scale ranges when window width and window level are changed from WW = 500/WL = 0 to WW = 500/WL = 250.

The spatial resolution of a given system is often expressed as a mathematical formula known as the **modulation transfer function** (MTF), which is used to represent the **fidelity** of the image to the original object. An optimal degree of image fidelity would have an MTF value of 1, while a complete failure to produce an image earns a score of 0. Thus, most images have an MTF value between 0 and 1. The phantom most often used to test spatial resolution of a CT system is a line pair object in which a series of descending sizes of pairs of bars are scanned. Spatial resolution is reported in line pairs/mm (lp/mm). The smallest line pair that can be resolved determines the resolution of the scanner. The relationship of the bars to the space between them is called the **spatial frequency**. The smaller the object, the greater the spatial frequency needed for scanning. It should be remembered that the ability to perceive the object is influenced by the contrast between the object and the surrounding tissues. All CT imaging protocols are designed to maximize this contrast in the area of interest.

Temporal Resolution

Temporal resolution, or time resolution, is the scan time the system requires to acquire the data. The faster the scan acquisition in a rotation, the less motion in the images and the better the image appears. Temporal resolution is best observed with cardiac imaging where the heart is continuously in motion, and the time acquisition of the images needs to be as short as possible in a gantry rotation. Currently 64, 128, 256, and 320 multi-slice and dual-source CT offer the ability to acquire the data with minimal time resolution. These scanners offer multisegmental reconstruction and step-and-shoot scanning methods to acquire data in phases when the heart is at rest in the R to R interval **(Figure 19)**. To date, dual-source CT offers the highest temporal resolution at 75ms.

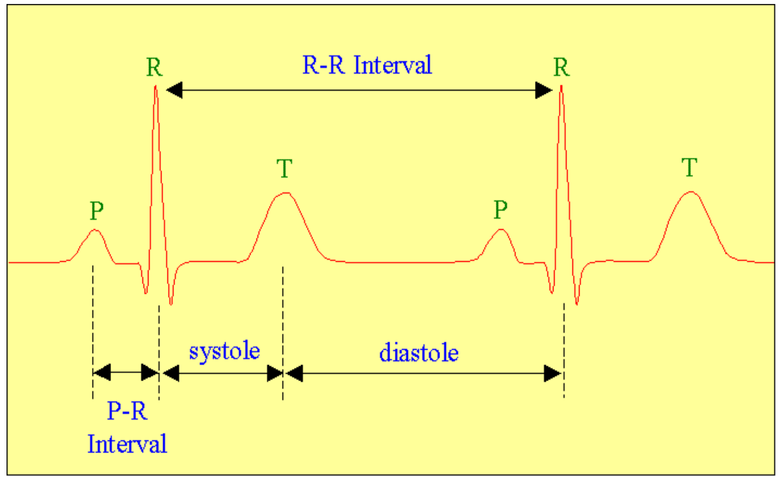


Figure 19. ECG QRS wave form, showing the R to R interval.
Courtesy of Dr. Ping He, Department of Biomedical, Industrial & Human Factors Engineering, Wright State University.

Noise

Noise significantly impairs resolution of low contrast objects, and a ―noisy" image appears grainy on the monitor **(Figure 20)**. Noise, or **quantum mottle**, represents deviations from uniformity of the image grid in which low-contrast HU values somewhat above or below 0 are interpreted by the computer as 0 values. The amount of noise in a given image is influenced by a number of factors, including detector size, spacing and efficiency, patient size and density, patient radiation dose, slice thickness, image matrix size, and reconstruction FOV and filter. Selecting the appropriate algorithm and patient radiation dose will assist in controlling noise.

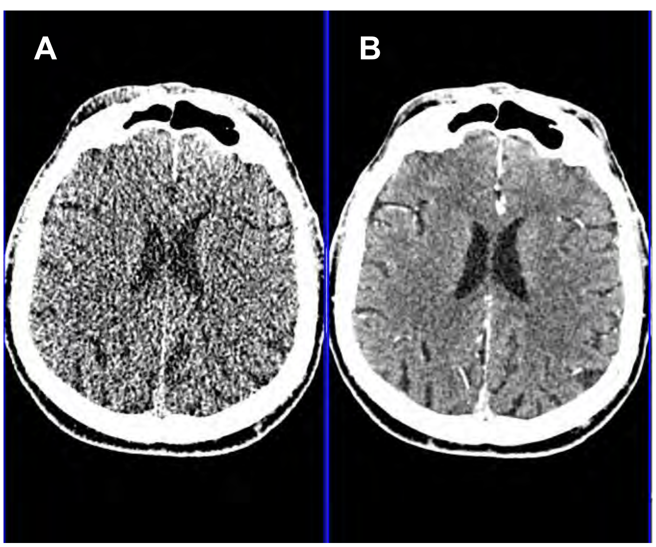


Figure 20. Axial brain. (A) With quantum mottle or noise. (B) Without noise.

Linearity

Linearity refers to the accuracy of the calibration of the scanner. The HU value of water must be 0 for all other values to be accurately recognized by the system. Daily calibration checks by the technologist are necessary to keep the CT system balanced and for images to be accurately depicted in reconstruction. This is accomplished by performing a test scan of a phantom, after which the values in the phantom object are recorded and the standard deviation is calculated and plotted. The plot line of the HU value and the linear attenuation coefficient should pass straight through 0 for the CT system to be calibrated perfectly. Any deviation from linearity would produce an inaccurate assignment of HU values to a given tissue. Smaller deviations would probably not affect the image display, but increasing deviations can seriously impair the resulting image.

IMAGE RECONSTRUCTION

Raw data contain information about the attenuation values of the beam mapped into the image matrix. The process of image reconstruction changes raw data into a readable image. Image data are filmed, interpreted, stored, and manipulated. Raw data are usually kept available on the system for at least a week so customized reconstructions can be performed if requested by the radiologist. Because of the significantly larger memory requirements (often in **gigabytes** [GB] and **terabytes** [TB]), it is not practical to permanently store raw data.

KEY TERMS
filtered back projection
filtering
reconstruction algorithms
retrospective reconstruction
postprocessing techniques
multiplanar reconstruction
3D imaging
maximum intensity projection
minimum intensity projection
volume rendering

Filtered Back Projection

Filtered back projection refers to a frequently used, uncomplicated method of image reconstruction. Back projection utilizes a basic numeric approach. Multiple ray beams are passed through an object, creating multiple **projections** which are then back-projected. The resulting images are calculated to create a single object image.

The projections are sent through correcting filters, which are computer programs designed to provide a more exact representation of the original object being reconstructed. Without a filter, the image will be severely distorted by the appearance of a star artifact, an effect of x-rays passing through some type of hardware in the patient's body, although the image will roughly approximate the original object (for more information on metal artifacts, see page 34). The main drawbacks to filtered back projection are a lack of a sharp image and the tendency toward other types of artifacts. Reconstruction algorithms and filters can compensate for the drawbacks of filtered back projection.

Reconstruction Algorithms

Reconstruction algorithms accentuate features within the image to optimize the reconstruction based on the anatomy. These algorithms work by weighting the contribution of surrounding pixels to the area of interest. The reconstruction algorithm is chosen prior to initiation of the reconstruction and is often incorporated into the imaging protocol. Several applications require the same data set be reconstructed multiple times because certain features of the anatomy are best shown with certain reconstruction algorithms. The most frequent example is in lung imaging where most radiologists prefer the ―standard” algorithm or soft tissue kernel for the mediastinum and the high frequency ―bone” or high frequency kernel for lung parenchyma evaluation. Filters may be high pass or low pass, depending upon the contrast required. The process of applying a reconstruction algorithm or filter is called **convolution** or filtering. In the early days of CT, application of the reconstruction algorithm required up to 30 seconds. Currently application of the reconstruction algorithm requires minimal time because of improved computer software and hardware.

High pass filters

High pass or high contrast filters are used when sharper resolution and better-defined edges to the objects in the image are desired. These filters work well with objects that naturally have a high level of contrast, such as studies of the musculoskeletal system and lung parenchyma, because the filters decrease the image contrast. However, high contrast filters are associated with an increased amount of noise attending the images. Examples of high pass filters are algorithms designed to display high-resolution images that are required for bone and lung imaging.

Low pass filters

Low pass or low contrast filters are associated with less noise than high pass filters, but they also decrease resolution. This type of filter provides enhanced contrast in all kinds of studies of low contrast soft tissues, such as the brain, abdominal organs and mediastinum.

Retrospective Reconstruction

Retrospective reconstructions are performed on raw data so that small regions of the anatomy can be optimally displayed. The reconstructions are performed on raw data sets; since raw data files are continuously being removed, the reconstructions are written into the imaging protocols so that the technologist performs the retrospective reconstructions immediately. It is possible to save raw data for the purposes of retrospective reconstruction, which can include reapplication of all of the algorithms, techniques, and filters that could have been applied at the time the projections were taken. As long as the raw data have been saved, new images can continue to be constructed. Retrospective reconstructions are most frequently employed in temporal bone imaging to look at the delicate structures of the inner ear.

Retrospective reconstructions can also be performed in the axial plane to improve the slice profile of the acquired images by overlapping the reconstructed slices. For example, using a single-slice CT scanner with a 5mm slice and a pitch of 2, the equivalent slice would have the z-axis resolution equivalent to a slice slightly greater than 6mm. By reconstructing the images with a slight overlap, the effective slice thickness can be reduced. Therefore a reconstruction interval may be used that is less than the slice width. It must be remembered that if a 5mm slice is reconstructed at a 3mm interval, the resultant slice is not a 3mm slice but instead two 5mm slices that have been overlapped.

Iterative Image Reconstruction

Another reconstruction process called **iterative image reconstruction** reduces radiation dose while providing and maintaining diagnostic image quality. Although the concept has been around for years, the call for reducing radiation dose has pushed the concept to the forefront, with more vendors incorporating this technology into CT software. Through a process of acquiring images and passing them through numerous software filters and noise-reducing calculations, the resultant image mirrors a full dose diagnostic scan **(Figure 21)**.

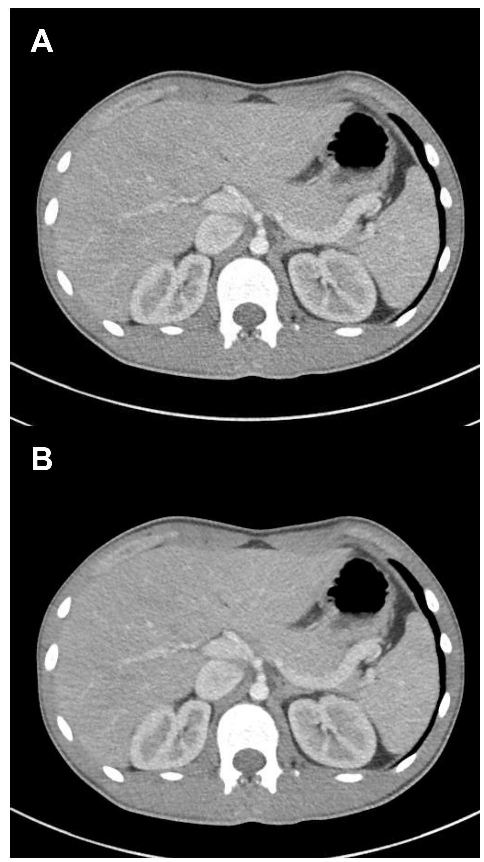


Figure 21. Axial image of the mid-liver. (A) Before iterative reconstruction. (B) After iterative reconstruction. Notice the noise in top image has been decreased on the bottom image.

POSTPROCESSING TECHNIQUES

Fusion Techniques

Fusion techniques are useful for overlapping preoperative CT with intraoperative fluoroscopic evaluation to better guide surgeons, especially during catheter guidance or ‑road mapping” **(Figure 22)**. After obtaining the preoperative CT scan, the images are converted on a workstation into a 3D format. Once the patient is in the operating room, the surgeon can call up the required images that contain the x- and y-coordinates necessary to view the exact patient orientation needed for catheter guidance. The ability to overlay CT data onto the fluoroscopic image not only assists the surgeon but also reduces radiation dose for the patient.

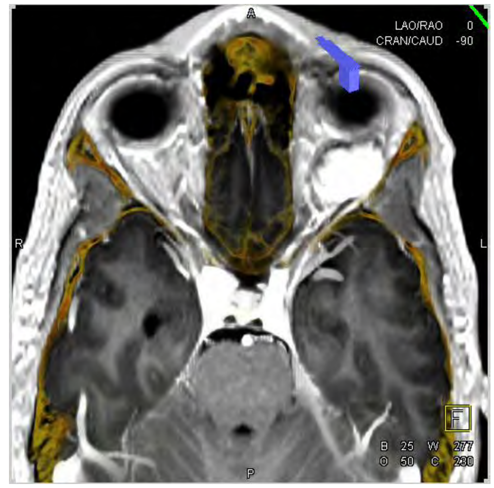


Figure 22. Fused preoperative MRI and CT scans. CT is in orange, representing bone; MRI is in gray scale representing soft tissue.

Multiplanar Reconstruction

Also called reformatting, 3D reformation, or MPR, this multiplanar reconstruction uses image data (as opposed to raw data) to create images of different orientations other than standard axial views. Traditional orientations include sagittal, coronal, or paraxial (**oblique**) views. Multiplanar reconstruction offers the advantage of creating new images without rescanning, which helps localize areas of interest and provide new information related to the relationship of a pathological process with surrounding structures. MPR requires voxels in which the dimensions in the x, y, and z planes are equal or isotropic. Multi-slice CT has been able to create such isotropic voxels because of the fixed distances between the detector rows. These displays are efficient ways of viewing large numbers of axial slices. Excellent MPR images also require that there be no gaps in the data set. Therefore, contiguous images are necessary.

Curved reformatting

Curved reformatting, a type of multiplanar reconstruction technique, results in the ‑unribboning” of tortuous vessels where they can be made straight, allowing more accurate measurements by the radiologist or surgeon **(Figure 23)**.

Figure 23. **Curved reformatting. (A) Coronal MPR of aorta. (B) Curved or "unribboned" MPR of aorta and iliac artery. (C) Volume-rendered image representing the path of the "unribboned" region of interest.**

2D/3D Imaging

Sophisticated computer techniques are available that facilitate looking at reconstructed data in different dimensions and in various planes. Today, these techniques include 2D imaging such as maximum intensity projection (MIP) and minimum intensity projection (MinIP), which are especially good for viewing vessels and airways, and 3D imaging such as volume rendering and surface shaded display, which are good for visualizing organs and bony anatomy, respectively **(Figure 24)**.

3D imaging is now common in the field of radiology. Although it has been around for more than 20 years, with the advent of multi-slice computed tomography 3D is now utilized in all types of CT scans. 3D differs from multiplanar reconstruction in that 3D applies *depth* to the image. Further, the difference between 3D and MIP/MinIP techniques is the *loss of the perspective of depth*. In maximum or minimum intensity, projections are based on the brightest or darkest pixel intensities, respectively. 3D techniques have great application in CT angiography.

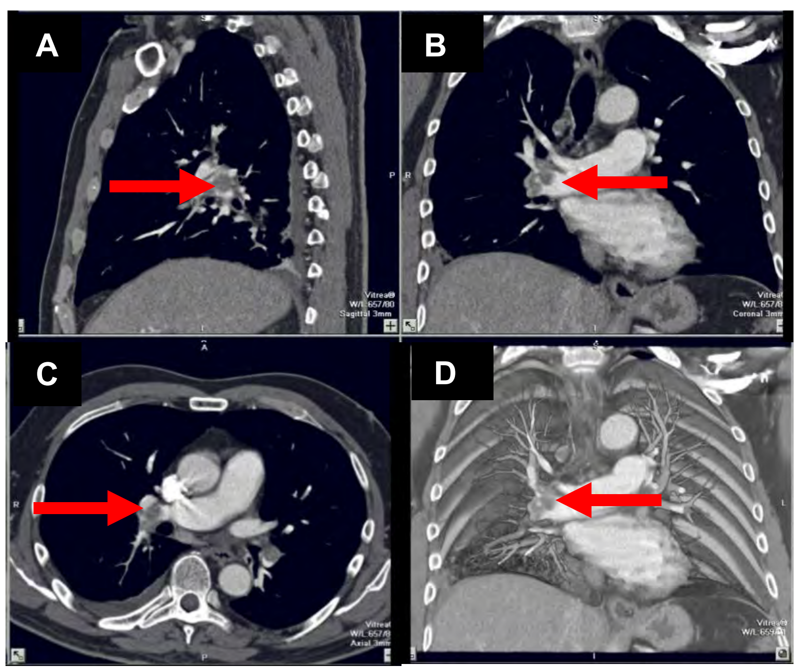


Figure 24. **MPR images. (A)Sagittal view, (B) Coronal plane or A/P, view. (C) Axial plane. (D) Volume-rendered image of chest shows the pulmonary artery. Note the pulmonary embolism (PE) on the right side and how easily it is visualized along with adjacent structures.**

Sophisticated volume-rendering techniques provide images that are indistinguishable from conventional CT reconstructions. Volume rendering may be the reconstruction technique of choice in the future to efficiently handle the large data sets being generated from current multi-slice scanners.

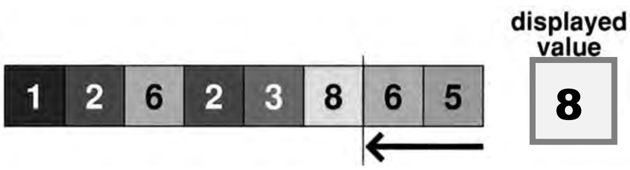


Figure 25. Illustration of MIP rendering. Courtesy of *Radiographics.*

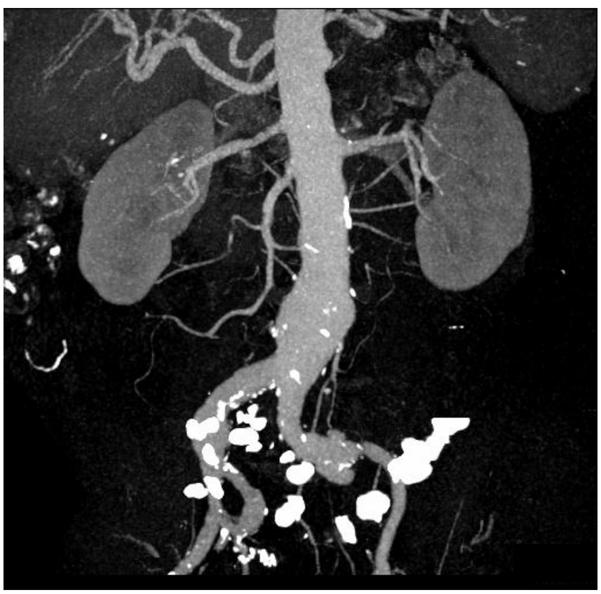


Figure 26. MIP rendering of the abdomen in A/P view. MIP allows the user to easily differentiate between blood vessels and dense calcifications. Note distinct difference in contrast between the vessel and the calcification on the vessel wall.

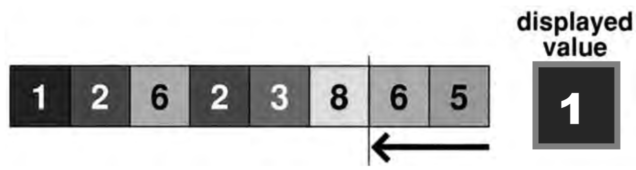


Figure 27. Illustration of MinIP rendering technique. Courtesy of *Radiographics.*

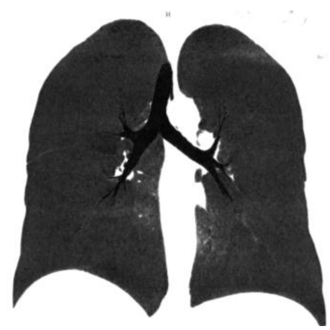


Figure 28. Coronal image of lungs. Small airways are more easily seen using MinIP when focusing on airway disease.

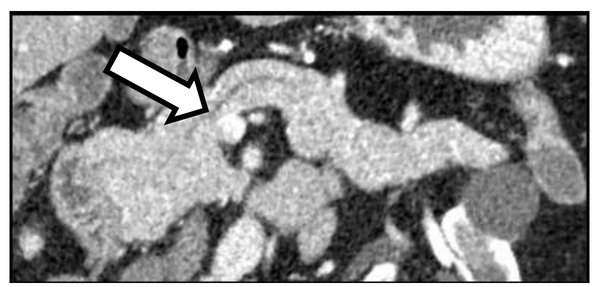


Figure 29. Curved MinIP image of the pancreatic duct. Notice low attenuation values are accentuated in the pancreatic duct, bringing out clarity where needed.

Maximum intensity projection

Maximum intensity projection (MIP) takes a 3D image and projects its highest attenuated voxels as a 2D object. Although this technique is often used in CT angiography, one must be cautious when using MIP as structures may appear superimposed due to loss of depth when in actuality there is space between the objects. The process of MIP evaluates the attenuation of each voxel, selecting the maximum voxel value based on its intensity. The maximum voxel value is then projected **(Figures 25, 26)**.

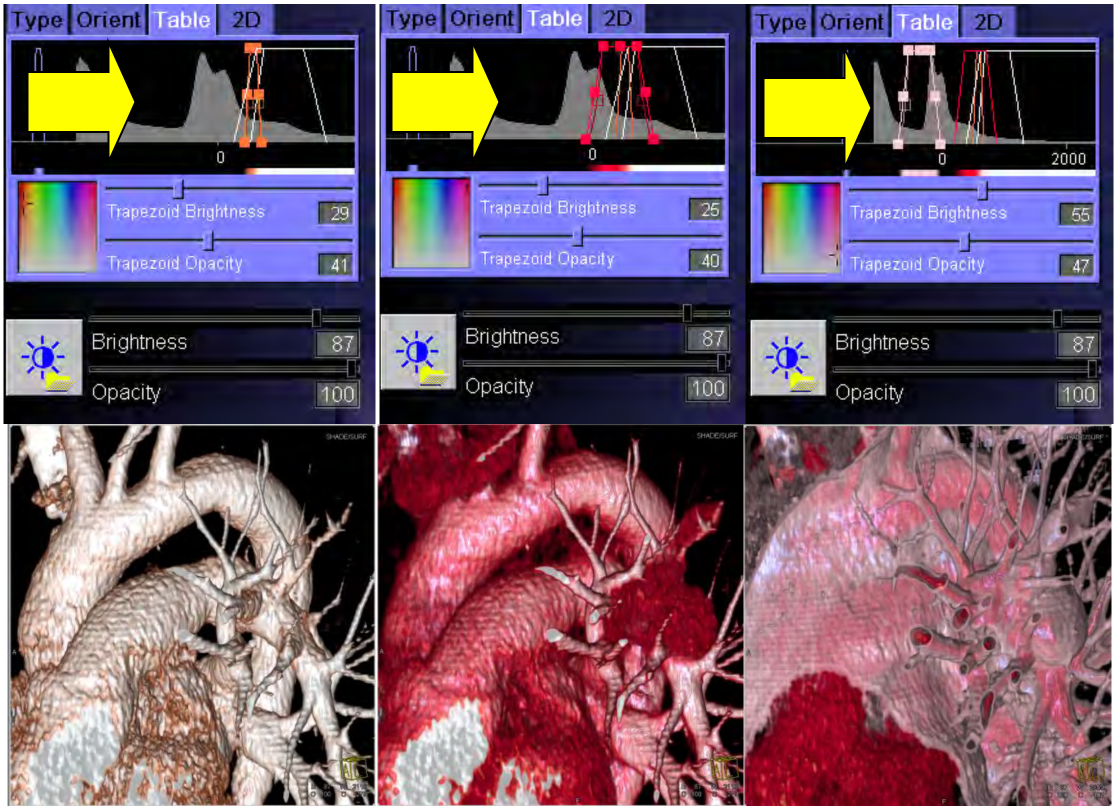


Figure 30. Trapezoids are the fundamental building blocks of creating volume-rendered images. The histogram shows how images can be generated by adding as many trapezoids as necessary to fully complement the different structures to be visualized.

(Left) Using just one trapezoid in a certain color will display vascular structures.

(Center) Adding another trapezoid allows setting the appropriate HU value to display additional anatomy of interest. This example shows soft tissue surrounding lung vasculature.

(Right) This example shows the wall of the vessels surrounding lung vasculature. It is very important to adjust the opacity and brightness display appropriately. If careful attention is not paid, it is possible to camouflage the area of interest.

Minimum intensity projection

In minimum intensity projection (MinIP), each voxel along a line from the viewer's eye through the volume of data is evaluated, and the minimum voxel value is selected on the basis of minimum intensity. The resulting displayed value of 1 corresponds to this minimum voxel value **(Figures 27, 28, 29)**.

Volume rendering

Volume rendering consists of image data that can be manipulated to display different opacities, lighting, shading, or transparencies. Underlying structures can be revealed by ―peeling back the layers of the onion.” Conversely, anatomy of interest can be unintentionally covered by applying too many trapezoids **(Figures 30, 31)**.

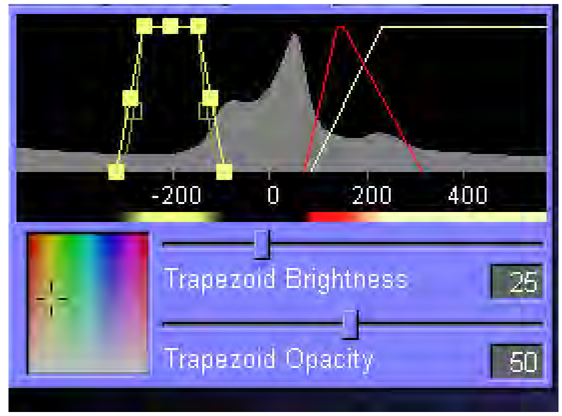


Figure 31. **Volume-rendered image with several trapezoids applied, revealing specific anatomy of interest.**

By selecting different trapezoids for brightness and opacity, human anatomy can be viewed as almost a "see-through" image, showing bony skeleton as well as arteries and transparent soft tissue.

Courtesy of NYU Langone Medical Center 3D

Surface shaded imaging

Surface shaded display (SSD) is a type of volume rendering where the surface of anatomy is visualized as opposed to underlying structures. While opacity and transparency cannot be manipulated, the benefits are faster processing time and thresholding of only the region of interest. SSD is especially useful in imaging of bone **(Figure 32)**.

Figure 32. **3D movie showing surface shaded display of craniofacial anatomy.**

CLICK HERE TO VIEW THE MOVIE AT YOUTUBE/ICPMEDUCATION
SSD Movie

~~I~~In general, surface-rendered images have the clearest volume depth cues of all 3D images. A common criticism of surface rendering is that the surface is derived from only a small percentage (less than 10% by some estimates) of the available data. In addition, surface rendering is not adequate for the visualization of structures that do not have naturally well-differentiated surfaces." [2]

4D Imaging

4D imaging has become a requirement in cardiac imaging for visualization of the heart in motion. The difference between 3D and 4D images are display of different time points. In other words, 4D is the adding the element of *time* to a 3D image, moving the display from a static to a dynamic visualization **(Figure 33)**.

There are many advantages to 4D over 3D imaging. 4D imaging:

- Allows estimation of the ejection fraction
- Demonstrates valves opening and closing, typically the mitral and tricuspid values, and visualizes the aortic root
- Allows ease of visualization while rotating the image as it travels through all phases of the heart; utilizing a **clip plane,** the motion of the valves is better visualized by “melting through” the entire data set at different angles of the anatomy
- Dynamically evaluates cycles of the heart
- Visualizes cardiac function and **morphology**
- Simulates flow of contrast media through vessels and organs

RECONSTRUCTION FACTORS

Reconstruction factors such as scan field of view (SFOV), display field of view (DFOV), and magnification are features of imaging that are applied after the collection of raw data is complete. These factors overlap with other features affecting image quality, such as image matrix, window size and level, tube current, and patient dose and can affect many aspects of image quality such as noise, contrast, and spatial resolution.

Figure 33. **4D movie of heart using volume-rendering technique.**

CLICK HERE TO VIEW THE MOVIE AT YOUTUBE/ICPMEDUCATION
[4D Cardiac Movie]

Scan Field of View

Quite simply, scan field of view or SFOV refers to the area of interest the scan field size is set to cover. The SFOV determines the number of detectors needed to collect data for a particular scan and must be larger than the area of interest. The entire patient anatomy should fall within the SFOV or out-of-field artifacts may occur, such as image shading, streaking, and misassignment of HU values. Any area outside of the SFOV will not be included in the data collected for reconstruction. Scanners often provide preselected settings (such as pediatric head, small head, medium head, or large head) for SFOV that include calibration algorithms specific to a particular part of the anatomy. Choosing a setting designed for a different area of the body, even if of the same size, will result in less-than-optimal image quality.

Reconstruction or Display Field of View

Reconstruction field of view, also called display field of view or DFOV, refers to the area of interest after it is reconstructed on the display. This representation of the original scan field of view can be zoomed in or out on a portion of the scanned area. A narrow DFOV displays a smaller area in a larger-than-scanned manner to magnify the area of interest. A wide DFOV zooms out to display the area of interest smaller than the original object in order to improve resolution and sense of perspective. Retrospective reconstructions can be performed to display both the original study and the ―zoomed" small DFOV images. Most modern scanners allow the minimal DFOV to be prescribed from the scout or topogram.

DFOV impacts both resolution and image noise. A wider DFOV increases the number of photons from data collected and reduces the amount of noise at the expense of resolution. Once again, these factors are also affected by other parameters, including focal spot size, scanner geometry, detector size, and the reconstruction algorithm used. As always, the goal is to strike a balance between noise and resolution. For images requiring a very high resolution (where a small DFOV is used), the accompanying noise can be somewhat compensated for by increasing the mAs. Scanners are often equipped with the ability to retrospectively produce more than one set of reconstruction images using the same raw data, allowing the technologist to use multiple combinations of DFOV and a reconstruction algorithm to achieve the optimal image.

Magnification

Magnification is the process of enlarging the scanned image for display. Selection of DFOV isolates the area to be scanned, after which the image can be magnified to improve viewing without affecting the spatial resolution or noise. For instance, after scanning the pelvis and hips, the right hip can be magnified post-scan to improve the viewing of this area. Magnification is a postprocessing technique that can make adjustments only to the reconstructed image. It has no impact on raw data and therefore cannot be used to improve the quality of the image. Magnification greater than two times the original image will generally result in a degradation of the viewed image. For these reasons, it is important to add magnification *after* all of the other factors in image formation and reconstruction have been adjusted for optimal quality and image sizing.

FILMING AND ARCHIVING

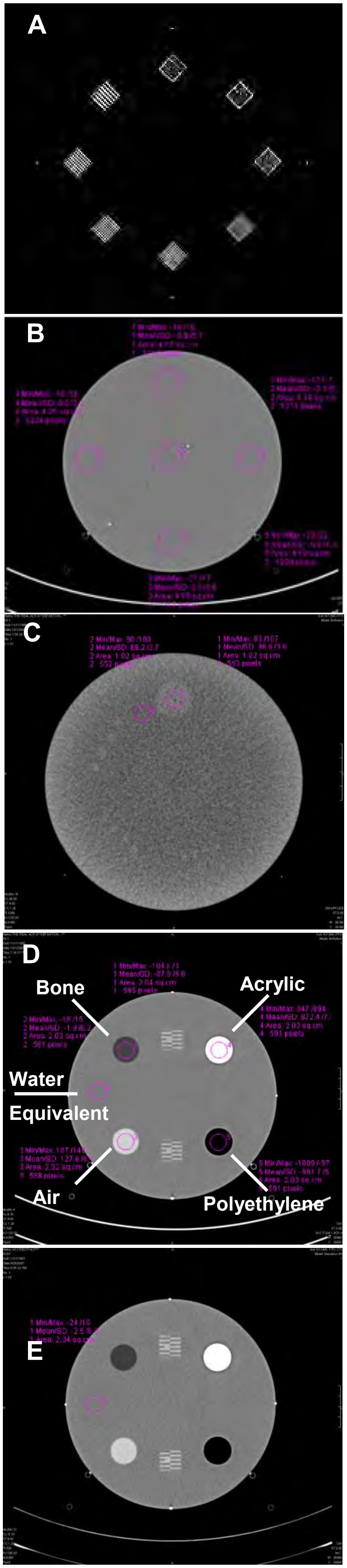
Image data can be permanently archived on a variety of media, including magnetic disk, CD-ROM, optical laser disk, DVD, and USB. In several states, CT data from hospitalized patients must be kept for a specified period of time (7 years in New York State). These public health mandates do not apply to outpatient facilities or imaging centers, although reputable operations will store their cases as well. In pediatric cases, the archiving period generally extends until the child has reached 18 years of age. Archiving CT data implies that the image sets are kept on storage media other than film. Simultaneously, diagnostic images need to be transferred to a medium that the radiologist can examine. Currently this involves storing the image on film, in a Picture Archiving and Communication System (PACS), or Radiology Information System (RIS).

PACS have the advantage of long-term storage without additional media. Films are created using a laser printer coupled with specialized single emulsion film, and the images are most commonly recorded as a 14” x 17” size format. A gray scale is provided along the film margin to help ensure that the contrast is correct on the film. It is critical that the film exactly matches what is on the screen at the display console as contrast and brightness levels of the film are determined by the laser camera — not at the CT console. Any discrepancy between the image on the monitor and that on the film signals the need for recalibration of the laser camera, which warrants a service call.

QUALITY ASSURANCE TESTS

Obviously, the execution of a quality image that offers appropriate detail is dependent on a complex interplay of a large number of variables, any of which can easily be misaligned, miscalculated, or simply malfunction. For these reasons, quality assurance testing is a routine part of CT scanning practice that should never be omitted or short-changed. The CT technologist, radiation safety officer, and system engineer are responsible for performing daily, weekly, and monthly maintenance checks to ensure optimal system performance

Each set of tests is designed to monitor a different aspect of CT scanner performance. Daily testing provides important information about the ―norms” of the system, allowing a diligent technologist to recognize patterns of drift from this performance over time. The tests are part of the training provided by the manufacturer of the equipment to ensure system stability. Additional testing should be performed on a weekly or monthly basis by a service engineer or site physicist to rigorously check total system performance. This process is referred to as preventative maintenance or PM.



Daily Tests

HU value calibration and standard deviation should be tested every day. A vendor-specific phantom is immersed in water and used to measure various performance parameters such as low- and high-contrast resolution and slice thickness display **(Figure 15)**. Uniformity – measuring average HU value at several different areas of the phantom including the phantom's center – is also tested. The values are then documented for internal and external reporting. Although water should be valued at absolute 0, a range of +/-3 HUs is acceptable. Anything outside of this range demands that a system recalibration program be performed and the test be repeated until it falls within acceptable limits. The ranges of acceptability for the standard deviation test depend upon the individual system and the scanning parameters for the image. Ranges outside of those the manufacturer deems acceptable suggest possible system failure at a number of junctures, including signal intensity, capture, or **amplification**, all of which will produce unacceptable levels of noise. Failure of the standard deviation test requires a service call for correction.

The American College of Radiology (ACR) also provides a vendor-neutral phantom to ensure the CT system meets minimum specifications and reproducibility and should be tested by the radiation safety officer and technologist on at least a semi-annual basis **(Figure 34)**. These tests measure performance on the basis of spatial resolution, contrast, table accuracy, dose, linearity, pixel consistency, density, radiation scatter, accuracy of slice thickness, and fidelity of the image. In addition, the technologist should keep a careful log of all system checks, results, CT workload, down time, and sample data for future reference.

Figure 34. ACR phantom. (A) This image represents the portion of a phantom used to test the spatial resolution/high contrast in line pairs. The smallest line pairs that can be resolved determine the resolution of the scanner.

(B) This image represents the portion of a phantom used to test uniformity.

(C) This image represents the portion of a phantom used to test low contrast detectability.

(D) This image represents the portion of a phantom used to test the HU value.

(E) Bar lines in the center represent slice thickness. This is used to test the accuracy of the reproducible slice thickness.

ARTIFACTS

KEY TERMS
aliasing artifact
metal artifact
motion artifact
beam hardening
out-of-field artifact
partial volume averaging artifact
ring artifact
tube arcing

One of the major limiting features of CT imagery is the preponderance of potential artifacts **(Table 2)**. These include a large array of disturbances in the display image that make it difficult to acquire the needed information. In the 1980s, C. Morgan defined an artifact as ~~a~~ distortion or error in an image that is unrelated to the subject being studied.” For CT, one can refine this definition as anything misread by the detectors that results in the assignment of an incorrect HU value. Artifacts are a by-product of many different functions of the CT system. Four predominant types of artifacts have been identified: streaks, shading, rings and bands, as well as miscellaneous patterns, for example, a basket weave. Artifacts can be caused by patient motion, difficulties with the imaging process, and equipment malfunctions. While identifying the source of artifacts is often difficult, identification is the key to correcting them.

Aliasing Artifacts

Aliasing artifacts are one of a number of streaking effects frequently seen. Objects with high spatial frequencies are best known to produce aliasing artifacts, which result from collection of too few data samples **(Figure 35)**. High-contrast CT images require a great deal of data collection, and sampling must occur at a rate of at least double the spatial frequency of the object being scanned. Third- and fourth-generation scanners have increased sampling capacity and take greater numbers of projections so are less likely to produce aliasing artifacts — although they may still occur when a partial scanning technique is used to reduce scan time.

Artifact	Type of Distortion		
	Streaks	Shading	Rings & Bands
Aliasing	●		
Beam hardening	●	●	
Bad detector channels			●
Incomplete projections		●	
Mechanical failure	●		
Metal	●		
Motion	●		
Noise	●	●	
Out-of-field radiation		●	
Partial volume averaging	●	●	
Ring			●
Scatter radiation		●	
Spiral/helical scanning	●	●	
Tube arcing	●		

Table 2. **Types of artifacts and the distortions they produce.**

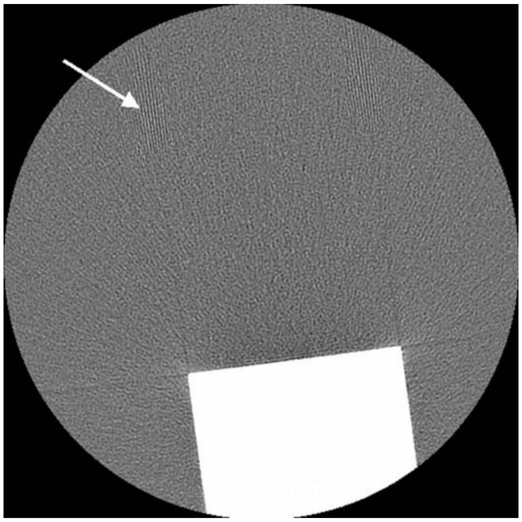


Figure 35. Aliasing artifact. CT image of a Teflon block in a water phantom shows aliasing due to undersampling of the edge of the block.

Courtesy of Julia F. Barrett and Nicholas Keat, St. George's University, UK

Figure 36. Star artifact. Tip of a feeding tube generates a dense, radically-oriented artifact that obscures significant anatomy.

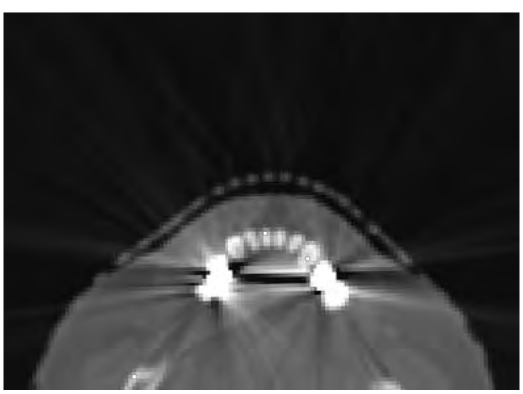


Figure 37. Star artifact. Presence of dental fillings creates a star artifact.

Courtesy of M. Garrett Larson, University of North Carolina Neuro Image Research and Analysis Laboratories.

Metal Artifacts

Metal artifacts are a common problem for all types of imaging modalities, including CT scanning, and produce a unique type of streaking called a “star artifact” **(Figures 36, 37)**. Metal artifacts are specifically caused by the presence of metal on or inside the patient during the scan in the form of jewelry, eyeglasses, metal in clothing, surgical clips, prosthetic devices, pacemakers, and dental fillings. The density of any metal is much greater than any tissue being scanned and therefore creates a radiating streak that is worsened by any table or patient movement. The obvious solution for external metal is removal of jewelry, clothing, and devices. Metal inside the body or teeth can be somewhat blocked by positioning the patient to keep the metal away from the scan field of view. Angling of the gantry may also help reduce the artifact from dental fillings. The scout view should be scrutinized for metal in belt buckles, pocket change, clasps, etc, and of course these items should be removed prior to the scan. If possible, metal wires from monitoring leads should be moved away from the range of anatomy being scanned.

Motion Artifacts

Patients may have difficulty lying still for CT scans, and even small motions can produce a type of streaking artifact known as “ghosting” in which the object of the scan appears doubled **(Figure 38)**. Motion artifacts may also produce a “blurring,” caused when the object of the scan gradually becomes displaced during the process.

Remember that patient motion may be voluntary or involuntary. Voluntary motion includes moving limbs or the head and generally occurs when the patient is anxious or does not understand the instructions. Carefully explaining the procedure to the patient beforehand and giving specific breathing instructions can minimize the occurrence of voluntary motion. It also may help to immobilize the patient or to use restraints.

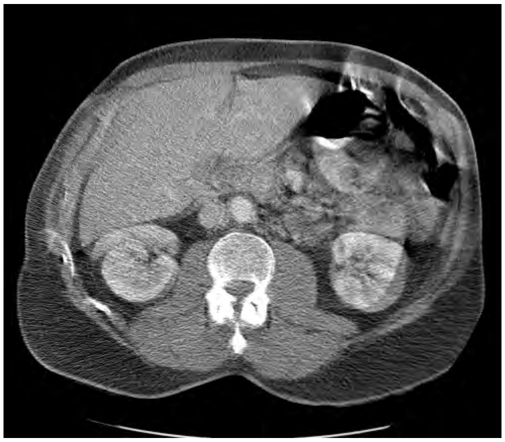


Figure 38. **Ghosting artifact. Image is degraded due to patient's respiratory motion. Kidneys appear as double structures. Motion also results in streaking from gas-filled bowel loops in the upper abdomen.**

Cardiac-related motion

Variations in ECG tracings can lead to step artifacts that look like ectopic or arrhythmic beats. When this occurs, images can have a step artifact that distorts part or all of the images. These variations can lead to additional or missed portions of the R to R intervals. Note the missing ECG data, causing a step artifact **(Figure 39)**.

Finally, helical scanning may be used to reduce scan time and therefore reduce motion artifacts. Naturally, involuntary motions such as breathing, heartbeats, hiccups, eye blinks, and swallowing should never be prevented, but they can be compensated for by selecting the fastest scan time possible to complete the exam.

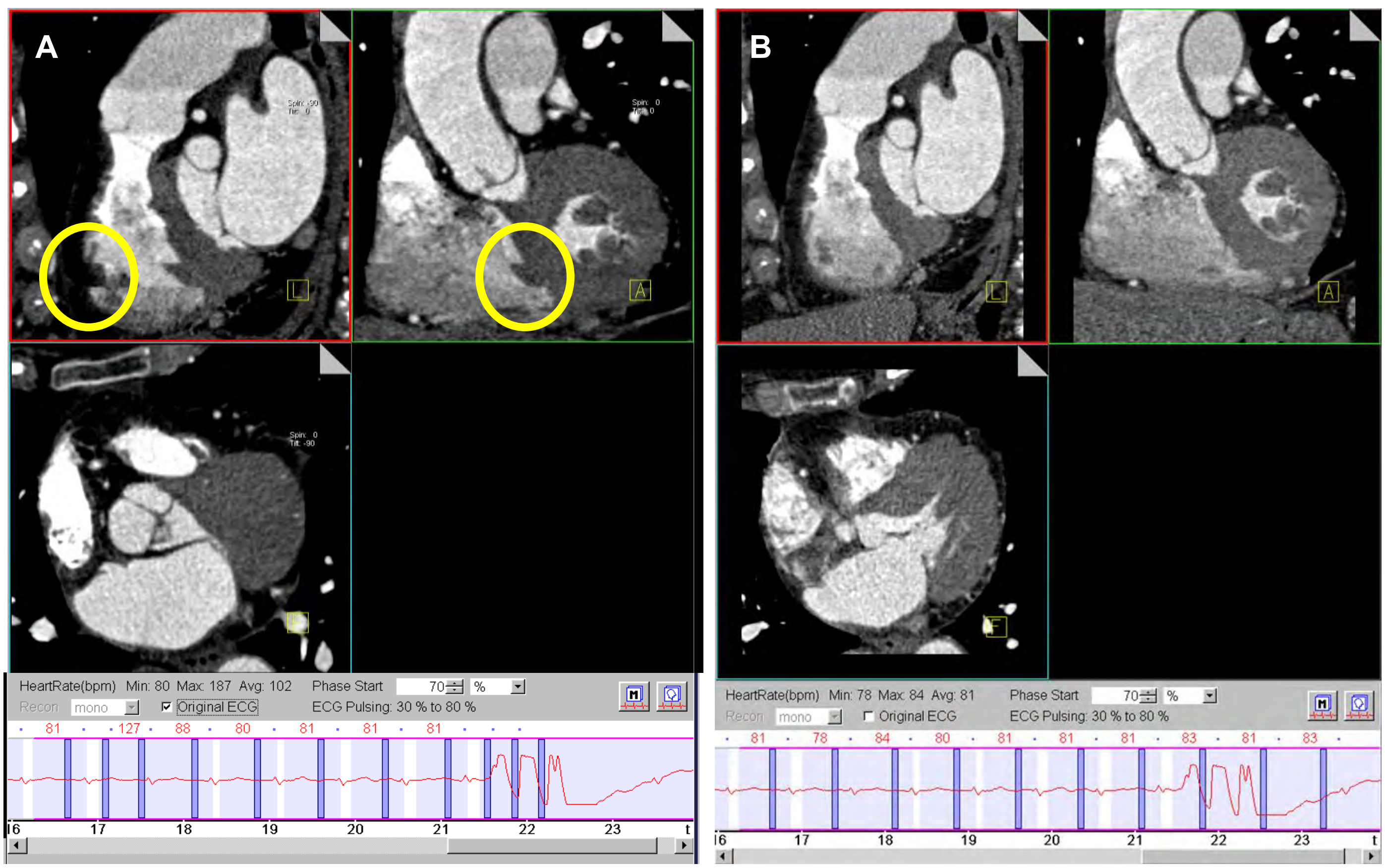


Figure 39. **Step artifact. (A) Original tracing. ECG triggering does not capture the heart rate correctly, resulting in misregistration in the signal and step artifact in the image. (B) Edited tracing. ECG triggering is corrected and synchronizations (time points along the ECG wave) are added to complete the gaps in the ECG signal, adding the missed data to give an artifact-free image.**

Courtesy of M. Barbara Srichai, MD, FAHA, FACC, NYU School of Medicine

Figure 40. **Beam hardening artifact. Severe streaking seen across the posterior fossa and temporal tips is due to beam hardening, resulting from the dense bone of the skull base.**

Beam Hardening

Beam hardening artifacts are produced when the mean energy of the x-ray beam increases as it passes through the patient. This occurs most often when a poly-energetic beam is used and the lower energy photons in the beam attenuate first, leaving the higher energy photons to pass through the patient and strike the detectors. Beam hardening artifacts can also occur when the path length of the beam varies as it passes through and around the object of interest — a longer path will further harden the beam. In either case, the hardening of the beam results in displacement of the HU value of the tissue and is more likely to appear on images of bone than on fat or less dense tissue **(Figure 40)**. The hardening is most pronounced at the center of the tissue and less so at the periphery, giving the artifact the appearance of a cup. Thus, the term ―cupping" is sometimes used to refer to this type of artifact **(Figure 41)**. Beam hardening may occur during scans of the temporal bones, the base of the skull, upper chest/shoulder area, hips, and with the patient's arms at the sides. CT scanners today generally come with special filters and algorithms designed to reduce the appearance of beam-hardening artifacts.

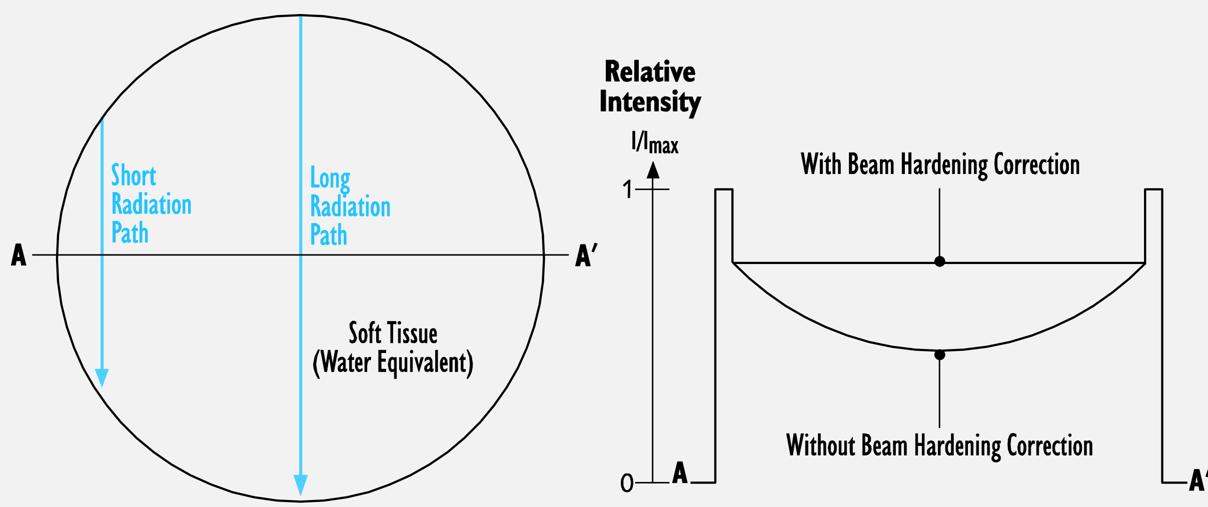


Figure 41. **Cupping occurs when the x-ray beam passing through an organ hardens the beam more in the center than at the periphery of the tissue. Beam hardening correction can help reduce this artifact.**

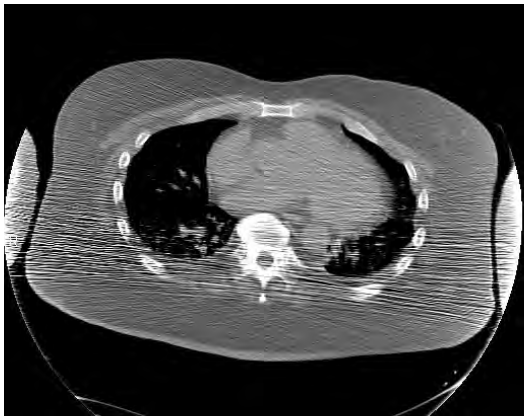


Figure 42. **Out-of-field artifact. Severe streaking across the chest is the result of partial occlusion of the arms within the SFOV. The patient had quadriplegia and therefore the arms could not be moved. Despite maximal SFOV, the entire anatomy could not be included because of obesity.**

Out-of-Field Artifacts

Out-of-field artifacts are commonly seen when attenuated x-ray beams from areas falling outside of the scan go unrecorded, although they still exert an effect. This type of artifact is caused by poor definition of the SFOV. The result is an inaccurate assignment of HU values to areas inside the scan **(Figure 42)**. Scanning of larger patients may present a challenge if part of the body does not fall within the SFOV. Obstruction of the detectors by parts of a patient's body may also cause streaks or shading artifacts. The extent of the artifact depends on the density of the tissue falling outside the SFOV. The best way for preventing out-of-field artifacts is to make sure the entire patient is within the scan field of view. For body scans, raising the patient's arms above the head so that the arms lie outside of the SFOV helps prevent out-of-field artifacts.

Partial Volume Averaging Artifact

Recall that partial volume averaging artifact is a direct result of the mathematic programming of the CT system, which assigns an HU value according to the *average* density of a scanned tissue without accounting for small differences within the tissue itself. Significant changes in density are notable; the presence of tumors or foreign bodies is detectable but very small areas of slight variation are hidden from the image. Partial volume artifacts can appear as a blurring of the margins between tissues with rounded edges or as a homogenized view of an area of different densities.

This particular artifact creates several problems for the radiologist who must make a diagnosis from the image as it may be difficult to identify normal changes in tissue from indicators of pathology. Additionally, volume averaging artifacts are sometimes seen in combination with other artifacts, resulting in what appears to be a real entity on the image. False-positive readings are possible, particularly when the slice is taken through the top or bottom of a tissue where rounding occurs, such as with the aortic arch or the dome of the liver. Partial volume averaging artifacts are often a function of obtaining thick slices, so reducing slice thickness will help reduce this artifact. Thinner slices are accompanied by more noise, and therefore mAs may need to be increased to compensate. Another solution is to acquire additional images from slightly different angles by repositioning the patient on the table or angling the gantry.

Figure 43. Ring artifact. Axial brain scan showing a dense ring super-imposed over the third ventricle in the mid-portion of the image.

Ring Artifacts

Ring artifacts are seen on images taken with third-generation scanners. The problem begins with a faulty or misaligned detector element that produces a ring effect emanating from the detector to the focal spot **(Figure 43)**. Occasionally, ring artifacts are produced by errors in back projection from fourth-generation scanners, but faulty detectors are more difficult to identify on fourth-generation scanners because the artifact may not be recognized as such. A single detector problem will not significantly alter the ultimate image but should still be serviced. The best way to identify a faulty detector on either a third- or fourth-generation scanner is to run an anterior-posterior or lateral localized image. If a ring artifact does appear, it helps the service engineer if the raw data set in which that artifact appears is saved so that the faulty detector can be quickly located. Any failure of a detector should be reported to a service engineer immediately so that it can be quickly corrected.

Tube Arcing

As scanner equipment ages, an artifact known as tube arcing is more likely to occur. Tube arcing is a function of high tube voltage and can produce severe artifacts similar to the streaking from metal artifacts, loss of image, and even total system shutdown. Tube-arcing artifacts can occur either inside or outside of the tube.

Arcing inside the tube

As x-ray tubes age, gas vapor emitted from the tungsten anode builds up. This build-up provides an environment for additional pathways to develop when the electron beam shoots down the tube, causing the beam to arc. As the level of tungsten gas increases over time, so does this phenomenon. Early signs of arcing inside the tube are random streaking and noise in images that increase in frequency and severity as the equipment continues in use. A service engineer usually performs a ―burn-off" to release the gases and allow the system to be recalibrated.

Arcing outside the tube

Arcing that occurs outside the tube is a more serious problem in which the oil surrounding the tube becomes impure. The tube of some CT units is set in a casing that contains **transformer** oil that helps dissipate the tremendous heat generated by the unit and conducts voltage to the anode. The unit collects gases and air bubbles that increase the conductivity of the oil, allowing the spontaneous creation of alternate high-voltage pathways that will divert the voltage from its usual course. This occurrence is known as "arcing in the oil" and creates a power surge so great that it usually shuts down the system. Although a cooling period may allow the system to scan again, it is recommended that the scanner be shut down temporarily until full service can be restored. An immediate service call is warranted.

SUMMARY

The ultimate quality of a CT image is determined by the interplay of six essential factors: contrast resolution, spatial resolution, temporal resolution, noise, linearity, and the presence of artifacts. These variables can be controlled but in order to do so, they must be thoroughly understood; by altering one parameter the others will be affected. Technologists must understand not only how radiation dose affects image noise, how changes in slice width affect dose, how dose affects tube heating and limitations of the scan, and how the choice of reconstruction algorithm affects the ability to diagnose certain diseases, but also how to execute the postprocessing requested by the radiologist. Approaching CT scanning equipped with these skills and knowledge will ultimately result in the production of a diagnostic image upon which patient recommendations can be confidently made.

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Aunt Minnie

CTisus

NetAnatomy

Journal of Radiology Case Reports

PubRad

RadioGraphics

Radiology

Radiology Case Reports

Radiolopolis

Tufts University School of Medicine Cross-Sectional Anatomy.

Visible Human Server

GLOSSARY

absorption
the taking up of energy by matter with which the radiation interacts.

algorithm
a mathematical process applied to raw data for the purposes of reconstruction, filtering, or enhancement of an image

amplification
increase in the strength of current or voltage

analog signal
a type of electrical signal that is directly measurable as voltage, light, x-rays, or number of rotations

analog-to-digital converter (ADC)
a device that converts analog signals into digital binary code that can be read and interpreted by the computer; its three major components are sampling, quantization, and coding

anode
the electrode at the end of an x-ray tube that, when struck by an electron beam, produces x-ray photons

aperture
the donut-shaped opening in the gantry through which the patient and patient table will pass during the scan

array processor
a device that reconstructs images from raw data at a very high rate of speed

artifact
an appearance on a radiographic image that is unrelated to the anatomical subject. Includes aliasing, beam hardening, edge-gradient effects, and metal and motion artifacts

attenuation
the reduction in the intensity of an x-ray beam as it passes through a patient

axial CT/conventional CT
a scanning method characterized by start-and-stop rotation of the gantry

back projection
a reconstruction process by which raw data are put into the image format (generally with filtering); also known as the **summation or linear supposition method**

beam hardening
an artifact produced when the mean energy of the x-ray passing through the patient increases, sometimes resulting in a cupping effect

bremsstrahlung
the term used for braking radiation, which comprises 80-90% of the x-rays used in CT scanning

cathode
the electrode in an x-ray tube from which an electron beam is initiated

cathode ray tube (CRT)
the monitor on which the image is displayed

central processing unit (CPU)
the ―brain" of the computer that directs and controls all of the other components

collimator
small electrical conductor that is used to shape the electron beam in the x-ray tube and reduce scatter radiation around the detectors

contrast
amount of differentiation between pixel shades of gray in a CT image

contrast resolution
the ability to differentiate density differences between tissues

convolution
the process of applying a reconstruction algorithm to an image; also known as **filtering**

clip plane
cutting away portions of the anatomy on one side of the plane

CT Dose Index (CTDI)
the most commonly used parameter for estimating and minimizing patient radiation dose

CT number or CT unit
the information contained in a single pixel assigned a value correlating to a shade of gray; also called **Hounsfield Unit**

data acquisition system (DAS)
the electronics positioned between the detectors and the computer responsible for collecting image data

detector
one of hundreds or thousands of identical receptor units that measure attenuated x-rays after they have passed through the patient

detector geometry
the relationship between the arrangement of the tube, the beam shape, and the detectors

digital signal
a type of electrical signal that is in the form of binary code that can be read by a computer - often called ―raw data”

digital-to-analog converter (DAC)
a device that converts digital signals from the scan controller into a continuous analog waveform signal so that the instructions from the scan controller can be interpreted; located between the scan controller and the gantry

display field of view (DFOV)
a reconstruction factor that refers to the area of interest after an image is reconstructed on the display; also called **reconstruction field of view**

dual-energy CT
an extension of dual-source CT; each of the two tubes has its own tube voltage, providing tissue differentiation by absorption of the energies

dual-source CT
CT unit utilizing two tubes and two detectors that rotate in the same plane and are at 90-degree angles from each other

dynode
a nonspecific type of electrode, including an anode or cathode

fidelity
a faithful reproduction that gives little to no distortion

filtered back projection
a frequently used, uncomplicated method of image reconstruction that uses a basic numeric approach. Multiple ray beams are passed through an object, creating multiple projections, which are then back-projected. The resulting images are calculated to create a single object image

filtering
the process of applying a reconstruction algorithm to an image; also known as **convolution**

gantry
the unit that houses the imaging components for x-ray production and data acquisition

gigabyte
one billion bytes

graphical user interface (GUI)
the combination of graphic images, eg, folders or icons, and the workflow

hertz (Hz)
the standard (SI) unit of frequency; equal to the old unit cycles per second; named for German physicist Heinrich Hertz (1857-1894)

high-voltage generator
the unit that generates the voltage necessary to produce x-ray photons

Hounsfield Unit (HU)
the information contained in a single pixel assigned a value correlating to a shade of gray; named for Sir Godfrey Hounsfield who shared the 1979 Nobel Prize for Physiology or Medicine for his part in developing the diagnostic technique of CT; also called **CT number** or **CT unit**

image matrix
the electronic plane of pixels that combine to form the image displayed on a monitor

interpolation
the process used to predict and calculate the image value between slices based on previously acquired data

isocentering
centering the region of interest so that it does not fall out of the scan field of view

isotropic
of equal physical properties along the x, y, and z axes, eg, a 0.5x0.5x0.5mm pixel

iterative reconstruction
the process of passing images through numerous software filters and noise-reducing calculations to reduce radiation dose while maintaining diagnostic image quality

kilovolts (kV)
unit of electromotive force, equal to 1000 volts

kVp
the maximum tube voltage from the cathode to the anode

lag (afterglow)
a type of artifact; can be produced when sodium iodide scintillation crystals are used

linear attenuation coefficient
the rate at which x-rays are diminished after passing through a patient

linearity
accuracy of a CT scanner, calibrated against the CT value of water

liquid crystal display (LCD)
device used to view CT images; it is of compact size, low power consumption, good color representation, excellent image resolution, and luminance (brightness); Resolution of monitors can vary from 1200x1600 pixels to 1536x2048 pixels depending on the type of LCD

localizer scan
a method of scanning that shows tissues superimposed without clear differentiation; also called a **scout**, **topogram**, or **reference image**

milliampere (mA)
the tube current that flows down the x-ray tube from the cathode filament to the anode

mAs
a scanner parameter used to reduce image noise; it refers to the tube current (mA) multiplied by the scan time.

modulation transfer function (MTF)
the frequency-dependent ratio of object contrast to image contrast. MTF permits qualitative determination of the spatial resolution of an imaging system.

mono-energetic beam
an x-ray beam that produces photons of the same energy

morphology
the study of the form and structure of organisms

noise
spots or blotches appearing on low-contrast images

nominal slice width
in multislice CT, the minimum slice thickness the detector elements can achieve; the collimation of detector elements configure various slice thicknesses, allowing flexibility in slice thickness choice

oblique
an angle that is not a multiple of 90 degrees

partial volume effect
when dissimilar objects occupy the same voxel, the resulting HU value becomes an average of the properties of those substances; also known as **partial volume averaging**

phantom
an artificial object, with a known CT value, that is weighted and configured for scanner testing of various parameters

photon
a particle of energy

pitch (P)
the distance the patient table travels in the time it takes the tube to complete one full 360° rotation divided by the slice width

pixel
a two-dimensional picture element that represents the smallest discrete block in a digital image display field; from ~~p~~icture element"

poly-energetic beam
an x-ray beam that produces photons of varying energies

projections
raw data resulting from conversion from analog to digital signals

quantum mottle
spots or blotches appearing on low-contrast images; also known as **noise**

random-access memory (RAM)
temporary storage of information in the CPU of a computer

quantizer
part of the analog-to-digital converter that converts large samples of data into smaller data samples

read-only memory
primary long-term data storage on the CPU of a computer

reconstruction field of view
a reconstruction factor that refers to the area of interest after an image is reconstructed on the display; also called **display field of view (DFOV)**

retrospective reconstruction
reconstruction of an image after raw data have been saved; includes multiplanar reconstruction and 3-D surface rendering

sample/hold (S/H)
the object being scanned is sampled to differentiate structures within the object and to assign various shades of gray to the pixels representing the structures

scan field of view (SFOV)
a reconstruction factor that refers to the area of interest the scan field size is set to cover

scintillation
the property of luminescence when excited by ionizing radiation; a scintillation crystal is used in the CT unit to help convert electrons into a digital signal for computer processing

scout
another term for a localizer scan; also called a **localizer scan**, **topogram,** or **reference image**

sensitivity
the number of light photons created by a rare earth material after the detector is struck by an x-ray

slice broadening
in spiral/helical CT, slices may not be as sharp due to the movement of the moving patient under the rotating tube and dependent on the pitch

slip ring
a large metal ring inside the gantry that rotates the scanning components and relays signals to and from the computer to perform the scan

spatial frequency
the number of occurrences between the visible line pairs and the gaps between them; the smaller the gaps between the lines, the better the spatial resolution

spatial resolution
defines how much detail is captured in an image and is dependent on the matrix size acquired

spiral/helical scanning
a scanning method in which certain x-ray tubes rotate continuously around a patient until the entire area of interest has been scanned; also called **volume data acquisition**

temporal resolution
the scan time the CT system requires to acquire the necessary data; also known as **time resolution**

terabyte
one trillion bytes or 1,000 gigabytes

thin element
refers to the physical detector that converts the xray signal to the electrical signal and also configures the slice and its thickness

transformer
amplifies the voltage to give the tube the necessary power it needs to generate x-rays

volume data acquisition
a scanning method in which certain x-ray tubes rotate continuously around a patient until the entire area of interest has been scanned; also called **spiral/helical scanning**

voxel
a three-dimensional volume of tissue that corresponds to an individual pixel on a CT image; from ―volume element"

windowing
a term describing computer manipulation of an image using the gray scale; includes both window level and window width

window level (WL)
the central point of the gray scale selected for the display window

window width (WW)
width of the display window given in Houndsfield Units

z-axis resolution
the ability to accurately characterize the CT number within a given voxel and related to slice thickness; is under the direct control of the operator

ABBREVIATIONS OF TERMS

AC	alternating current
ADC	analog-to-digital converter
AP	anterior/posterior
CBCT	cone beam CT
cm	centimeter
CPU	central processing unit
CRT	cathode ray tube
CT	computed tomography
CTDI	CT Dose Index
CVD	chemical vapor deposition
DAC	digital-to-analog converter
DAS	data acquisition system
DECT	dual-energy CT
DFOV	display field of view
DSCT	dual-source CT
EBCT	electron beam computed tomography
FOV	field of view
GB	gigabyte
GOS	gadolinium oxysulfide
GUI	graphical user interface
HU	Hounsfield Unit(s)
HVG	high-voltage generator
HVSR	high-voltage slip rings
Hz	hertz
IRS	image reconstruction system
kB	kilobytes
kV	kilovolts
kVp	kilovolt peak
LCD	liquid crystal display
lp/mm	line pairs/mm
LVSR	low-voltage slip rings
mA	milliampere
MB	megabyte
MinIP	minimum intensity projection
MIP	maximum intensity projection
mm	millimeter
MOD	magnetic optical disk
MPR	multiplanar reconstruction
MTF	modulation transfer function
P	pitch
PA	posterior/anterior
PACS	picture archiving and communication system
PC	personal computer
PM	preventive maintenance
RAM	random-access memory
RIS	radiology information system
ROM	read-only memory
RPM	rotations per minute
SFOV	scan field of view
S/H	sample/hold
SSD	surface shaded area
TB	terabyte
USB	universal serial bus drives
WORM	write-once-read-many
WL	window level
WW	window width