CT Instrumentation & Physics

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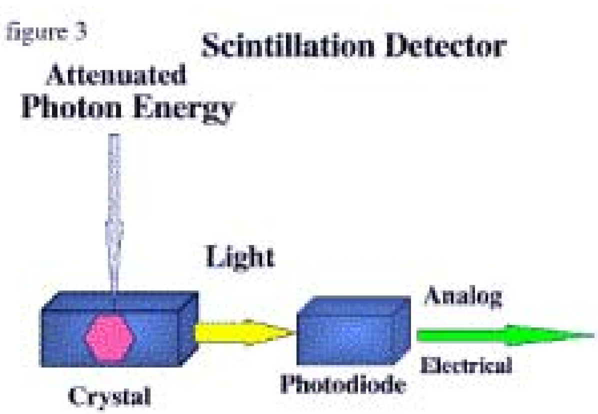
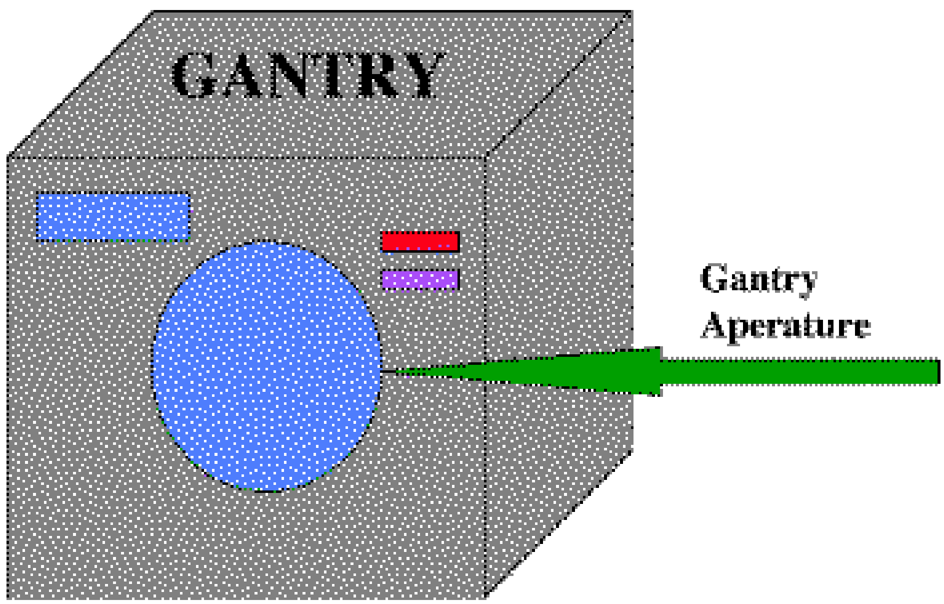


Computed tomography (CT) is the science that creates two-dimensional cross-sectional images from three-dimensional body structures. Computed tomography utilizes a mathematical technique called reconstruction to accomplish this task. It is important for any individual studying the CT science to recognize that CT is a mathematical process. In a basic sense, a CT image is the result of "breaking apart" a three-dimensional structure and mathematically putting it back together again and displaying it as a two-dimensional image on a television screen. The primary goal of any CT system is to accurately reproduce the internal structures of the body as two-dimensional cross-sectional images. This goal is accomplished by computed tomography's superior ability to overcome superimposition of structures and demonstrate slight differences in tissue contrast. It is important to realize that collecting many projections of an object and heavy filtration of the x-ray beam play important roles in CT image formation. Each component of a CT system plays a major role in the accurate formation of each CT image it produces.

CT Gantry

The first major component of a CT system is referred to as the scan or imaging system. The imaging system primarily includes the gantry and patient table or couch. The gantry is a moveable frame that contains the x-ray tube including collimators and filters, detectors, data acquisition system (DAS), rotational components including slip ring systems and all associated electronics such as gantry angulation motors and positioning laser lights. In older CT systems a small generator supplied power to the x-ray tube and the rotational components via cables for operation. This type of generator was mounted on the rotational component of the CT system and rotated with the x-ray tube. Some generators remain mounted inside the gantry wall. Some newer scanner designs utilize a generator that is located outside the gantry. Slip ring technology eliminated the need for cables and allows continuous rotation of the gantry components. The inclusion of slip ring technology into a CT system allows for continuous scanning without interference of cables. A CT gantry can be angled up to 30 degrees toward a forward or backward position. Gantry angulation is determined by the manufacturer and varies among CT systems. Gantry angulation allows the operator to align pertinent anatomy with the scanning plane. The opening through which a patient passes is referred to as the gantry aperture. Gantry aperture diameters generally range from 50-85 cm. Generally, larger gantry aperture diameters, 70-85 cm, are necessary for CT departments that do a large volume of biopsy procedures. The larger gantry aperture allows for easier manipulation of biopsy equipment and reduces the risk of injury when scanning the patient and the placement of the biopsy needle simultaneously. The diameter of the gantry aperture is different for the diameter of the scanning circle or scan field of view. If a CT system has a gantry aperture of 70 cm diameter it does not mean that you can acquire patient data utilizing a 70 cm diameter. Generally, the scanning diameter in which patient or projection data is acquired is less than the size of the gantry aperture. Lasers or high intensity lights are included within or mounted on the gantry. The lasers or high intensity lights serve as anatomical positioning guides that reference the center of the axial, coronal, and sagittal planes.

Figure 1



X-ray Tube, Collimation, Filtration

CT procedures facilitate the use of large exposure factors, (high mA and KvP values) and short exposure times. The development of spiral/helical CT allows continuous scanning while the patient table or couch moves through the gantry aperture. A typical spiral/helical CT scan of the abdomen may require the continuous production of x-rays for a 30 to 40 second period. The stress caused by the constant build up of heat can lead to a rapid decrease of tube life. When an x-ray tube reaches a maximum heat value it simply will not operate until it cools down to an acceptable level. CT systems produce x-radiation continuously or in short millisecond bursts or pulses at high mA and KvP.

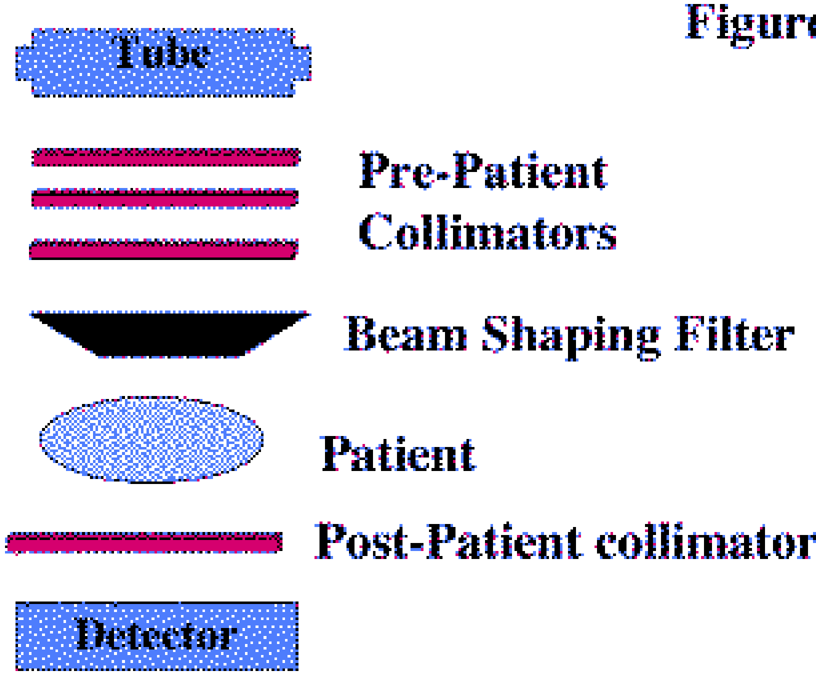
values. CT x-ray tubes must possess a high heat capacity which is the amount of heat that a tube can store without operational damage to the tube. The x-ray tube must be designed to absorb high heat levels generated from the high speed rotation of the anode and the bombardment of electrons upon the anode surface. An x-ray tubes heat capacity is expressed in heat units. Modern CT systems utilize x-ray tubes that have a heat capacity of approximately 3.5 to 5 million heat units(MHU). A CT x-ray tube must possess a high heat dissipation rate. Many CT x-ray tubes utilize a combination of oil and air cooling systems to eliminate heat and maintain continuous operational capabilities. A CT x-ray tube anode has a large diameter with a graphite backing. The large diameter backed with graphite allows the anode to absorb and dissipate large amounts of heat.

The focal spot size of an x-ray tube is determined by the size of the filament and cathode which is determined by the manufacturer. Most x-ray tubes have more than one focal spot size. The use of a small focal spot increases detail but it concentrates heat onto a smaller portion of the anode therefore, more heat is generated. As previously described, when heat is building up faster than the tube can dissipate it the x-ray tube will not produce x-rays until it has sufficiently cooled. CT tubes utilize a bigger filament than conventional radiography x-ray tubes. The use of a bigger filament increases the size of the effective focal spot. Decreasing the anode or target angle decreases the size of the effective focal spot. Generally, the anode angle of a conventional radiography tube is between 12 and 17 degrees. CT tubes employ a target angle approximately between 7 and 10 degrees. The decreased anode or target angle also helps eleviate some of the effects caused by the heel effect . CT can compensate any loss of resolution due the use of larger focal spot sizes by employing resolution enhancement algorithms such as bone or sharp algorithms, targeting techniques, and decreasing section thickness.

In CT collimation of the x-ray beam includes tube collimators, a set of pre-patient collimators and post-patient or pre-detector collimators . Some CT systems utilize this type of collimation system while other do not. The tube or source collimators are located in the x-ray tube and determine the section thickness that will be utilized for a particular CT scanning procedure. When the CT technologist selects a section thickness he or she is determining tube collimation by narrowing or widening the beam. A second set of collimators located directly below the tube collimators maintain the width of the beam as it travels toward the patient. A final set of collimators called post-patient or pre-detector collimators are located below the patient and above the detector. The primary responsibilities of this set of collimators are to insure proper beam width at the detector and reduce the number of scattered photons that may enter a detector.

There are two types of filtration utilized in CT. Mathematical filters such as bone or soft tissue algorithms are included into the CT reconstruction process to enhance resolution of a particular anatomical region of interest. Inherent tube filtration and filters made of aluminum or Teflon are utilized in CT to shape the beam intensity by filtering out low energy photons that contribute to the production of scatter. Special filters called "bow-tie" filters absorb low energy photons before reaching the patient. X-ray beams are polychromatic in nature which means an x-ray beam contains photons of many different energies. Ideally, the x-ray beam should be monochromatic or composed of photons having the same energy. Heavy filtration of the x-ray beam results in a more uniform beam. The more uniform the beam, the more accurate the attenuation values or CT numbers are for the scanned anatomical region.

Figure 2



Detectors

When the x-ray beam travels through the patient, it is attenuated by the anatomical structures it passes through. In conventional radiography we utilize a film-screen system as the primary image receptor to collect the attenuated information. The image receptors that are utilized in CT are referred to as detectors. The CT process essentially relies on collecting attenuated photon energy and converting it to an electrical signal, which will then be converted to a digital signal for computer reconstruction. A detector is a crystal or ionizing gas that when struck by an x-ray photon produces light or electrical energy. The two types of detectors utilized in CT systems are scintillation or solid state and xenon gas detectors. Scintillation detectors utilize a crystal that fluoresces when struck by an x-ray photon which produces light energy. A photodiode is attached to the scintillation portion of the detector. The photodiode transforms the light energy into electrical or analog energy. The strength of the detector signal is proportional to the number of attenuated photons that are successfully converted to light energy and then to an electrical or analog signal. The most frequently used scintillation crystals are made of Bismuth Germinate (Bi4Ge3012) and Cadmium Tungstate (CdWO4). Earlier designs utilized Sodium and Cesium Iodide as the light producing agent. One of the problems associated with these element was that at times it would fluoresce more than necessary. The after glow problems associated with Sodium and Cesium Iodide altered the strength of the detector signal which could cause inaccuracies during computer reconstruction.

figure 3

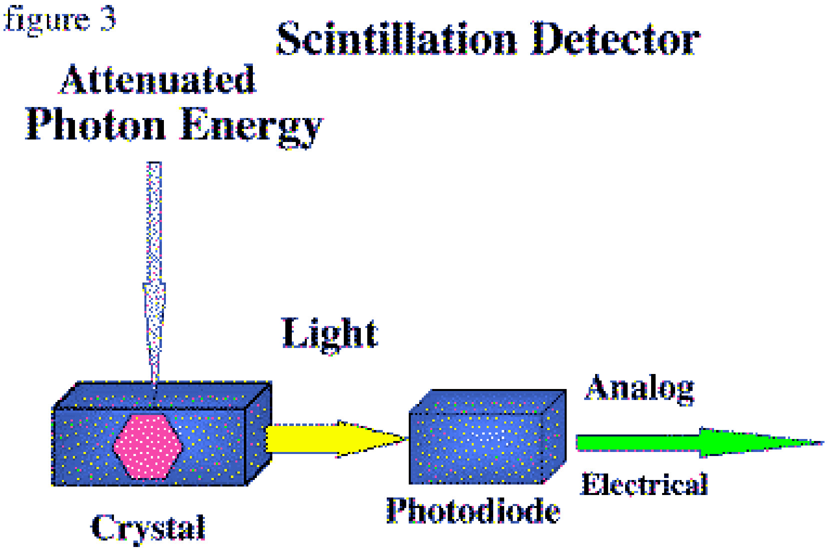
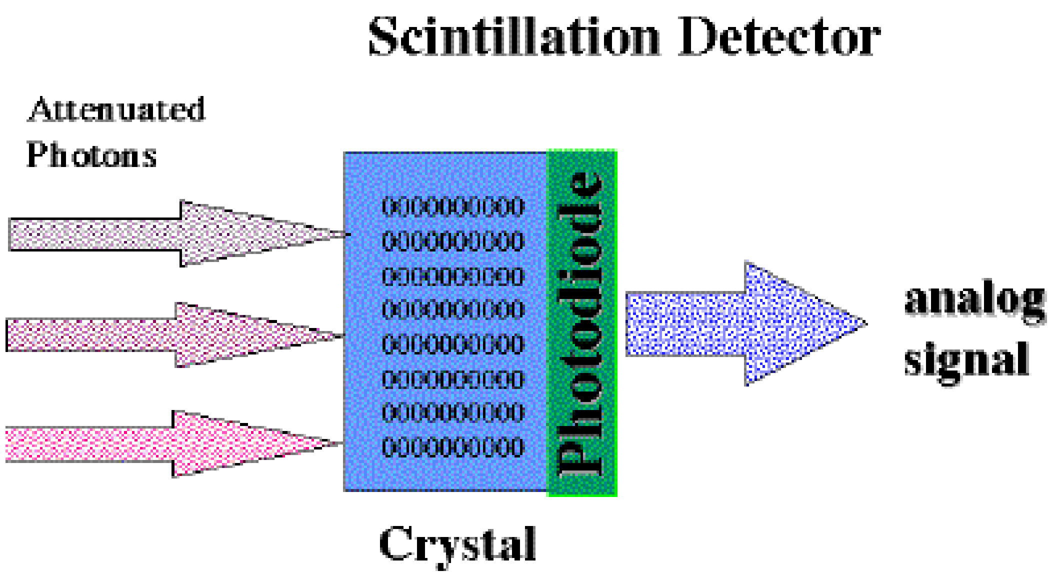


Figure 4

Cross Section



The second type of detector utilized for CT imaging system is a gas detector. The gas detector is usually constructed utilizing a chamber made of a ceramic material with long thin ionization plates usually made of Tungsten submersed in Xenon gas. The long thin tungsten plates act as electron collection plates. When attenuated photons interact with the charged plates and the xenon gas ionization occurs. The ionization of ions produces an electrical current. Xenon gas is the element of choice because of it's ability to remain stable under extreme amounts of pressure. Utilizing more gas in a detector increases the number of molecules that can be ionized therefore, the strength of the detector signal or response is increased. The long thin tungsten plates of the gas detector are highly directional. Ionization of the plates and the resultant detector signal rely on attenuated photons entering the chamber and ionizing the gas. If the xenon gas detectors are not positioned properly there is a chance that the ability of the detector to produce an accurate signal is compromised because the photons may miss the chamber. The xenon gas detectors are generally fixed with the position of the x-ray tube which occurs with 3rd generation scanner geometry designs.

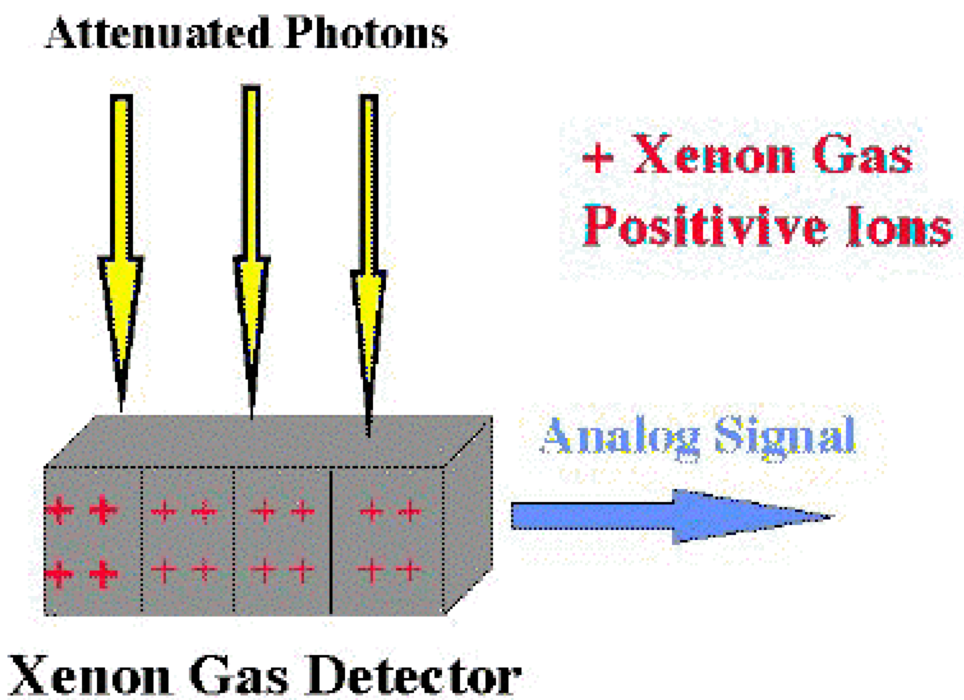


Figure 5

The term detector refers to a single element or a single type of detector used in a CT system. The term detector array is used to describe the total number of detectors that a CT system utilizes for collecting attenuated information. 3 rd generation CT imaging systems employ 800-1000 detectors while 4th generation scanners include 4000-5000 individual detectors in a detector array.

Overview

The path that an x-ray beam travels from the tube to a single detector is referred to as a ray. After the x-ray beam passes through the object being scanned, the detector samples the beams intensity. The detector reads each ray and measures the resultant beam attenuation. The attenuation measurement of each ray is termed a ray sum. A complete set of ray sums is referred to as a view or projection. It takes many views to create a computed tomography image. Obtaining a single view does not give the entire perspective of the object being scanned. Therefore, we can say that the detector is "seeing" an insufficient amount of information. The attenuation properties of each ray sum are accounted for and correlated with the position of each ray. At this point, the detector has "collected" the projection or raw data. The more photons collected, the stronger and more accurate the detector signal. This is essential for accurate image reconstruction. The detector accomplishes this task by adding together all the photon energy it has received. The detector receives all the projection data and subsequently generates an electrical or analog signal. The signal represents an absorption or attenuation profile. An attenuation profile is obtained for each view or projection. Every detector in the detector array is responsible for this task.

Detector efficiency describes the percent of incoming photons that a detector converts to a useable electrical signal. The two primary factors that determine how well a detector can capture photons relative to efficiency is the width and the distance between each detector. It is important that detectors are placed as close to one another as possible. Scintillation detectors convert 99-100 percent of the attenuated photons into a useable electrical signal. Xenon gas detectors are less efficient, converting 60-90 percent of the photons that enter the chambers. The efficiency of the xenon gas detector is compromised by the absorption of some of the photons by the ionization plates. Additionally, photons may pass through the chamber without interacting with the gas molecules. However, one advantage to this situation may be that some of the photons absorbed by the plates were scattered photons. As in conventional radiography scatter also adversely effects the CT image. Therefore, it is reasonable to conclude that the gas detectors have low scatter acceptability. Scintillation detectors convert almost all the information it receives including scattered photons therefore, the detectors have high scatter acceptability.

The dynamic range describes how many levels of information a detector can detect. The dynamic range determines the ability of a detector to detect and differentiate a wide range of x-ray intensities. "Dynamic range of a detector describes the range of x-ray exposures at the detector to which the system can respond without saturation and produce satisfactory gray-scale images (Morgan, 1983)." Current CT systems have an approximate dynamic range of 1,000,000 to 1 and 1,100 views or projections a second. CT systems have the ability to respond to 1,000,000 x-ray intensities at approximately 1,100 views per second. Unfortunately, display systems and human visual perception limits the full use of this massive amount of data.

Data Acquisition System (DAS)

Once the detector generates the analog or electrical signal it is directed to the data acquisition system (DAS). The analog signal generated by the detector is a weak signal and must be amplified to further be analyzed. Amplifying the electrical signal is one of the tasks performed by the data acquisition system (DAS) (Seeram, 1994). The DAS is located in the gantry right after or above the detector system. In some modern CT scanning systems the signal amplification occurs within the detector itself. Before the projection or raw data, which is currently in the form of an electrical or analog signal, goes to the computer it must be converted to digital information. The computer does not "understand" analog signals therefore, the information must be converted to digital information. This task is accomplished by an analog to digital converter which is an essential component of the DAS. The digital signal is transferred to an array processor. The array processor solves the statistical information using algorithmic calculations essential for mathematical reconstruction of a CT image. An array processor is a specialized high speed computer designed to execute mathematical algorithms for the purpose of reconstruction (Berland, 1987). The array processor solves reconstruction mathematics faster than a standard microprocessor. It is important to note that special algorithms may require several seconds to several minutes for a standard microprocessor to compute. Recently, processors that compute CT reconstruction mathematics faster than an array processors have been utilized to solve reconstruction mathematics essential to the development of CT fluoroscopy. The term image or reconstruction generator is used to describe this type of computer.

Further discussion of the CT system computer and image reconstruction will be provided in a future module titled CT Instrumentation and Physics Part 2.

CT Patient Table or Couch

The final component of the scan or imaging system is the patient table or couch. CT tables or couches should be made with a material that will not cause artifacts when scanned. Many CT tables or couches are made of a carbon fiber material. The movement of the table or couch is referred to as incrementation or indexing. Helical/spiral CT table incrementation or indexing is quantified in millimeters per second mm/sec because the table is moving for the entire scan.

All table or couch designs have weight limits that if exceeded may compromise incrementation or indexing accuracy. Various attachments are available for different types of scanning procedures. Attachments for direct coronal scanning and therapy planning are commonly used in many CT departments.

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