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College of Engineering

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Lecture Four

Data Acquisition, part 1

INTRODUCTION

CT scanners are complex, with many different components involved in the process of creating an image. Adding to the complexity, different CT manufacturers often modify the design of various components. From a broad perspective, all makes and models of CT scanners are similar in that they consist of a scanning gantry, x-ray generator, computer system, operator's console, and physician's viewing console.

To help simplify the process, CT can be broken down into three segments: data acquisition, image reconstruction, and image display. Data are acquired when x-rays pass through a patient to strike a detector and are recorded. The major components that are involved in this phase of image creation are the gantry and the patient table (Fig. 2-1).

FIGURE 2-1 The gantry and patient table are major components of a CT image system. (Courtesy of Siemens AG.)

GANTRY

The gantry is the ring-shaped part of the CT scanner. It houses many of the components necessary to produce and detect x-rays (Fig. 2-2). Components are mounted on a rotating scan frame. Gantries vary in total size as well as in the diameter of the opening, or aperture. The range of aperture size is typically 70 to 90cm.

The CT gantry can be tilted either forward or backward as needed to accommodate a variety of patients and examination protocols. The degree of tilt varies among systems, but ±15° to ±30° is usual. The gantry also includes a laser light that is used to position the patient within the scanner.

Control panels located on either side of the gantry opening allow the technologist to control the alignment lights, gantry tilt, and table movement. In most scanners, these functions may also be controlled via the operator's console. A microphone is embedded in the gantry to allow communication between the patient and the technologist throughout the scan procedure.

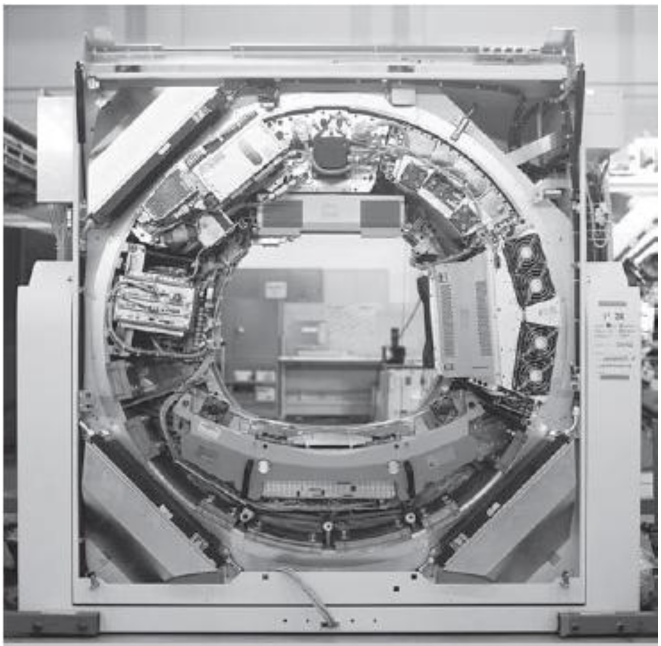
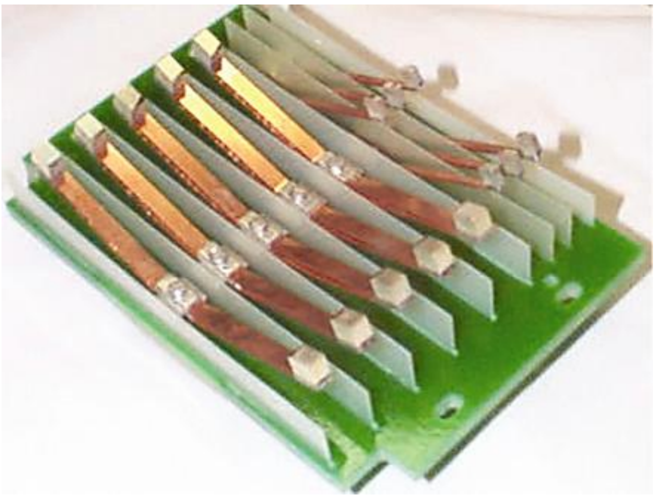
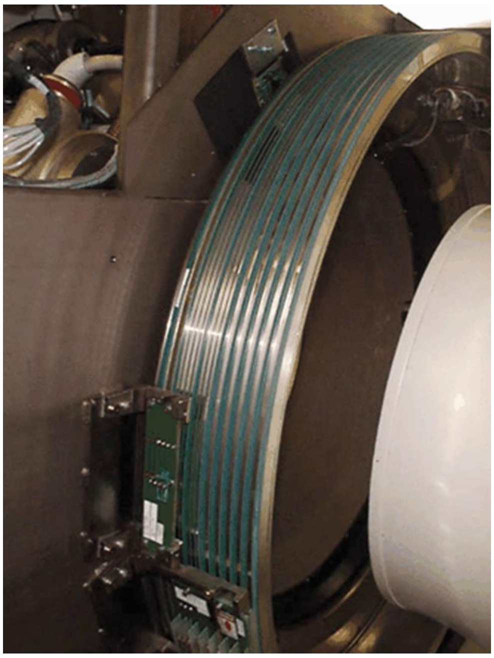


FIGURE 2-2 The gantry houses many of the components necessary to produce and detect x-rays. The gantry cover is removed on this third-generation scanner configuration to reveal the components necessary for data acquisition, including the x-ray tube and detector array. Image courtesy of Siemens AG.

Slip Rings

Early CT scanners used recoiling system cables to rotate the gantry frame. This design limited the scan method to the step-and-shoot mode and considerably limited the gantry rotation times. Current systems use electromechanical devices called slip rings. Slip rings use a brush-like apparatus to provide continuous electrical power and electronic communication across a rotating surface. They permit the gantry frame to rotate continuously, eliminating the need to straighten twisted system cables.





Slip rings used to bring power to x-ray tube on rotating gantry of a helical CT machine and, for some designs, to acquire information from the detector array. The shiny metal strips carry electric signals that are swept off by special brushes. The brushes are not in the form of bristles but rather of metal blocks (in this case a silver alloy). The five pairs of larger brushes provide the voltage required by the x-ray tube, and the three pairs of smaller ones transfer signals from the gantry controller.

Generator

High-frequency generators are currently used in CT. They are small enough so that they can be located within the gantry. Highly stable three-phase generators have also been used, but because these are stand-alone units located near the gantry and require cables, they have become obsolete.

Generators produce high voltage and transmit it to the x-ray tube. The power capacity of the generator is listed in kilowatts (kW). The power capacity of the generator determines the range of exposure techniques (i.e., kV and mA settings) available on a particular system.

CT generators produce high kV (generally 120–140 kV) to increase the intensity of the beam, which will increase the penetrating ability of the x-ray beam and thereby reduce patient dose. In addition, a higher kV setting will help to reduce the heat load on the x-ray tube by allowing a lower mA setting. Reducing the heat load on the x-ray tube will extend the life of the tube.

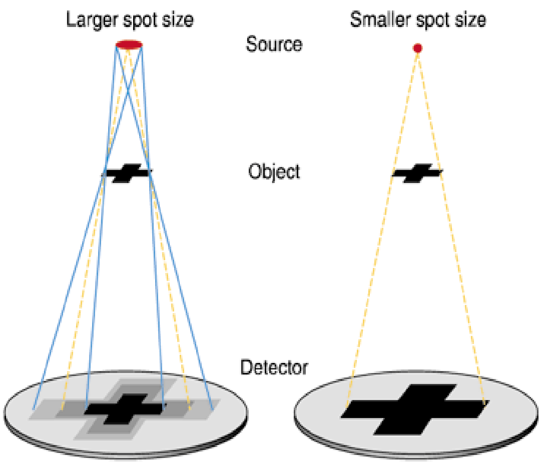
Cooling Systems

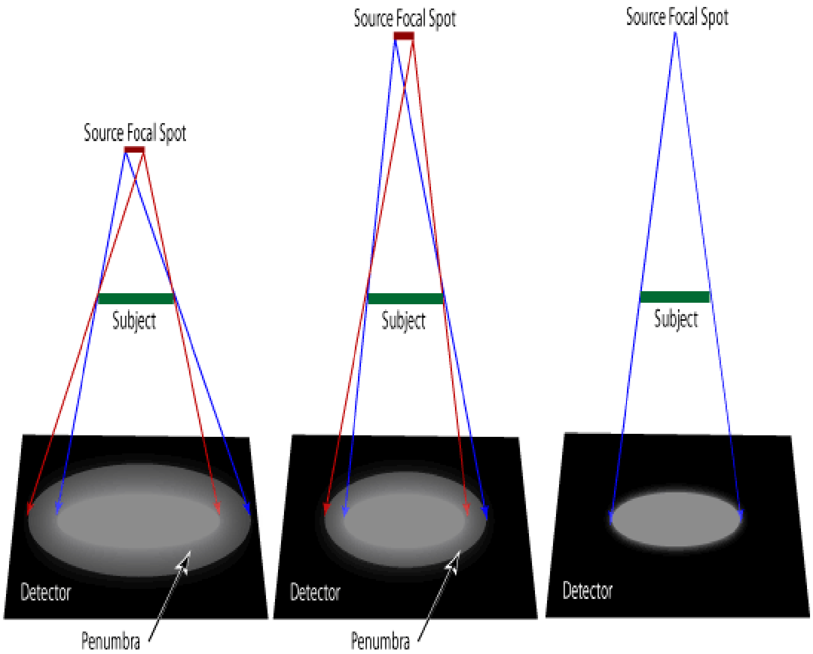
Cooling mechanisms are included in the gantry. They can take different forms, such as blowers, filters, or devices that perform oil-to-air heat exchange. Cooling mechanisms are important because many imaging components can be affected by temperature fluctuation.

X-ray Source

X-ray tubes produce the x-ray photons that create the CT image. Their design is a modification of a standard rotating anode tube, such as the type used in angiography. Tungsten, with an atomic number of 74, is often used for the anode target material because it produces a higher-intensity x-ray beam. This is because the intensity of x-ray production is approximately proportional to the atomic number of the target material. CT tubes often contain more than one size of focal spot; 0.5 and 1.0 mm are common sizes.

Just as in standard x-ray tubes, because of reduced penumbra small focal spots in CT tubes produce sharper images (i.e., better spatial resolution), but because they concentrate heat onto a smaller portion of the anode they cannot tolerate as much heat.





An enormous amount of stress is placed on the CT tube. Scanning protocols often require multiple long exposures performed on numerous patients per day. A CT tube must be designed to handle such stress. The way a tube dissipates the heat that is created during x-ray production is critical. All manufacturers list generator and tube cooling capabilities in their product specifications. These specifications usually list the system generator’s maximum power in kW. Also listed is the anode heat capacity in million heat units (MHU) and the maximum anode heat dissipation rate in thousand heat units (KHU). These specifications can be helpful in comparing various CT systems. It is important to remember that these values represent the upper limit of tube performance. It is also important to compare the length of protocols that the tube will allow and how quickly they can be repeated.

Filtration

Compensating filters are used to shape the x-ray beam. They reduce the radiation dose to the patient and help to minimize image artifact. Since radiation emitted by CT x-ray tubes is polychromatic, filtering the x-ray beam helps to reduce the range of x-ray energies that reach the patient by removing the long-wavelength (or "soft") x-rays. These long-wavelength x-rays are readily absorbed by the patient; therefore, they do not contribute to the CT image but do contribute to the radiation dose to the patient. In addition, creating a more uniform beam intensity improves the CT image by reducing artifacts that result from beam hardening.

Different filters are used when scanning the body than when scanning the head. Human body anatomy typically has a round cross section that is thicker in the middle than in the periphery. Hence, body-scanning filters are used to reduce the beam intensity at the periphery of the beam, corresponding to the thinner areas of a patient's anatomy. Because of their shape they are often referred to as bow tie filters (Fig. 2-3).

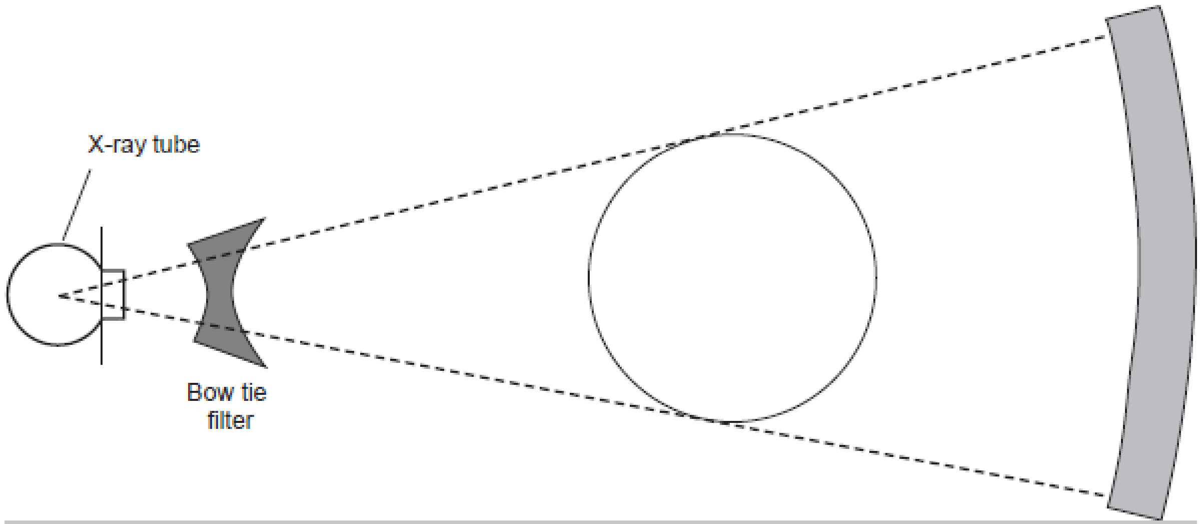


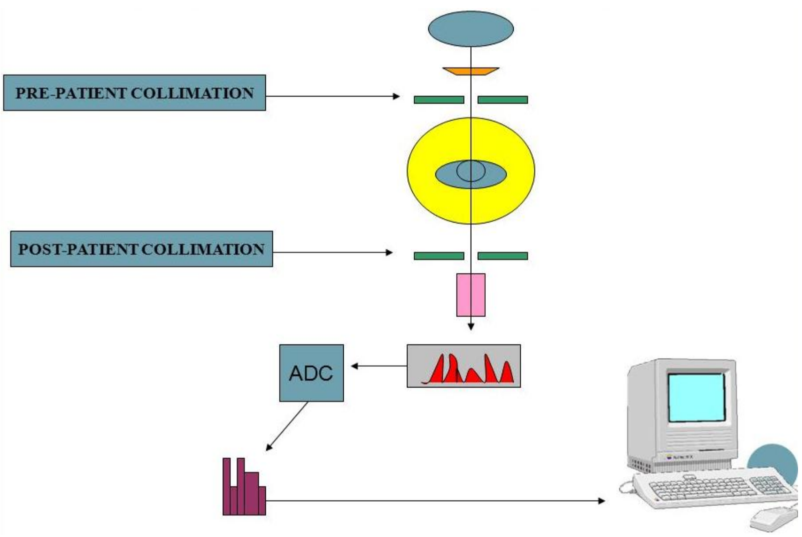
FIGURE 2-3 Filtering shapes the x-ray beam intensity. Removing low-energy x-rays minimizes patient exposure and produces a more uniform beam.

Collimation

Collimators restrict the x-ray beam to a specific area, thereby reducing scatter radiation. Scatter radiation reduces image quality and increases the radiation dose to the patient. Reducing the scatter improves contrast resolution and decreases patient dose. Collimators control the slice thickness by narrowing or widening the x-ray beam. The source collimator is located near the x-ray source and limits the amount of x-ray emerging to thin ribbons. Because it acts on the x-ray beam before it passes through the patient it is sometimes referred to as pre-patient collimation.

The source collimator affects patient dose and determines how the dose is distributed across the slice thickness (i.e., dose profile). The source collimator resembles small shutters with an opening that adjusts, dependent on the operator's selection of slice thickness. In MDCT systems (Multidetector computed tomography), slice thickness is also influenced by the detector element configuration. Scanners vary in the choices of slice thickness available. Choices range from 0.5 to 10 mm.

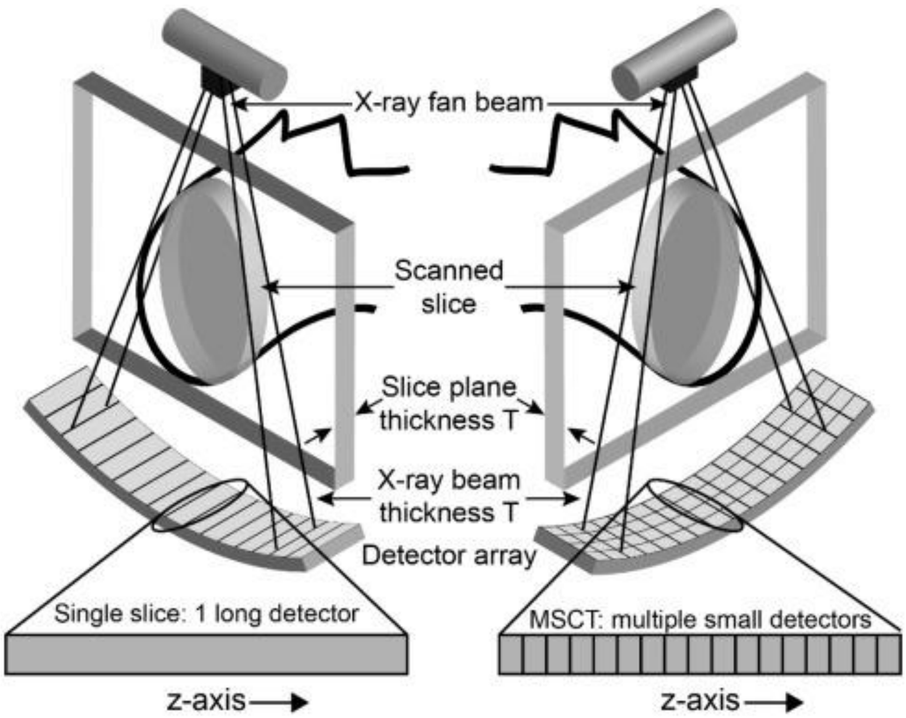
Some CT systems also use pre-detector collimation. This is located below the patient and above the detector array. Because this collimator shapes the beam after it has passed through the patient it is sometimes referred to as post-patient collimation. The primary functions of pre-detector collimators are to ensure the beam is the proper width as it enters the detector and to prevent scatter radiation from reaching the detector.



Detectors

As the x-ray beam passes through the patient it is attenuated to some degree. To create an x-ray image, we must collect information regarding the degree to which each anatomic structure attenuated the beam. In CT, we use detectors to collect the information. 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 entire collection of detectors included in a CT system. Specifically, the detector array comprises detector elements situated in an arc or a ring, each of which measures the intensity of transmitted x-ray radiation along a beam projected from the x-ray source to that particular detector element. Also included in the array are elements referred to as reference detectors that help to calibrate data and reduce artifacts.

The scan field of view determines the size of the fan beam, which, in turn, determines the number of detector elements that collect data. Detectors can be made from different substances, each with their own advantages and disadvantages.



The optimal characteristics of a detector are as follows:

1) high detector efficiency, defined as the ability of the detector to capture transmitted photons and change them to electronic signals;

2) low, or no, afterglow, defined as a brief, persistent flash of scintillation that must be taken into account and subtracted before image reconstruction;

3) high scatter suppression;

4) high stability, which allows a system to be used without the interruption of frequent calibration.

Overall detector efficiency is the product of a number of factors. These are

1) stopping power of the detector material;

2) scintillator efficiency (in solid-state types);

3) charge collection efficiency (in xenon types);

4) geometric efficiency, defined as the amount of space occupied by the detector collimator plates relative to the surface area of the detector;

5) scatter rejection.

Other terms are sometimes used to describe aspects of a detector's efficiency. Capture efficiency refers to the ability with which the detector obtains photons that have passed through the patient. Absorption efficiency refers to the number of photons absorbed by the detector and is dependent on the physical properties of the detector face (e.g., thickness, material). Response time is the time required for the signal from the detector to return to zero after stimulation of the detector by x-ray radiation so that it is ready to detect another x-ray event.

The detector response is generally a function of the detector design. Dynamic range is the ratio of the maximum signal measured to the minimum signal the detectors can measure. All new scanners possess detectors of the solid-state crystal variety. Detectors made from xenon gas have been manufactured but have largely become obsolete as their design prevents them from use in MDCT systems. However, some xenon gas detector systems may still be in use on older models.

Xenon Gas Detectors

Pressurized xenon gas fills hollow chambers to produce detectors that absorb approximately 60% to 87% of the photons that reach them. Xenon gas is used because of its ability to remain stable under pressure. Compared with the solid-state variety, xenon gas detectors are significantly less expensive to produce, somewhat easier to calibrate, and are highly stable.

xenon detector channel consists of three tungsten plates. When a photon enters the channel, it ionizes the xenon gas. These ions are accelerated and amplified by the electric field between the plates. The collected charge produces an electric current. This current is then processed as raw data. A disadvantage of xenon gas is that it must be kept under pressure in an aluminum casing. This casing filters the x-ray beam to a certain extent. Loss of x-ray photons in the casing window and the space taken up by the plates are the major factors hampering detector efficiency. Figure 2-4 demonstrates the basic structure of a xenon gas detector array.

Solid-State Crystal Detector

Solid-state detectors are also called scintillation detectors because they use a crystal that fluoresces when struck by an x-ray photon. A photodiode is attached to the crystal and transforms the light energy into electrical (analog) energy. The individual detector elements are affixed to a circuit board (Fig. 2-5).

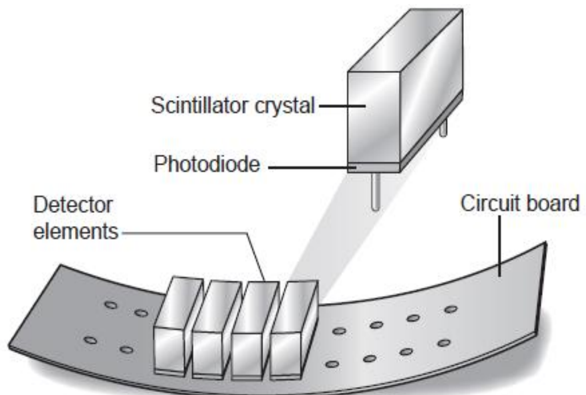


FIGURE 2-5 Structure of a solid-state detector array.

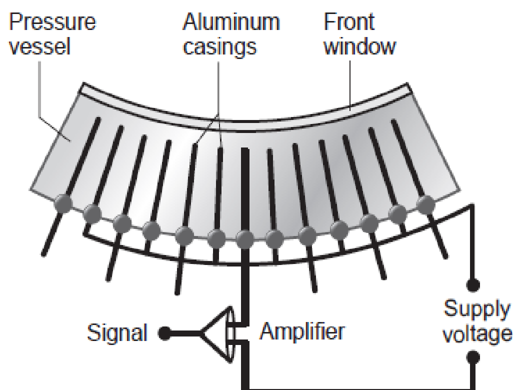


FIGURE 2-4 Structure of a xenon gas detector array.

Solid-state crystal detectors have been made from a variety of materials, including cadmium tungstate, bismuth germinate, cesium iodide, and ceramic rare earth compounds such as gadolinium or yttrium. Because these solids have high atomic numbers and high density in comparison to gases, solid-state detectors have higher absorption coefficients. They absorb nearly 100% of the photons that reach them. In addition, there is no loss in the front window, as in xenon systems. This increased absorption efficiency is the chief advantage of solid-state detectors.

Solid-state detectors may produce a brief afterglow. However, this has been greatly reduced or eliminated in modern CT detectors. Solid-state detectors are more sensitive to fluctuation in temperature and moisture than the gas variety. Table 2-1 compares solid-state detectors to the xenon gas variety.

TABLE 2-1 Characteristics of Detectors

Solid-State Crystal	Pressurized Xenon Gas
High photon absorption	Moderate photon absorption
Sensitive to temperature, moisture	Highly stable
Solid material	Low-density material (gas)
Can exhibit afterglow	No afterglow
No front window loss	Losses attributable to front window and the spaces taken up by the plates

The relative placement, shape, and size of the detectors affect the amount of scatter radiation that reaches the image. A deep, narrow detector design will accept less scatter than short, wide detectors. Detectors are separated using spacing bars. This allows the detectors to be placed in an arc or circle. Detector spacing is measured from the middle of one detector to the middle of the neighboring detector and accounts for the spacing bar. Ideally, detectors should be placed as close together as possible, so all x-rays are converted to data.

The size of the detector opening is called the aperture. A small detector is important for good spatial resolution and scatter rejection. Maximum utilization and small aperture are desirable. Figure 2-6 shows the relationship between detector arrangement and scatter acceptance.

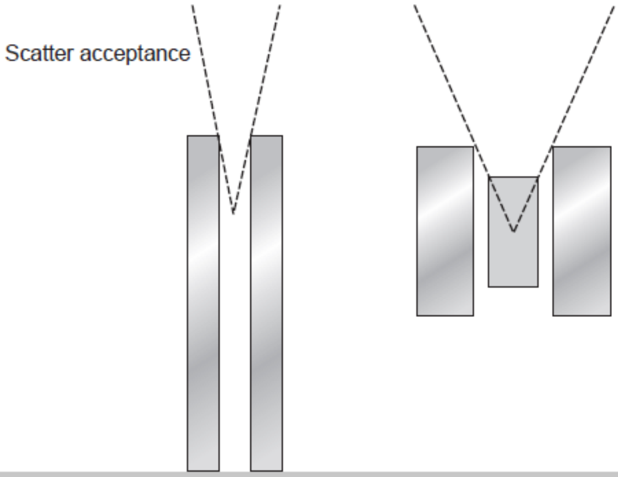


FIGURE 2-6 Detector spacing and aperture. **A.** The width and spacing of the detectors affect the amount of scatter that is recorded. **B.** Low scatter acceptance is desirable. Simple geometric principles affect scatter acceptance.