PART I
X-Ray Imaging and Computed Tomography

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1 X-Ray and Computed Tomography Imaging Principles

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1.1. INTRODUCTION TO X-RAY IMAGING

X-ray imaging is a well-known imaging modality that has been used for over 100 years since Röntgen discovered X-rays based on his observations of fluorescence. His initial results were published in 1885, and reports of diagnoses of identified fractures shortly followed. A year later, equipment manufacturers started selling X-ray equipment. Today, X-ray and its three-dimensional (3D) extension, computed tomography (CT), are used commonly in medical diagnosis.

X-rays are high-energy photons. Their generation creates incoherent beams that experience insignificant scatter when passing through various media. As a result, X-ray imaging is based on through transmission and analysis of the resulting X-ray absorption data. Typically, X-rays are detected through a combination of a phosphor screen and a light-sensitive film, as shown in Fig. 1.1. The current system, which has been used for mammography and radiography for many years, provides a good-quality analog image that is not compatible with digital storage and transmission requirements of the modern digital era. A slight variation of this common technique is used in fluoroscopy where image intensifier is used as transition stage to supply signals to CMOS cameras producing an analog image directly on a TV screen. Multiple conversions steps in this case from X-rays to electrons to light to camera display lead to poor image quality.

An alternative to the conventional detection technique, also shown in Fig. 1.1, uses a digital detector that converts X-ray photons directly into an electrical signal of digital nature. Chapter 2 in this book discusses an example of this direct detection technology using a large-area active matrix flat panel based on the amorphous silicon (a-Si). Having a digital image leads to lower storage cost and ease of electronic transmission in a future e-Healthcare era.

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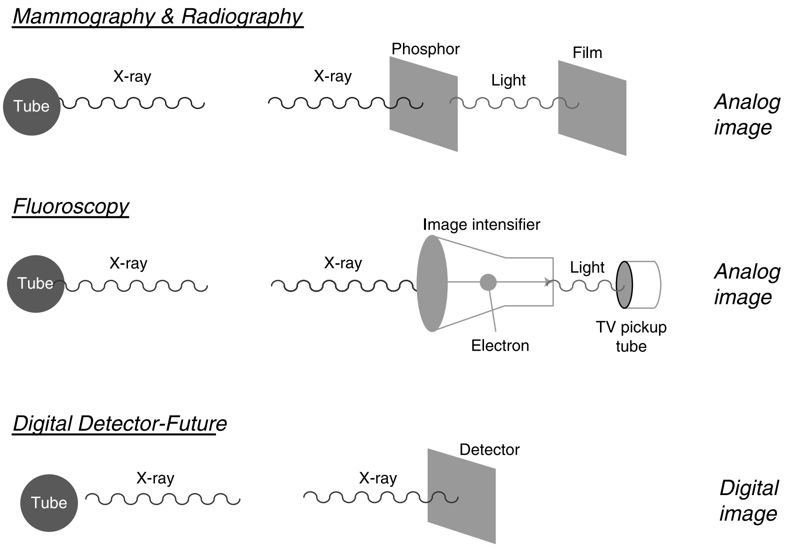

Fig. 1.1. Typical X-ray detection methods used today that provide an analog image (**top** and **middle**) and that in the future will use a digital detector (**bottom**). (From http://www.ecse.rpi.edu/censsis/.)

There are some important differences in characterization of film-based imaging and digital imaging. In this chapter and the rest of the book we will focus on digital imaging because it is a more modern technique which with time is expected to completely replace film-based imaging the same way that digital cameras have displaced analog films in consumer cameras. In digital imaging we use terms such as brightness, dynamic range, linearity, or signal-to-noise ratio instead of density, latitude, film speed, or image sharpness, the terms associated with film-based technology.

Digital X-ray detectors can operate in two regimes: photon counting and integration. In the photon counting mode, each individual photon is detected; and if its energy is higher than the set threshold, the photon is counted with its corresponding energy registered if desired. In the integrating mode the charge generated by the incoming photon is integrated in a selected time interval. Due to this principle of operation, a count rate in the photon counting mode is limited, typically to 10^6 counts per second (c/s), while it is virtually unlimited for the integrating mode. The photon counting mode can detect smaller signals, down to an individual photon, and offers a higher dynamic range (typically 10^6) compared to the integrating mode (10^4). Advantages of the photon counting mode include higher detector quantum efficiency (DQE), lower electronic noise, no need for signal digitization, and possibility of energy discrimination. The integrating mode, in turn, can operate with high count rates, and it is simple and inexpensive to implement. Chapter 3 in this book discusses differences in both photon counting and integration modes of operation in more detail.

While operation of modern X-ray based scanners can be quite complex, a basic principle behind X-ray imaging is quite simple. The technique relies on analyzing attenuation data of the object (patient) that undergoes X-ray exposure. Because different materials (internal organs) experience different levels of X-ray intensity attenuation, an image corresponding to these properties can be readily created. The attenuation characteristics are governed by the so-called Beer–Lambert law, which is expressed as follows:

$$I(z) = I_0 * \exp(-\mu z) \tag{1.1}$$

where $I(z)$ is the X-ray intensity at the detector, I_0 is the X-ray intensity at the source, z is the distance between the source and the detector planes, and μ is the attenuation coefficient that has a different value for different materials. By measuring $I(z)$ for a set of detectors, one can establish the corresponding value of the attenuation coefficients that give a representation of the image. X-ray imaging is particularly good for providing a contrast between soft and hard tissues, because the attenuation coefficient has a quite different value in both media; hence one of the first applications was to identify fractured bones.

To operate as a diagnostic technique, X-ray imaging needs a radiation source, a means of interactions between the X-ray beam and the object to be imaged, ways of registration of the radiation carrying information about the object, and finally the ability to convert that information into an electrical signal. Although widely used and inexpensive, standard X-ray technique have quite severe limitations. First, 3D structures are collapsed into 2D images, leading to highly reduced image contrast. Second, it is difficult to image soft tissues due to small differences in attenuation coefficients. Finally, standard film-based technology does not provide quantitative data and requires specialized training for accurate image assessment.

Fortunately, a 3D extension of 2D X-ray technology, called computed tomography (CT), was invented in 1972 and is in widespread use today. A basic principle behind CT is to take a large number of X-ray images at multiple angles and, based on that information, calculate the 3D image of the imaged object. CT hardware used for this application is typically called a CT scanner and is similar to an ordinary X-ray machine, albeit with much more computational power. With today's multiple-row detector helical CT scanners, 3D images can be obtained with spatial resolution approaching that of conventional radiographic images in all three dimensions.

This chapter is organized as follows. The radiation source, a well-known X-ray tube, is discussed briefly in Section 1.2. Details of interaction between photons and the object, which include absorption, reflection, scattering, and diffraction, are considered in Section 1.3. Detectors used to register the radiation events are discussed in Section 1.4, while conversion of electrical signals is mentioned in Section 1.5. Principles of computed tomography (CT) are introduced in Section 1.6, while CT scanner design is described in Section 1.7. Extension of X-ray imaging that takes into account photon energy, referred to as "color" X-ray imaging, is discussed in Section 1.8 followed by summary of future trends in Section 1.9. For more details on X-ray imaging modalities, the reader is referred to numerous books on this subject [1–7].

1.2. X-RAY GENERATION

A typical X-ray tube is shown in Fig. 1.2. Generation of X-rays depends on thermionic emission and acceleration of electrons from a heater filament. During that process, electrons emitted from cathode are accelerated by anode voltage. Kinetic energy loss at an anode is converted to X-rays. The relative position of an electron with respect to the nucleus determines the frequency and energy of the emitted X-ray.

X-rays produced in an X-ray tube contain two types of radiation: *Bremsstrahlung* and characteristic radiation. The word *Bremsstrahlung* is retained from the German language to describe the radiation that is emitted when electrons are decelerated. It is characterized by a continuous distribution of X-ray intensity and shifts toward higher frequencies when the energy of the bombarding electrons is increased. Characteristic X-rays, on the other hand, produce peaks of intensity at particular photon energies as shown in Fig. 1.3. In practice, emitted radiation is filtered, intentionally or not, producing high-pass filter response as low-energy radiation is completely attenuated. As a result, the final X-ray spectrum has band-pass type characteristics with several local peaks superimposed on it (Fig. 1.4).

The filtering effect shown in Fig. 1.4 is intentional, used to cut off X-ray energies below 20 keV in the shown example. A similar effect can be achieved unintentionally if the gap between the source and the detector is large. Figure 1.5 shows transmission characteristics through air. While 40-keV radiation is not affected by the air gap, 10-keV rays are severely attenuated, and the degree of their attenuation is dependent on the distance.

X-ray generation is a fairly inefficient process because most of the electrical power ends up as heat at the anode. Therefore, an X-ray tube is also a heater, and heat

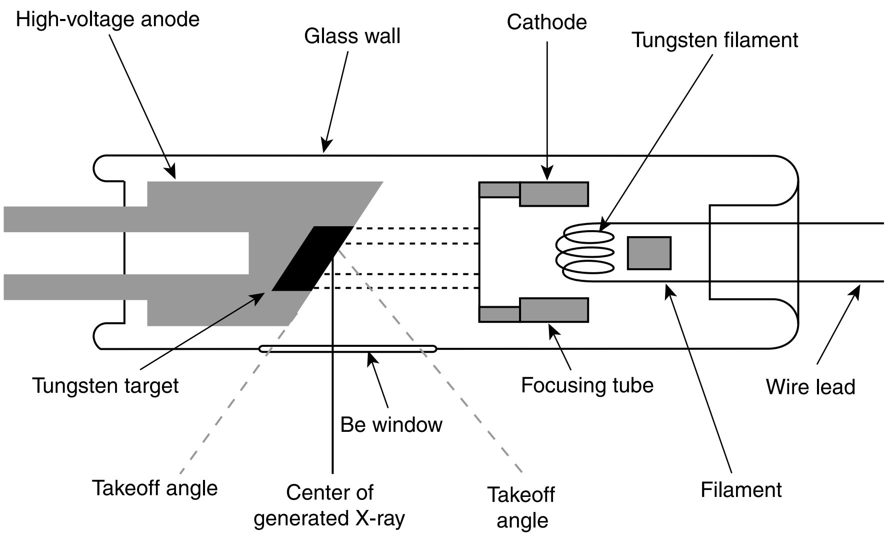

Fig. 1.2. X-ray tube. From http://www.siint.com/en/technology/xrf_descriptions1_e.html.

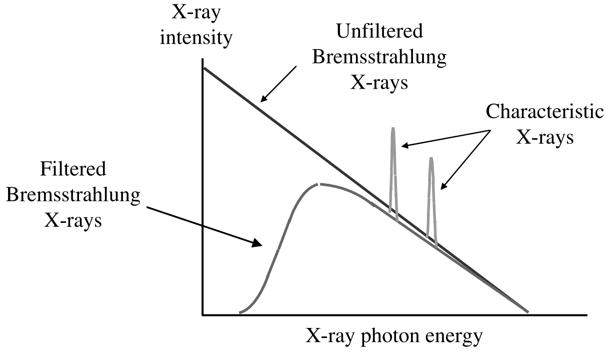

Fig. 1.3. Schematic representation of X-ray intensity frequency characteristics.

extraction problems are primary problems in the equipment design and manufacturing. In addition, only a few percent of the generated X-rays end up being absorbed at the detector because the X-ray beam is not collimated and photons are radiated in all possible directions. X-ray photon energy is related to acceleration voltage; so if the acceleration voltage is 20 kV, it will produce 20-keV photons. Clearly, X-ray-based equipment is clearly not suitable for home use! The total number of photons generated is proportional to the cathode current, which typically is several milliamperes. A typical X-ray system uses step-up transformers to produce high-voltage (HV) as schematically shown in Fig. 1.6.

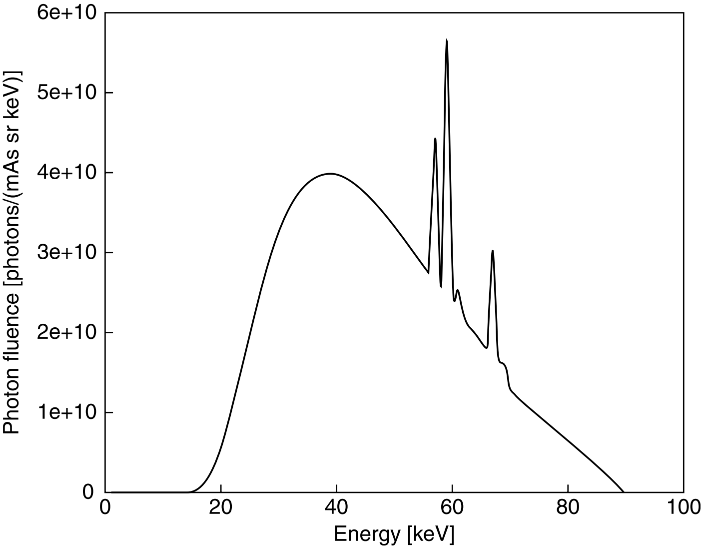

Fig. 1.4. Simulated X-ray intensity characteristics for a 90-keV tube with a 1.5-mm Be and 2.7-mm Al filter. (From Roessl and Proksa [8], with permission.)

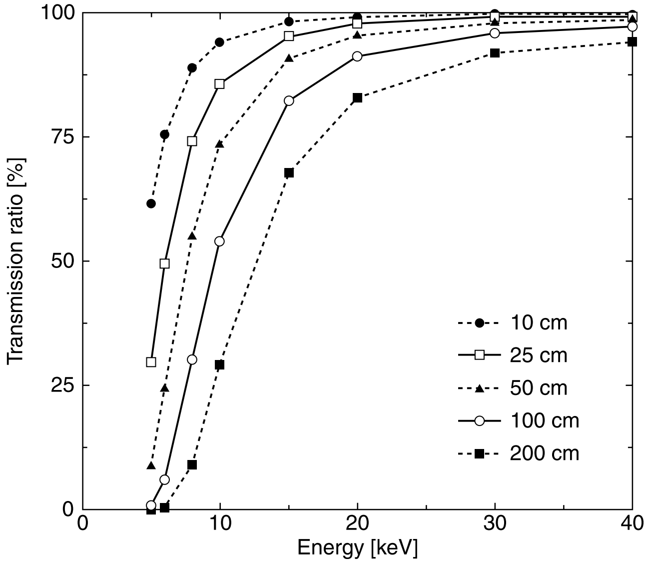

Fig. 1.5. X-rays transmission characteristics for the air path as a function of X-ray energy. (From Miyajima and Imagawa [9], with permission.)

X-ray tubes used in computed tomography (CT) are subjected to higher thermal loads in than in any other diagnostic X-ray application. In early CT scanners, stationary anode X-ray tubes were used, since the long scan times meant that the instantaneous power level was low. Long scan times also allowed significant heat dissipation. Shorter scan times in later versions of CT scanners required high-power

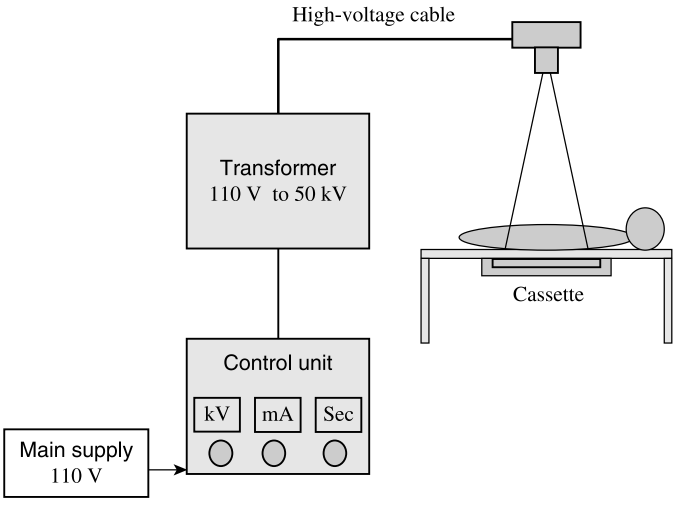

Fig. 1.6. Schematic representation of a standard X-ray system.

X-ray tubes and use of liquid-cooled rotating anodes for efficient thermal dissipation. The recent introduction of helical CT with continuous scanner rotation placed even more demands on X-ray tubes; this is clearly a challenging engineering problem because the dissipated power is in the kilowatt range.

X-rays represent ionizing radiation that at significant dose will cause tissue damage. The traditional unit of absorbed dose is the rad. 1 rad is defined as the amount of X-ray radiation that imparts 100 ergs of energy per gram of tissue or, as re-stated in SI units, causes 0.01 joule of energy to be absorbed per kilogram of matter. As a frame of reference, a typical chest X-ray exposure is about 50 mrads, while exposure of 50 rads causes radiation sickness. In the SI system, rad is now superseded by gray, with the following simple relationship between the two: 1 gray equals 0.01 rad.

1.3. X-RAY INTERACTION WITH MATTER

X-rays interact with matter in several ways that can be divided into absorption and scattering effects. Primary effects at energies of interest in medical applications are photoelectric effect, Compton scatter, and coherent scatter. In the photoelectric effect, the energy of an X-ray photon is absorbed by an orbital electron, which in turn is ejected from an atom. During this process, X-rays are converted into electric charges, a process very useful for radiation detection. Scattering effects can be of Compton nature, where some energy loss is involved, and coherent, without any energy loss. In Compton scatter, some of the X-ray energy is transferred to an electron, and the X-ray photon travels on with an altered direction and less energy. The Compton process might sometime be utilized in medical imaging in so-called Compton cameras, but frequently it is an undesired effect. As opposed to the Compton effect in coherent scatter, all X-ray energy interacts with the atom, but is later re-radiated with same energy in an arbitrary direction. As a result, the photon changes direction but still carriers the same energy, a process quite detrimental to medical imaging because the original path of photon from a source to a detector is altered.

The relative probability of above processes is dependent on the photon energy and characteristics of the matter with which it interacts. In order to focus our discussion here, we will discuss some details of photon interaction with a semiconductor material called CZT. CZT stands for cadium zinc telluride and is currently considered as the most promising detector material for X-ray and γ-ray direct detection in medical imaging for reasons that are explained later in this chapter. The relative probability of absorption/effect is plotted in Fig. 1.7. The photoelectric effect is a dominant one in the considered energy range of 20–300 keV; although at higher energies, Compton scattering becomes equally probable. At the energy of 122 keV, which represents a characteristic cobalt radiation line, the photoelectric effect has 82% probability of happening, Rayleigh scattering 7%, and Compton scattering 11%.

Note that the photoelectric line shows an interesting behavior in the 20- to 40-keV range due atomic structure. The corresponding attenuation length, shown in Fig. 1.8, varies from 0.05 to 0.17 mm. This indicates that even a thin CZT detector will effectively absorb all radiation in that energy range.

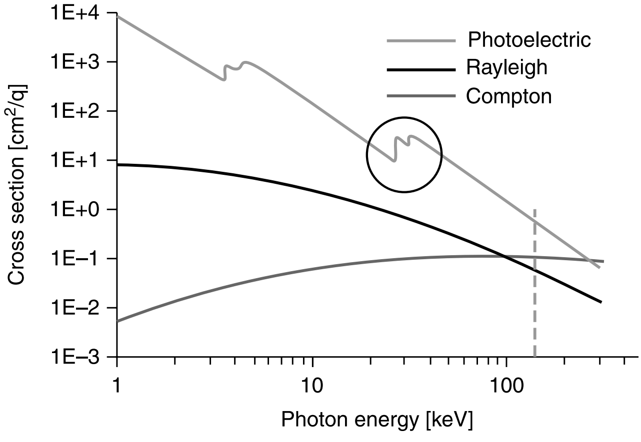

Fig. 1.7. Effective photon cross sections for the photoelectric effect, Rayleigh scattering, and Compton scattering in CZT. The dashed vertical line indicates the 122-keV cobalt line.

The photoelectric effect is one of the energy loss processes where the photon effectively disappears after the interaction. A complete absorption of the photon energy is the desired effect for X-ray detection. The name photoelectron comes from a process of ejecting an electron from one of the atomic shells of the media. After the ejection of the photoelectron, the atom is ionized. The vacancy in the bound shell is refilled with

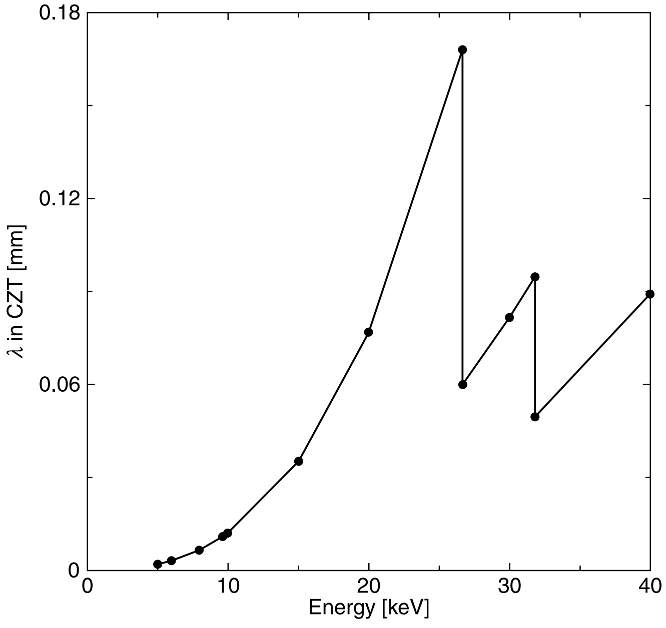

Fig. 1.8. Attenuation length in CZT for X-ray energies from 5 to 40 keV.

an electron from the surrounding medium or from an upper atom shell. This may lead either to the emission of one or more characteristic fluorescence X-rays or to the ejection of an electron from one of the outer shells, called an Auger electron. Depending whether tellurium, cadmium, or zinc atoms are involved, the resulting soft X-rays energies might be in an 8- to 31-keV range (Te 27–31 keV; Cd 23–26 keV; Zn 8–10 keV).

Having addressed absorption effects, let us change our focus to scattering events: Rayleigh and Compton. Rayleigh scattering describes photon scattering by atoms as a whole, frequently also called coherent scattering because the electrons of the atom contribute to the interaction in a coherent manner and there is no energy transferred to the CZT material. The elastic scattering process changes only the direction of the incoming photon. For these reasons, Rayleigh scattering is detrimental to medical imaging as the original photon trajectory is changed.

Unlike Rayleigh scattering, the Compton effect deals with photons that are scattered by free electrons and as a result lose some of their primary energy. This mechanism can contribute significantly to the measured energy spectrum. The change in the photon energy increases with increasing scattering angle. The energy and momentum lost by the photon is transferred to one electron, called the recoil electron, which is emitted under a certain angle with respect to the direction of the incoming photon and can have a maximum kinetic energy defining the so-called Compton edge. The Compton edge can frequently be visible in the measured spectrum as an abrupt end to the energy tail caused by Compton scattering (Fig. 1.9).

The Compton scattering equation describes the change in photon energy and its corresponding wavelength as

$$\lambda' - \lambda = \frac{h}{m_e c}(1 - \cos\theta)$$

where λ is the wavelength of the photon before scattering, λ' is the wavelength of the photon after scattering, m_e is the mass of the electron, θ is the angle by which the

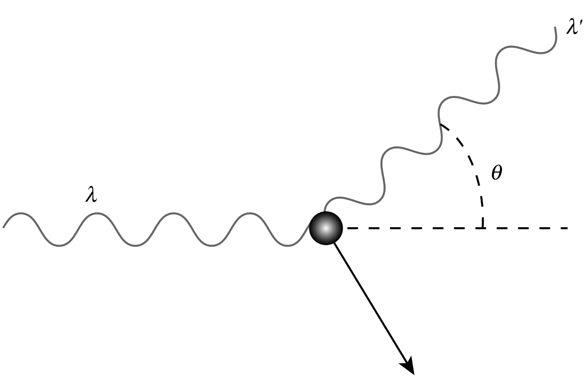

Fig. 1.9. Illustration of Compton scattering event.

photon's trajectory changes, h is Planck's constant, and c is the speed of light. Substituting textbook values for m_e, c, and h, we can obtain a characteristic Compton wavelength defined as $h/(m_e c)$ to be equal to 2.4 picometers (pm).

The Compton equation has two interesting properties. First, the characteristic Compton wavelength value is small compared to the typical X-ray wavelengths used in medical imaging (10-keV wavelength is about 120 pm). As a result, the maximum wavelength change is only a fraction of the original wavelength value. Second, the largest change in the photon energy can only be expected for angles θ close to 180 degrees. The maximum wavelength change is twice the Compton wavelength change.

1.4. X-RAY DETECTION

To detect X-rays, one needs a material that is capable of absorption of the incoming radiation and transformation of the corresponding photon energy into an electrical signal. There are two ways of accomplishing this result: indirectly and directly. The indirect process uses materials called scintillators that convert X-rays photons to visible light. The generated light is in turn detected using conventional photodetectors. The direct conversion process uses semiconductor detectors that convert X-ray photons directly into electrical signals.

In typical X-ray indirect detection applications, either thallium-doped sodium ionide NaI(Tl) or thallium-doped cesium iodide CsI(Tl) is used as a scintillator because thallium-doped crystals are one of the best-known materials for scintillation processes. One of the fundamental problems with scintillators is a low efficiency of the indirect process. Once converted to visible light, the signal is no longer proceeding along the direction of the incoming photon; instead light is emitted in all directions. As a result, very few generated photons reach the subsequent photosensitive detector.

Other manufacturers use direct conversion techniques with amorphous selenium, amorphous silicon, or cadmium zinc telluride (CZT) semiconductor detectors. Chapter 2 of this book deals with circuit, process, and device development for large-area, digital imaging applications using amorphous silicon (a-Si), and it discusses in detail the implementation of this new technology. For low-energy X-ray detection, standard silicon is frequently used, mainly because of good homogeneity and very well stabilized technology. A 300-μm-thick silicon detector converts nearly all 8-keV X-rays, but only 2% of 60-keV X-rays. High-purity germanium (HPGe) and compound semiconductors, like gallium arsenide (GaAs), cadmium telluride (CdTe), cadmium zinc telluride (CdZnTe), and mercuric iodide, show much better radiation stopping power compared to silicon due to their higher atomic numbers.

In the indirect detection mode, an additional photomultiplication is required to compensate for very low signal magnitude. The process can utilize devices called photomultiplier tubes, which provide very high gain (10^6), low noise, fast response, and capability of detecting single photoelectrons. Unfortunately, these devices are sensitive to magnetic field and bulky. Their large size makes some medical imaging equipment look like vacuum tube technology used in electronics before transistors was invented.

More miniaturized equipment can be built using avalanche photodiodes (APD). Silicon APDs have high quantum efficiency (QE over 60%) and broad spectral response, but their gain is not that high (10^3 at best) and their additional noise degrades system performance. Another promising approach is based on silicon photomultipliers (SiPM). These devices can be manufactured in a standard CMOS process and offer very high gain as PMTs. They are, however, sensitive to voltage and temperature fluctuations and present several operation problems that include cross-talk and afterpulsing degradation issues.

PMTs are still the main technology in today's clinical nuclear medicine. Ongoing improvements in QE and timing performance while reducing already low cost for large sizes promise this incumbent technology to be dominant for a number of years at least in low-end, low-performance equipment. APDs have matured and a number of APD-based scanners have been built. SiPM is a new technology with a promising future, and it remains to be seen whether it is applied in commercial equipment. However, ultimate gain in performance can only be obtained with the direct detection. CZT-detector-based scanners offer the highest performance currently available in X-ray and γ-ray medical imaging, albeit at high cost of manufacture at the moment.

CZT detectors have high potential as X-ray spectrometers due to relatively high atomic number (Cd 48, Zn 30, and Te 52) and high density (approximately 5.9 g cm^{-3}). The wide CZT bandgap (about 1.6 eV) enables the detectors to be operated at room temperature. The main drawback of CZT detectors is their poor hole charge transport properties that create polarization problems in high-count rate systems. Another disadvantage is high cost of manufacturing defect-free CZT crystals. With recent improvements in CZT volume production, these problems are expected to be overcome soon [10–12].

1.5. ELECTRONICS FOR X-RAY DETECTION

Advanced Very Large Scale of Integration (VLSI) electronics and techniques of designing and prototyping integrated circuits (ICs) allow integration of a large number of channels into a single silicon chip, so each imaging pixel of the detector array can be read out by an individual electronic channel. Such hardware offers post-processing capability and good spatial resolution of an image, as well as large dynamic range. If, in addition, digital imaging systems are able to extract some energy information about X-ray radiation, this information can be used to extract additional media information in radiology, material screening, or diffractometry as discussed later in the color X-ray section of this chapter (Section 1.8).

The fast front-end electronics for a large array of X-ray sensors should amplify and filter small signals from the each sensor element, perform analog-to-digital conversion, and then store the data in digital form. Because of the complexity of the multichannel mixed-mode integrated circuit, the important problems such as power limitation, low level of noise, good matching performance, and crosstalk effects must be solved simultaneously. Chapters 3, 4, and 6 in this book, as well as references 13–16, discuss architectures and related circuit solutions for the readout electronics for the digital X-ray imaging systems.

A primary example of VLSI electronics used for X-ray detection is Medipix consortium and related chip development (see http://medipix.web.cern.ch/MEDIPIX/ for details). A third-generation Medipix3 chip is a CMOS pixel detector readout chip designed to be connected to a segmented semiconductor sensor such as CZT. Like its predecessor, Medipix2, it acts as a camera taking images based on the number of X-ray photons which hit the pixels when the electronic shutter is open. Medipix3 aims to allow for color imaging and dead time free operation. It also seeks to mitigate the effect of charge sharing by summing charge between neighboring pixels and allocating the sum or hit to the individual pixel with the highest collected charge.

1.6. CT IMAGING PRINCIPLE

Standard X-ray imaging poses fundamental limitations, making it only useful for clinical diagnosis in cases of initial screening or simple bone fractures. The primary limitation comes from the fact that three-dimensional (3D) objects are collapsed into 2D images. The technique offers only low soft-tissue contrast and is not very quantitative. As mentioned earlier X-ray CT tomography relies on taking a large number of X-rays at multiple angles. The Greek roots of the term "tomography" are "tomo"—act of cutting and "graphos"—image. Based on the X-ray tomographical information a 3D image is calculated although the computational load can be very heavy. In the era of powerful Intel processors this is no longer a limitation, a conventional personal computer (PC) is sufficient in most cases.

As shown in Fig. 1.10, a CT scanner makes many measurements of attenuation through the plane of a finite-thickness cross section of the body. The system uses these data to reconstruct a digital image of the cross section, with each pixel in the image representing a measurement of the mean attenuation of a boxlike element (called a voxel) that extends through the thickness of the section. An attenuation measurement quantifies the fraction of radiation removed in passing through a given amount of a specific material of thickness. Each attenuation measurement is called a *ray sum* because attenuation along a specific straight-line path through the patient from the tube focal spot to a detector is the sum of the individual attenuations of all materials along the path. Rays are collected in sets called *projections*, which are made across the patient in a particular direction in the section plane. There may be about a thousand rays in a single projection. To reconstruct the image from the ray measurements, each voxel must be viewed from multiple different directions. A complete data set requires many projections at rotational intervals of 1° or less around the cross section. Back-projection effectively reverses the attenuation process by adding the attenuation value of each ray in each projection back through the reconstruction matrix.

The attenuation coefficient of water is obtained during calibration of the CT machine. Voxels containing materials that attenuate more than water (for example, muscle tissue or bones) have positive CT numbers, whereas materials with less attenuation than water (for example, lungs) have negative CT numbers.

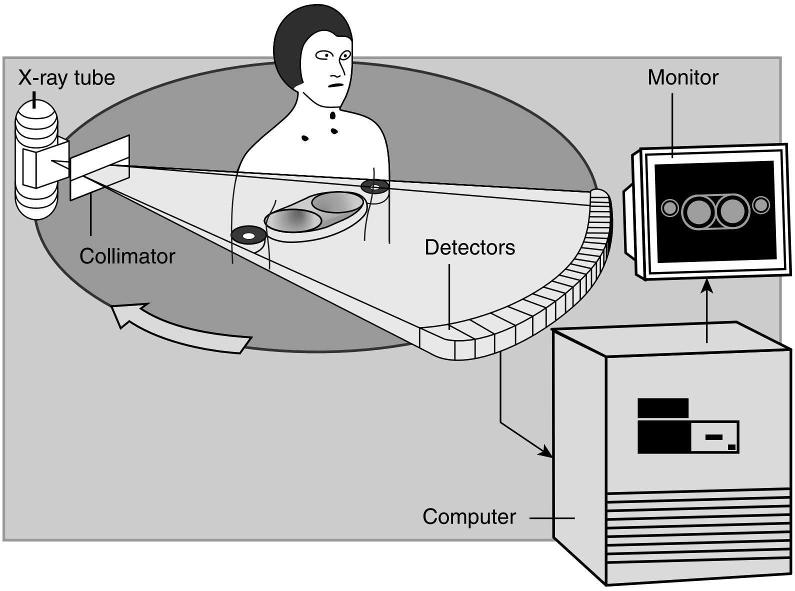

Fig. 1.10. Schematic illustration of CT imaging operation.

The CT concept was invented in 1972 by Godfrey Hounsfield, who introduced computed axial transverse scanning. As a result, the first CT machines were known as CAT scanners. The scanner is quite similar to other pieces of X-ray imaging equipment with additional requirements for movements and signal processing. In the next section we will review architecture of CT scanners as they evolved from 1972 until now.

1.7. CT SCANNERS

Computed tomography (CT) is a method of acquiring and reconstructing the image of a thin cross section on the basis of measurements of X-ray attenuation. In comparison with conventional radiographs, CT images are free of superimposing tissues and are capable of much higher contrast due to elimination of scatter. Most of the developments in CT since its introduction can be considered as attempts to provide faster acquisition times, better spatial resolution, and shorter computer reconstruction times. From the early designs, the technology progressed with faster scanning times and higher scanning plane resolution, but true three-dimensional (3D) imaging became practical only recently with helical/spiral scanning capabilities.

Despite its internal complexity and hefty price, a CT scanner is basically a set of rotating X-ray tubes and radiation detectors. Figure 1.11 shows a schematic representation of a typical CT system. In order to achieve a 3D image, lots of computing power is required so every scanner has a built-in personal computer (PC). In addition to

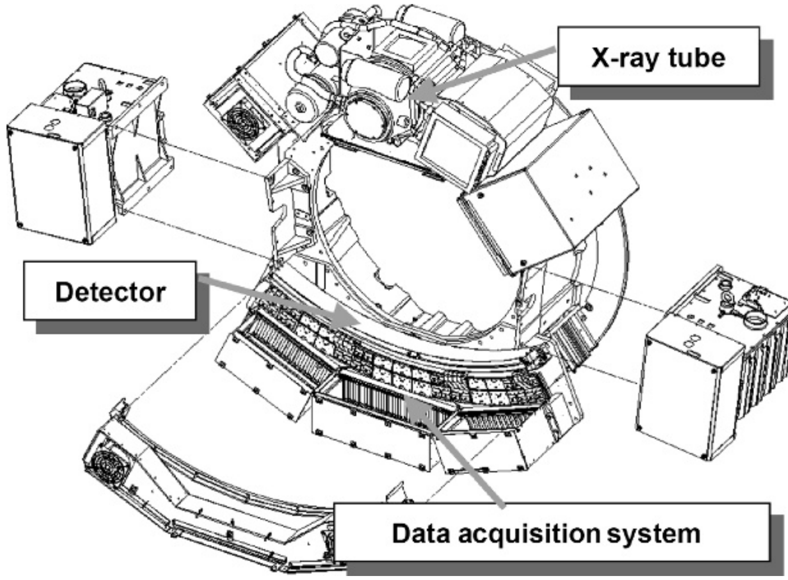

Fig. 1.11. Schematic representation of a typical CT scanner.

medical imaging CT scanners are being used for nondestructive evaluation of materials and luggage inspection.

First generation of scanners used parallel beam design with one or two detectors used in parallel. The early versions required about 5 minutes for a single scan and thus were restricted to regions where patient motion could be controlled (head, for example). Since procedures consisted of a series of scans, procedure time was reduced somewhat by using two detectors so that two parallel sections were acquired in one scan. Although the contrast resolution of internal structures was unprecedented at the time, images had poor spatial resolution of the order of a few millimeters.

Second-generation scanners introduced a small fan beam, rotation, and a larger number of detectors. Later designers realized that if a pure rotational scanning motion could be used, then it would be possible to use higher-power, rotating anode X-ray tubes and thus improve scan speeds in thicker body parts. One of the first designs to do so was the so-called third-generation or rotate–rotate geometry. In these scanners, the X-ray tube is collimated to a wide, fan-shaped X-ray beam and directed toward an arc-shaped row of detectors. The number of detectors varied from few hundred in early versions to nearly a thousand in modern scanners.

Fourth-generation scanners made additional improvements and introduced even more detectors (up to 5000 of them) in one machine placing them in a concentric ring. It was later realized that if multiple sections could be acquired in a single breath hold, a considerable improvement in the ability to image structures in regions susceptible to physiologic motion could result. However, this required some technological advances, which led to the development of helical/spiral CT scanners. In conventional CT, 360 degrees of data are acquired for one image. To get another image,

Fig. 1.12. Manufacturing of a modern CT scanner. (From Siemens web site.)

the gantry is moved to next location. Helical CT, on other hand, covers a nonplanar geometry. In this technique a table with a patient moves while the image is being taken. This movement is slow: A typical speed is a few millimeters per second. Most algorithms used in helical scanners are the same as with conventional CT, but an added interpolation step in the direction of the table movement is required. The development of helical or spiral CT around 1990 was a truly revolutionary advancement in CT scanning that finally allowed true 3D image acquisition within a single breath hold.

CT scanners for full-body analysis are large industrial machines. Figure 1.12 shows a manufacturing step of a dual-source CT scanner, which in this example is primarily used for heart examinations. The device contains two X-ray tubes and two set of detectors which allow taking clinical images at two different X-ray energies simultaneously. This dual-energy scanning offers potential of differentiating materials beyond the visualization of morphology—for example, direct subtraction of either vessels or bone during scanning. Using multiple energy scanning leads eventually to color X-ray imaging, a technique described in the next section. Before we conclude the CT scanning section, let us note that the hardware shown in Fig. 1.12 has 78-cm gantry bore, 200-cm scan range, and 160-kW generator power. One can easily imagine special physical accommodation conditions required for these machines to be deployed in hospitals.

1.8. COLOR X-RAY IMAGING

A traditional X-ray imaging relies on analysis of attenuation data without paying any attention to particular energy of transferred X-ray photons. However, some additional, useful information can be obtained if energy information is retained, even in a crude fashion by binning the entire energy range into few energy bins. This technique is called "color" X-ray imaging and is getting increasing attention in the imaging field [13].

As discussed previously, attenuation of X-rays by matter is dependent on the energy of the photons. The Bremsstrahlung sources of radiation in clinical X-ray computed tomography (CT) systems are characterized by a broad-energy spectrum. Therefore, different spectral components are attenuated with different strengths, a process called beam hardening. In conjunction with integrating X-ray detectors, beam hardening causes characteristic artifacts in the reconstructed images when left uncorrected.

Imaging modalities based on X-ray transmission measurements are usually regarded as rather insensitive with respect to the detection of specific drugs, contrast materials, and other substances present in the object of interest. This is due to the fact that in the transmission case the signal coming from these samples is always superimposed on the signal resulting from the anatomic background. Conventional X-ray transmission systems based on current integration do not allow us to separate the two signals. The spectral decomposition allows the selective separation of the anatomic background and contributions coming from heavy elements. In connection with photon counting detectors with reasonable energy resolution, X-ray transmission modalities—for example, X-ray CT—could allow the selective, high-resolution imaging of targeted contrast materials in addition to the conventional imaging of the anatomy.

Conventional X-ray/CT imaging relies on measuring total absorption of an object regardless of the particular energy of the incoming radiation. Different combinations of objects can produce equal absorption, while a color X-ray can differentiate between these different objects. The goal of a color X-ray is therefore to improve detectability of details, improve signal difference-to-noise ratio (SDNR), discriminate between different materials and different types of tissue, and enhance visibility of contrast media.

Color X-ray imaging is already used in some baggage inspection systems, as an automatic exposure control in mammography, and for bone mineral density measurement discrimination of bones and soft tissue using dual-energy contrast-enhanced digital subtraction techniques. VLSI chips that can discriminate against X-ray photon energies are already available in production volumes. The use of a color X-ray in medical X-ray imaging is expected to increase dramatically in the next 5–10 years.

1.9. FUTURE OF X-RAY AND CT IMAGING

The most basic task of the diagnostician is to separate abnormal subjects from normal subjects. In many cases there is significant overlap in terms of the appearance of the image. Some abnormal patients have normal-looking films, while some normal patients have abnormal-looking films. Technology optimization can be used to select an abnormality threshold; however, true medical diagnosis must be determined independently, based on biopsy confirmation, long-term patient follow-up, and other medical factors.

Medical X-ray imaging can be viewed from the physics, application, or medical points of view. The physical point of view is originating in technological aspects of X-ray imaging. What wavelength and photon energy are used? What are available sources of radiation? What electronics is used for processing? While this point of view is a dominant one in this book, we have to briefly address the other two.

The application point of view addresses the use of the technology. Is technology used for diagnosis or intervention? Morphological or functional imaging? What spatial resolution can be achieved? The medical point of view, on the other hand, relates heavily toward a patient. Is this screening procedure performed on a presumably healthy person? Or is it used for a diagnosis if patient is suspected of having a disease already? Is the image used for diagnostic purposes or as a helping tool in image-guided therapeutic interventions?

Standard X-ray screen-film technology is in widespread use but it needs to be replaced/complemented with much more accurate CT imaging, which, as discussed, is a method for acquiring and reconstructing a 3D image of an object. CT differs from conventional projection in two significant ways. First, CT forms a cross-sectional image, eliminating the superimposition of structures that occurs in plane film imaging because of compression of 3D body structures onto the two-dimensional recording system. Second, the sensitivity of CT to subtle differences in X-ray attenuation is at least a factor of 10 higher than normally achieved by screen-film recording systems because of the virtual elimination of scatter.

One obvious further development in X-ray-based imaging is digitization. As discussed in this chapter, modern CT equipment is capable of producing digital images of underlying organs by directly transforming photon energy information into electrical signals. Availability of digitally captured images throughout the whole health-care system provides enormous advantages and is instrumental in so-called telemedicine. Needless to say, there are also important security and privacy issues related to electronic patient records. In fact, one might argue that the largest obstacles to widespread implementation of this new digital imaging technology are of nontechnical nature.

While this introductory chapter is primary devoted to transmission-based X-ray modalities, there are alternative medical imaging techniques called nuclear medicine that can be used as complementary techniques that rely on a somewhat different principle. Nuclear medicine contains two major modalities: positron emission tomography (PET) and single particle emission computed tomography (SPECT). In both techniques, no external radiation is used to scan the object as in X-ray imaging. Instead, the patient is subjected to injection or oral administration of tracers with radioactive isotopes. During the procedure the clinician waits for distribution and accumulation of these isotopes in the region of interest—for example, in tumor. Subsequently, radiation is detected to determine the source distribution of the isotopes. Mapping of emitted radiation is performed either by collimator (SPECT) or by coincidence using two detectors (PET). 2D or 3D images can be created using iterative reconstruction and filtered back projection. As opposed to CT, nuclear medicine does not rely on attenuation data; in fact it is desirable to correct for attenuation effects if possible. Nuclear medicine techniques are described extensively in Chapter 5.

X-ray and γ-ray detection is performed in a very similar manner in both nuclear medicine and CT. There are, however, some important differences in how the electronics is being to used to process the collected information. SPECT imaging requires that the detector channels count individual photons (γ-ray events), which produce small quantities of charge in proportion to their energy. These charge pulses range from 0 to 20 fC and occur at rates in the range of 10–1000 photons/s/pixel. The charge (energy) from each γ-ray must be measured with a resolution better than of 0.1 fC for low image noise and good contrast. In contrast to SPECT, CT detector channels measure X-ray events at a relatively high flux (tens of millions of photons per second per pixel) producing a continuous current due the overlapping of individual charge pulses. This photon flux can vary over a 140-dB dynamic range, producing detector currents ranging from 1 pA to 10 μA, and must be measured with better than 74-dB SNR (signal-to-noise ratio) at a kilohertz sampling rate.

Standard X-ray and CT provide an anatomic image of the underlying organ. Other modalities discussed later in this book (SPECT, PET, functional MRI) are used to detect an organ function. It would be extremely advantageous if both the anatomic image and the organ function were combined in one. The area of multimodality image scanning is emerging rapidly as a main direction toward improvements in medical imaging. Combining anatomical imaging (CT, MRI, ultrasound, diffuse optical tomography) and functional imaging (SPECT, PET) promises to lead to dramatic improvements in medical diagnosis. The early multimode X-ray-based implementations (PET with CT, SPECT with CT) simply join two separated apparatus; this offers limited advantages. The true integration would involve using the same data acquisition electronics and detector technologies to obtain a perfect fusion of anatomical and metabolical images.

To accomplish the goal of multimode operation with high spatial resolution and sensitivity, several breakthrough technologies have to be used. First, novel semiconductor detectors such as CZT offer unprecedented energy resolution that is currently being exploited in SPECT gamma cameras. Avalanche photodiode and silicon multiplier coupled individually to crystal arrays of LSO scintillators would be another potential detector technology. Second, innovative circuit techniques that utilize subthreshold operation can be implemented in front-end ASICs to provide very low power systems (reduction of power dissipation by an order of magnitude compared to the state-of-the-art). Third, creative use of FPGA resources is required to provide efficient signal processing of hundreds of channels simultaneously. Finally, proven signal processing algorithms in FPGA can be implemented in medical imaging signal processor that would serve as the "Pentium" equivalent used for personal computing.

CT imaging systems create high-resolution three-dimensional X-ray images that provide anatomical definition with a millimeter spatial resolution, but lack direct physiological information. In the medical literature, it has been recognized that combining CT and SPECT in a single machine would improve the correlation of these two data sets and improve diagnostic process. Let us hope that the evolution from film-based 2D

X-ray screening toward precise imaging using 3D techniques such as CT and/or SPECT will happen in the not-too-distant future in order to improve quality of life for all of us.

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