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Photon-Counting Detectors in Computed Tomography: A Review

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Abstract

Photon-counting computed tomography (CT) is a new technique that has the potential to revolutionize clinical CT and is predicted to be the next significant advancement. In recent years, tremendous research has been conducted to demonstrate the developments in hardware assembly and its working principles. The articles in this review were obtained by conducting a search of the MEDLINE database. Photon-counting detectors (PCDs) provide excellent quality diagnostic images with high spatial resolution, reduced noise, artifacts, increased contrast-to-noise ratio, and multienergy data acquisition as compared with conventionally used energy-integrating detector (EID). The search covered articles published between 2011 and 2021. The title and abstract of each article were reviewed as determined by the search strategy. From these, eligible studies and articles that provided the working and clinical application of PCDs were selected. This article aims to provide a systematic review of the basic working principles of PCDs, emphasize the uses and clinical applications of PCDs, and compare it to EIDs. It provides a nonmathematical explanation and understanding of photon-counting CT systems for radiologists as well as clinicians.

Keywords

- photon-counting detectors
- photon-counting computed tomography
- energy-integrating detector

Introduction

Computed tomographic (CT) scanners were introduced early in the 1970s, and they have since improved diagnostic radiology.[1,2] A CT image is described as a spatial distribution of the imaged object's linear attenuation coefficients. The coefficient at each site is determined by the object's chemical composition, mass density, and X-ray photon energy.[3] A vital component of the CT scanner is its detector that is responsible for image production and has a significant impact on image quality and radiation exposure. All modern commercial CT scanners utilize solid-state detectors and have a third-generation rotate–rotate design.[4] However, the currently used energy-integrating detector (EID) system has disadvantages as of electronic noise and low signal-to-noise ratios resulting in an increase in dose to the patient.[5–7] The next significant advancement in CT is anticipated to be the advent of energy-resolved, photon-counting detectors (PCDs).[8] According to various studies previously performed with PCDs, CT has shown to have multiple advantages. This includes remarkably high spatial resolution, reduced electronic noise and beam-hardening and metal artifacts, increased contrast-to-noise ratio and radiation dose efficiency, and multienergy data acquisition while maintaining diagnostic quality.[9,10]

Energy-Integrating Detectors

Standard CT detectors use a two-step indirect conversion technique. First, a scintillator converts the X-ray energy into visible light. Second, visible light is collected and converted into an electrical charge using a photodiode (PD). ►**Fig. 1** shows the construction of an indirect CT array detector.[11,12]

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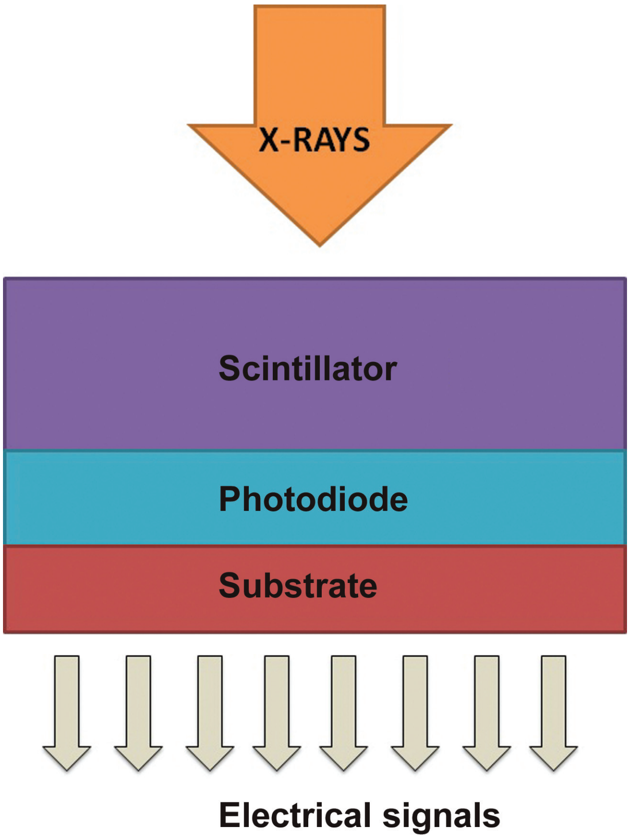

Fig. 1 Diagram depicting energy-integrating scintillator detector.

Scintillators are commonly organized in two-dimensional arrays in current CT scanners, which include a multislice CT geometry. A reflecting material matrix forms part of the scintillator array's construction. The PDA is physically supported by the reflecting matrix, which reduces cross-talk between PDA units. A separate low-noise preamplifier is attached to each element of the PDA. This PDA collects light signals from the scintillator array and converts them to electrical impulses. PDA electrical signals are then collected and converted to digital signals by the integrated acquisition electronics. The signal is digitalized and delivered to the image reconstruction module after being integrated over a period of time.[13]

The individual detector cells of an EID are separated by optically opaque layers. This is done to avoid optical cross-talk. They have a minimum width of ~0.1 mm. They also lower the detector's geometric dosage efficiency. The X-ray photons absorbed in the separation layers do not contribute to the measured signal, despite having passed through the patient.[14] Moreover, electronic noise significantly influences EID images in low-dose settings (up to a 5.8% increase).[15] In addition to this, another disadvantage comes with spectral CT. A tube potential of at least 120 kVp is required for adequate spectrum separation. As a result, lowering the radiation dosage for young individuals depends on reducing the tube current. Increased kVp, ultimately, reduces the contrast.[16] Rapid voltage switching using a single X-ray tube degrades average image quality, limits dual-energy spectrum contrast, and inhibits dose reduction due to lack of tube current modulation.[17] The sensitivity to optical photons is lower in dual-layer detectors with single tubes. There is more cross-talk between the two detector layers, resulting in greater noise levels in material decomposition imaging.[18]

Photon-Counting Detectors

EIDs are used in currently available multidetector CT scanners, wherein the total energy deposited by all photons is proportional to the measured signal without precise information about any individual photon or its energy. PCDs use direct-conversion techniques to count individual photons while measuring energy information simultaneously.[6] These detectors, which are based on arrays of pixelated cadmium telluride and cadmium zinc telluride, are electrically connected with application-specific integrated circuits for reading purposes. These will perform photon-counting that is quick and highly efficient.[19–21]

Photon-Counting Detectors—Basic Principles

During direct conversion, there is a production of electron-hole pairs within the semiconductor due to the absorption of X-rays. Between the cathode situated at the top and the pixelated anode electrodes situated at the bottom of the detector, the charges are separated by a strong electric field (►**Fig. 2**).

The electrons migrate to the anodes, where they cause small current pulses lasting a few nanoseconds (109 seconds). The current pulses are converted into voltage pulses using a full width at half maximum of 10 to 15 nanoseconds in a pulse-shaping circuit. The pulse height of the voltage pulses is translated from the quantity of charge in the current pulses. As a result, the pulse height is proportional to the absorbed X-ray photons' energy E. When these pulses reach a certain threshold, they are individually measured. Typically, the minimum threshold energies are more than 20 keV.[11]

The threshold is set so that the pulses are greater than the electronic noise but lesser than the pulses produced by the incoming photons. Additionally, the detector may categorize incoming photons into several energy bins (usually two to eight) based on their energy by comparing each pulse to numerous threshold values. As a result, the pulse counts are nearly free of electronic noise. On the contrary, the total energy deposited throughout the measurement interval, the electronic noise included, is measured and integrated using EIDs.[21]

Advantages and Applications of Photon-Counting Detectors

Compared with current detectors, one of the critical advantages of the PCD is the higher spatial resolution. PCDs developed for clinical CT feature pixel sizes that are smaller than today's typical EIDs, which have pixel sizes of the order of 1 mm. Hence, it offers a higher spatial resolution.[22,23] There is no requirement for extra separation layers between the individual detector cells since they are defined by the strong electric field between the electrodes. To significantly enhance spatial resolution, each "macro" detector pixel

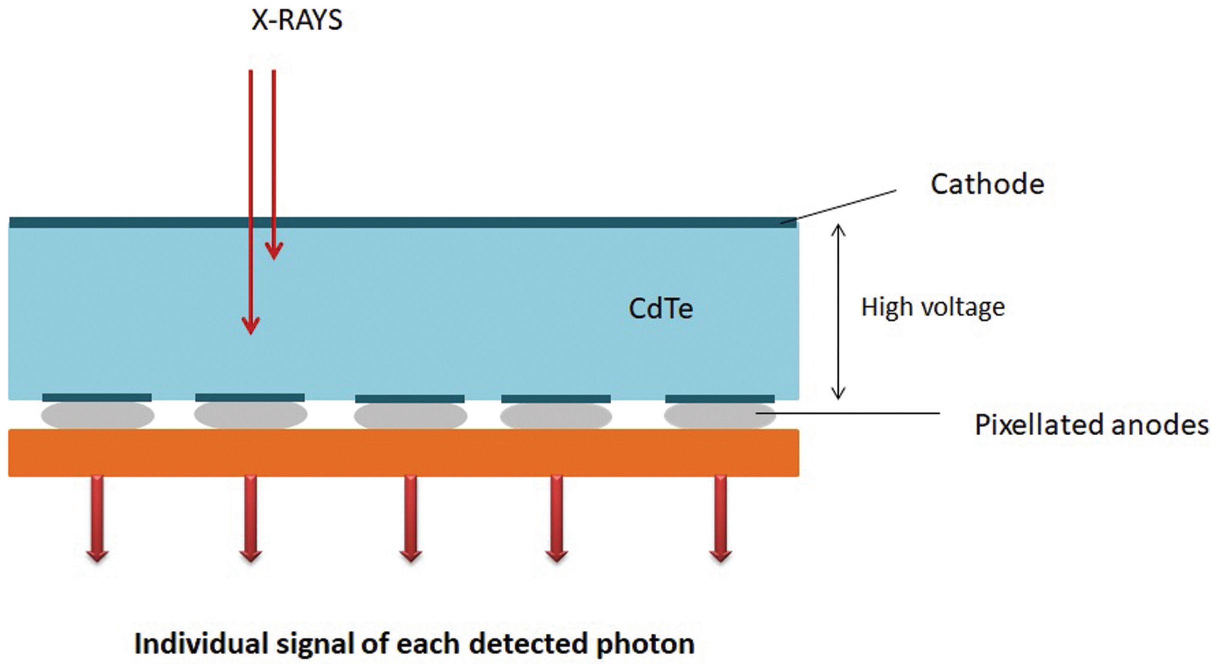

Fig. 2 Direct conversion photon-counting detector. CdTe, cadmium telluride.

bounded by collimator blades can be split into smaller detector subpixels that are read out individually.[12] Increasing spatial resolution further might assist some applications, such as lung and temporal bone imaging.[24,25]

When compared with solid-state scintillation detectors, there is the removal of electronic noise in PCD, which results in reduced noise in the image, reduction in streak artifacts, and stability of CT numbers in scanning with low dose or bariatric imaging.[11]

Photon weighting: Different weights can be assigned to photons of different energies in the processes of photon detection and image generation, impacting the signal's contrast as well as noise. The transmitted X-ray photons' energy is influenced by the contrast between two projection measurements. Low-energy photons, on average, contain more contrast information than high-energy photons. In addition, if the material composition of two projection measurements differs, the dependence of contrast on photon energy is enhanced, and higher contrast can be produced by providing more weight to photons that contain more contrast information. By giving specific photons more weight than others, the variance is increased compared with the mean value, therefore lowering the signal-to-noise ratio. Photon weighting, in many instances, is done indirectly through the material decomposition process. This process can be thought of as using the contrast and noise of the signal to estimate the thickness of the material with minimal variance.[26] Because tissue attenuation in the low-energy region of the X-ray spectrum is less homogenous, the higher weighting of low-energy photons results in an increase in beam-hardening artifacts. This problem can be corrected by material decomposition.[27]

Low-energy photons are attenuated more by the photoelectric effect, whereas high-energy photons are attenuated mainly through the Compton scattering effect. As a result of higher photon attenuation in the low-energy spectrum due to the photoelectric effect, materials with high atomic numbers like iodine contribute to improved contrast on CT scans. The detector signal in a typical EID CT system is proportional to the total energy of all collected X-ray photons. As a result, lower-energy photons contribute very little to the detector signal than higher-energy photons from high-Z materials like iodine, which have less information. Hence, in an EID CT system, the signal produced by lower-energy photons is underweighted, lowering the contrast-to-noise ratio of the iodine signal. PCD CT systems count each individual photon uniformly regardless of the detected photon energy. As a result, low-energy photons offer more image contrast in PCD CT than in EID CT, enhancing image contrast and contrast-to-noise ratio of iodine contrast material. If the radiation dosage is maintained, the increased iodine contrast on a PCD CT system results in a better iodine contrast-to-noise ratio.[4]

Spectral CT, sometimes also referred to as DECT or dual-energy CT, refers to CT that makes use of two-photon spectra.[28] At the energy levels utilized in diagnostic imaging, X-ray photons interact with matter primarily through photoelectric effect and Compton scatter. The photoelectric effect is the process wherein an incident photon causes the ejection of an electron from an atom's K shell, causing a void in the K shell. The void is filled by an electron from a neighboring shell, resulting in the release of energy in the form of a photoelectron. When an incoming photon has adequate energy to overcome an electron's K-shell binding energy, the photoelectric effect occurs.[29] The photoelectric effect is energy-dependent, that is, it is dependent on the atomic number of an element to a great extent. The probability of occurrence of photoelectric interactions increases as the atomic number of an element increases. The photoelectric effect is the primary process behind the energy-dependent or "spectral" variations in attenuation observed with DECT. Therefore, higher-atomic number materials are frequently described as having significant "spectral" features.[30] Compton scatter affects organic compounds with a low atomic number, whereas the photoelectric effect affects organic molecules with a higher atomic number.[31] The so-called dual-energy technique decomposes the linear attenuation coefficient into elements from the photoelectric and Compton effects. To distinguish the photoelectric- and Compton effects in the projection domain, two spectrally different measurements of the same X-ray path through the object of interest must be taken.[32] One way to determine materials in a structure using a PCD is using a rigorous basis decomposition technique. This technique, however, necessitates precise calibration of the imaging system's basic functions and energy response. On the other hand, reconstructed data from each energy window can be used in conjunction with a classification algorithm to divide the structures into different materials or regions. PCDs can discriminate between different types of materials, for example, soft tissue, bone, and contrast agents having K-edges in the energy range of interest.[33]

Another advantage of photon-counting imaging is its capability of low-dose imaging. There are two reasons for this. The first reason is the average image quality enhancement that PCDs provide. This involves increasing contrast-to-noise ratio (CNR) by using the energy information in the detected X-ray beam and improving small object visualization. PCDs have a distinct benefit when imaging with low dosage levels, that is, their ability to decrease electronic noise. PCDs employ a threshold to distinguish true counts from noise, and electronic noise may be eliminated by setting the threshold levels high enough above the noise.[15,27,34]

Discussion

PCD helps attain increased contrast among soft tissues due to a greater weighting of low-energy photons. This aids in the identification of small attenuation changes, as well as other intracranial conditions such as hemorrhage, demyelinating diseases, and tumors.[35–37]

One of the most well-established clinical uses of dual-energy CT is the characterization of renal stones. PCD-CT can be utilized for stone characterization since one of its potential advantages is the ability to add spectral information to any CT examination. A study performed by Ferrero et al showed that it was possible to accomplish a complete separation between uric acid and nonuric acid stones.[38] Another study depicted that PCD is capable of differentiating stones comprising of uric acid, calcium oxalate monohydrate, cystine, and apatite.[39]

Multiple contrast agents might be represented simultaneously, in vivo, using photon-counting spectral CT. Tissue enhancement at various phases can be detected in a single CT scan, thereby possibly eliminating the requirement for multiphase CT scans and ultimately lowering radiation exposure.[40]

Photon-counting has a greater detectability for low contrast bleeds, microbleeds, and sulcus when compared with multi-detector CT, indicating that it may be useful in identifying preexisting hemorrhages.[41]

In a study performed by Mannil et al, they aimed to compare the imaging features of coronary stents utilizing the new PCD to a traditional EID on CT. They observed that when compared with traditional EID arrays, the PCD delivers improved in-stent lumen delineation of coronary artery stents at similar CT scan protocol settings and the same image reconstruction parameters.[42]

One study aimed to determine the clinical feasibility, image quality, and radiation dose, which were achieved by halving the size of the PCD compared with standard-resolution photon-counting CT. It showed that, compared with standard-resolution photon-counting, halving the size of PCD demonstrated enhanced image quality with respect to image noise and spatial resolution.[43] Similarly, another study aimed to see if a high-resolution PCD system might help visualize minute features in the lungs better than a traditional high-resolution CT system. They found that compared with conventional chest CT, PCDs helped see higher-order bronchi and bronchial walls without affecting nodule assessment.[44]

For the identification of microcalcifications in breast tissues, Wetzl et al compared a spiral breast CT system to a digital breast tomosynthesis system. A cadmium telluride PCD was installed in the SBCT scanner, which aided in detecting microcalcifications, thereby increasing diagnostic rates.[45]

Conclusion

Photon-counting CT is a promising method that is on the approach of being practically viable. This may lead to modifying CT's clinical utility in the upcoming years drastically. With its high spatial resolution, reduced noise and artifacts, increased CNR, and providing multienergy data acquisition while maintaining diagnostic quality images, photon-counting CT systems are likely to replace EIDs as technology advances.

Description

This article reports the current status, basic working principles, and applications of photon-counting detectors (PCD) compared with the conventional EID used for CT scanners.

Conflict of Interest

None declared.

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