International Journal of Community Medicine and Public Health
Singh P et al. Int J Community Med Public Health. 2024 Aug;11(8):3287-3294
http://www.ijcmph.com
pISSN 2394-6032 | eISSN 2394-6040

DOI: https://dx.doi.org/10.18203/2394-6040.ijcmph20242192

Review Article

Technical challenges and benefits of photon counting detector computed tomography

Priyanka Singh*, Mamta Verma

Department of Radiological Imaging Techniques, College of Paramedical Sciences, Teerthanker Mahaveer University, Moradabad, Uttar Pradesh, India

Received: 24 May 2024
Revised: 03 July 2024
Accepted: 04 July 2024

***Correspondence:**
Priyanka Singh,
E-mail: priyankasingh.nov.26@gmail.com

ABSTRACT

X-ray detector is the essential part of a computed tomography (CT) system that controls dose efficiency and image quality. All clinical CT scanners utilized scintillating detectors up until the first clinical photon-counting-detector (PCD) system was approved in 2021. These detectors do not record information about individual photons during the two-step detection process. PCDs, on the other hand, employ a single step process that transforms X-ray radiation straight into an electrical signal. This retains information on individual photons, allowing one to count the quantity of X-rays in various energy ranges. Better spatial resolution, reduced dosages of iodinated contrast material, enhanced iodine signal, enhanced radiation dose efficiency, and the lack of electronic noise are the main benefits of PCDs. PCDs with multiple energy thresholds are able to separate detected photons into two or more energy bins, allowing for the availability of energy-resolved data for each record. In the event of dual-source CT, this permits high pitch or high temporal resolution acquisitions in addition to high spatial resolution for tasks involving material quantification or classification. PCDCT imaging of anatomy, where excellent spatial resolution provides clinical benefit, is one of the most promising uses of the technology. Inner ear, bone, small blood artery, heart, and lung imaging are among them. Photon-counting CT will become the wave of the future for workhorse CT imaging systems. An overview of the PCDCT principle, possible clinical benefits, and limitations of conventional CT is provided in this review paper, along with potential future developments for this CT imaging technology.

Keywords: Vascular disease, PCDCT, CCTA, EID, Dual energy CT, Cadmium telluride, CNR

INTRODUCTION

Photon counting computed tomography (CT) is a trending imaging technology which has been adapted after so many years of research development. The main application of photon-counting detector computed tomography (PCDCT) is the vascular disease of the human body. In CT, PCDCT is nearly new technology that has the potential to be the next significant technological advancement in the field. In a nutshell, energy-resolving detectors are used in photon-counting CT to enable scanning at many energies, because PCDCT reduces artefacts and has a higher spatial and contrast resolution of soft tissues than traditional computed tomography (also known as energy-integrating detector CT, or EID CT), it performs better in basic terms. Any pathological condition that affects the blood vessels is vascular disease. The primary reason of vascular abnormality is narrowing or hardening of arteries.[1]

CT is a non-invasive technique. It has wide focal view, fast scanning speed, excellent spatial resolution, but there are some limitations of CT as it uses ionizing radiation and complete CCTA iodinated administered contrast media.

Photon-counting detector computed tomography

Photon-counting detector computed tomography (PCDCT) is advance technology to identify any vascular disease with minimal radiation dose. The reduction or elimination of a number of parameters influencing detector performance and, consequently, the final quality of reconstructed pictures has allowed PCDs to be used in clinical CT. The primary barriers to the adoption of counting mode CdTe detectors in clinical CT was count rate performance.[2] Commercial scanners are capable of producing photon fluences of more than 108 photons/(mm^2 s), which is several orders of magnitude higher than hit rates typically found in applications such as nuclear medicine. In clinical PCDs, monolithic cadmium telluride (CdTe) layers linked to pixelated application-specific integrated circuits (ASIC) with modest pitch (<200 m) can alleviate the need for high hit-rate-capable detectors (>106 counts per second) (Figure 1).[3]

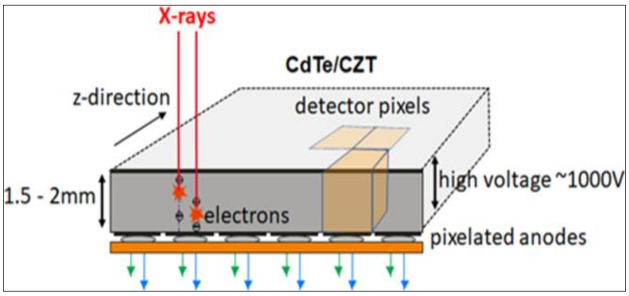

Figure 1: Technical development of photon counting computed tomography.

PRINCIPLE

Counting photons CT detectors measure photon energy and count the number of entering photons. Improved contrast-to-noise ratio, better spatial resolution, and optimal spectrum imaging are the outcomes of this approach. Hounsfield's invented computed tomography converts the visible light into electric signals through two step detection process, while PCD CT transforms the energy in a single step. Using exploratory PCD-CT devices, an innovative one-step direct X-ray conversion method that uses energy-resolving, PCDs has been thoroughly examined and early therapeutic benefits have been observed.[4]

Current solid-state scintillation detector properties

It is useful to quickly review the characteristics of solid-state scintillation detectors, which are the basis for all modern medical CT scanners, in order to better understand the features of photon-counting CT detectors and their clinical implications. Individual detector cells, measuring 0.8–1 mm in side length, comprise a scintillator (such as gadolinium oxide or gadolinium-oxy sulfide GOS) with a photodiode glued to the rear, generating solid-state scintillation detectors (Figure 2).[5] To comprehend the characteristics of the scintillator uses the absorbed X-rays to produce visible light, which is then detected by the photodiode and transformed into an electrical current. The energy E of the absorbed x-ray photon determines both the scintillation light's intensity and the induced current pulse's amplitude. Every current pulse that was recorded during one reading (projection) of the measurement is integrated. As a result, detector signal S is produced.

$$S = \int 0^{Emax} D(E)N(E)dE \approx \int 0^{Emax} EN(E)dE$$

In the equation, D(E)=detector responsivity, N(E)=absorbed X-ray flux during one reading, and Emax is the maximum energy of the X-ray beam (Figure 3).

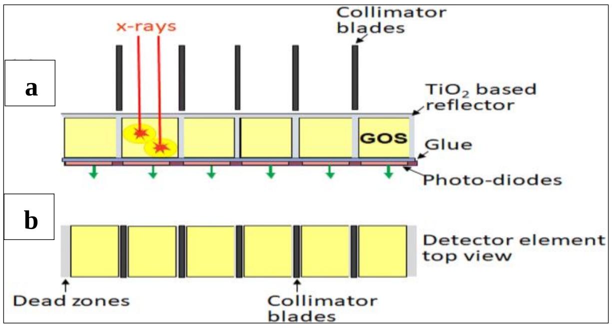

Figure 2: An energy-integrating scintillation detector schematic drawing, (a) side view, and (b) top view.

The X-rays (red arrows) are absorbed by individual detector cells composed of a scintillator such as gadolinium oxide or gadolinium oxysulfide (GOS), which transforms their energy into visible light. Photodiodes affixed to the rear of every detector cell pick up this light and transform it into an electrical current. It takes collimator blades to reduce stray radiation. In order to avoid optical crosstalk, the individual detector cells must also be separated by optically opaque layers (such as those made of TiO_2). These layers are known as "dead zones" since X-rays absorbed do not contribute to the measured signal.

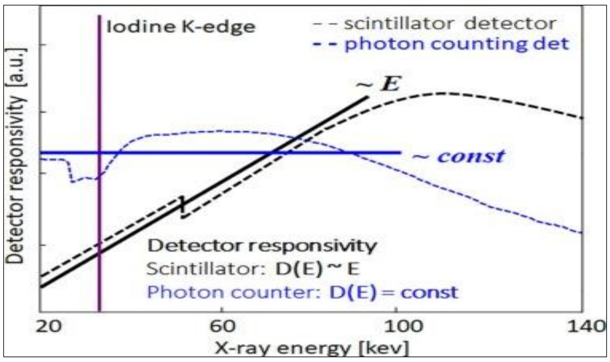

Figure 3: The response of a GOS scintillator detector (dotted black line) and a CdTe photon-counting detector (dotted blue line) to X-ray energy E is represented by the detector responsively D (E).

A vertical line indicates iodine K-edge at 33 keV. When compared to a PCD, a scintillator detector's signal is less affected by low-energy X-rays that are directly above the K-edge of iodine. As a result, a scintillator detector produces an image with less iodine contrast than a PCD at the same X-ray tube voltage.

Compared to X-ray photons with higher energy, those with lower energy E, which carry the majority of the low contrast information, contribute less to the integrated detector signal. As a result of the energy-weighting, the contrast-to-noise ratio (CNR) in CT pictures is decreased. This is especially true for CT scans that use an iodinated contrast agent, as iodine's x-ray absorption is best at lower energies that are just above its K-edge at 33 keV.[6]

Electronic noise distorts the low-level analogue electric signal of the photodiodes at low X-ray flux, this noise becomes larger than the quantum noise (Poisson noise) of the X-ray photons and, if the flux is further reduced, causes a disproportional increase in image noise and instability of low CT-numbers (e.g., in the lungs). Potentially further reducing the radiation dosage in medical CT is limited by this significant rise in noise and the drift of CT-numbers.[7]

To avoid optical crosstalk, layers that are optically opaque are placed between the individual detector cells. Their minimum width is around 0.1 mm, and they lower the detector's geometric dosage efficiency: even if they have gone through the patient, X-ray photons absorbed in the separation layers do not add to the recorded signal; hence, from the standpoint of radiation dosage, they constitute a wasted dose.[8] Detector cells in medical CT detectors today range in active size from 0.8×0.8 mm^2 to 1×1 mm^2.

With an extra 0.2 mm dead zone in the in-plane direction and 0.1 mm dead zone in the z-axis direction, a detector element's total area ranges from 1.0×0.9 mm^2 to 1.2×1.1 mm^2. Thus, the geometric dosage efficiency is between 70 and 80 percent. It is therefore problematic to increase the spatial resolution of solid-state scintillation detectors beyond today's performance levels. Notably, significantly reducing the scintillators' size to increase spatial resolution while maintaining the width of the separation layers constant will further reduce the geometric efficiency.

Photon counting detector's properties

Photon-counting detectors can be made with semiconductors such as silicon (Si), cadmium telluride (CdTe), or cadmium zinc telluride (CZT). Because of their high atomic number, the CdTe and CZT layers can be relatively thin (1-2 mm) and still provide outstanding X-ray absorption. For Si photon-counting detectors to effectively absorb photons, they need to be significantly thicker (30–60 mm). According to published research, every PCDCT used in preclinical or clinical settings has either a CdTe or a CZT detector. On the upper side of the semiconductor layer, large-area cathode electrodes are created, while pixelated anode electrodes are deposited on the lower side. Applying a high voltage of 800–1000 V between the cathode and each individual anode will create a strong electric field. As the incident X-rays are absorbed by the semiconductor, charges (electron-hole pairs) are separated in this electric field. Once at the anodes, the electrons produce short current pulses (10−9 seconds), which are then converted by an electronic pulse shaping circuit into voltage pulses with a half-width (FWHM) of 10–15 nanoseconds. During the measurement time of the projection, the voltage pulse height is directly proportional to the absorbed energy E of the X-rays.[9]

Comparing photon-counting detectors to solid-state scintillation detectors reveals a number of advantages. There are no extra separation layers; instead, the strong electric field between the pixelated anodes and common cathode defines each individual detector cell (Figure 4). The only thing that lowers a photon-counting detector's geometrical dosage efficiency is the inevitable presence of anti-scatter collimator grids or blades.[10]

Unlike scintillator detectors, each "macro pixel" enclosed by collimator blades can be separated into smaller sub-pixels for independent readings, hence enhancing spatial resolution.

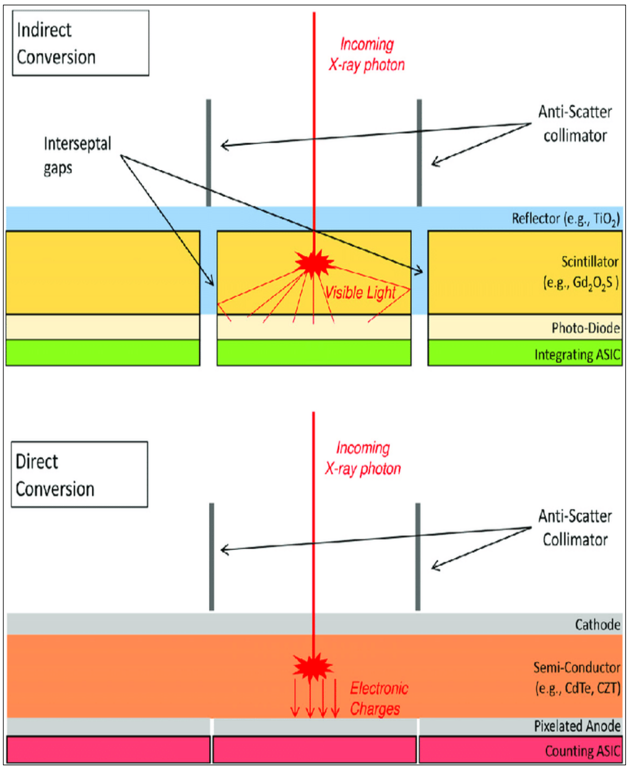

Figure 4: Schematic diagram of photon counting detector.

COMPARISON BETWEEN CONVENTIONAL CT/EID AND PCD CT

Traditional EIDs release light and produce a signal that is proportionate to the total energy of all X-rays they detect. Given that lower energy X-rays are more likely to be attenuated by iodine and other physiological tissues, PCDs uniformly weigh detected X-rays of different energies, giving greater signal to lower energy photons that may contribute to significant portions of a CT image. The electrical signal that each X-ray deposits is proportionate to its energy, which gives PCDs the ability to discriminate between different energies.[11]

Table 1: The physical and working comparison between EID and PCDCT.

Variables	Conventional CT/EID	PCD CT
Energy conversion process	Indirect conversion	Direct conversion
Artefacts	Produce different artefacts	Reduce artefacts
Performance on vascular disease	Average	Better than EID
Detectors	Scintillator detectors made up of $Gd_2O_2S_2$/ $CdWO_4$	Silicon, CdTe/CZT
Arrangement of detector	Arranged in 8-16 rows	Arranged in 16-320 rows
Applied voltage	120Kv	800-1000 v
Radiation exposure	Higher radiation dose	Lower radiation dose
Reconstruction process	Filtered back	Quantum iterative
Image resolution	Poor resolution acquisition	Ultra high resolution acquisition

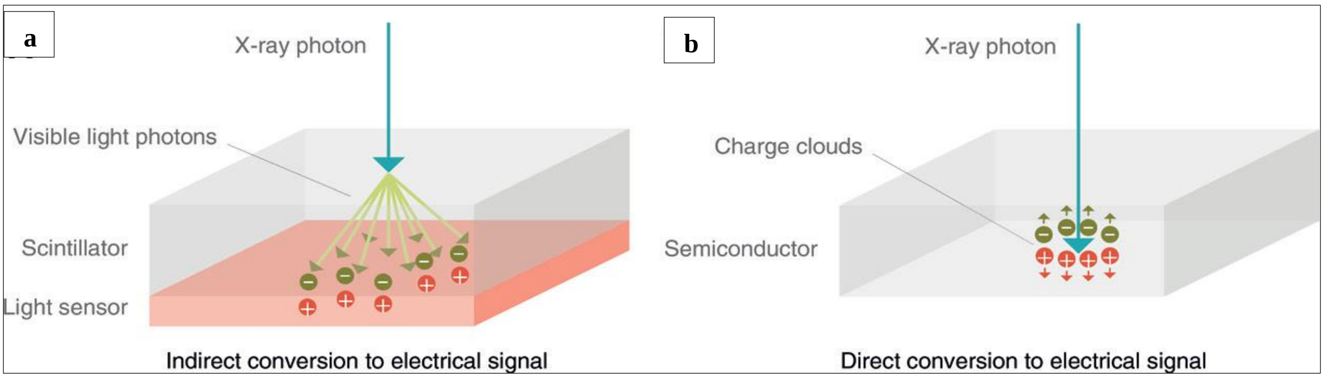

Figure 5: Detector kinds are shown in diagrams.

(a) An incident X-ray photon in a typical energy-integrating detector is transformed into a shower of visible light photons in a scintillator. When visible light strikes a light sensor beneath it, positive and negative electrical charges are produced; and (b) The X-ray photon is absorbed in a semiconductor material in a photon-counting detector, where it produces charge, both positive and negative. An electrical signal is produced when the positive and negative charges are drawn in opposing directions by a strong electric field.

ENERGY INTEGRATING DETECTOR (EID)

Energy-integrating detectors (EIDs) that allow for proportionate detection of the total energy delivered by all photons without presenting precise information about a single photon or its energy are used in the construction of multi detector CT scanners that are sold professionally.

The incoming X-rays are converted into visible light inside a ceramic scintillator of an energy integrating detector. After that, a photo diode gathers the produced visible light, and the signal is progressively integrated for the duration of the single picture acquisition (about 100 to 300 µs). The detector's individual pixels must be forcibly separated due to the noteworthy spatial dispersion of visible light inside the scintillator (the light must propagate extremely well to emit to the photodiode) (Figure 5a). To do this, the ceramic is cut, the gaps are stuffed with reflecting material, and light is only permitted to exit the ceramic in the path of the photo diode. Because the area between the pixels is lost during the conversion process and is therefore unusable for measuring X-rays, the pixels cannot be arbitrarily reduced in size. Furthermore, spectral information of the individual X-ray photons is lost in this technique because all generated light is collected inside the temporal frame of a single image. The total amount of light released from the scintillator cannot distinguish between a large number of high energy photons and a small number of low energy photons, despite the fact that low energy photons contribute more to the contrast of the image (Figure 4).[12]

PHOTON COUNTING DETECTOR COMPUTED TOMOGRAPHY (PCDCT)

A novel detector is PCD-CT's main component. A CT system's detector captures the X-rays that the patient has attenuated. With its 16–320 rows and roughly 1000 individual detector pixels, it has the appearance of a quick digital X-ray camera.

Silicon (Si), CdTe, or CZT are examples of semiconductors used to make photon-counting detectors. The CdTe and CZT layers can be relatively thin (1–2 mm) and still produce excellent x-ray absorption because of their high atomic number. Si photon-counting detectors must be substantially thicker (30–60 mm) in order to absorb photons adequately. The literature currently in publication states that CdTe or CZT detectors are present in every PCD-CT used in preclinical or clinical settings. Large-area cathode electrodes are formed on the upper side of the semiconductor layer, whereas pixelated anode electrodes are deposited on the lower side (Figure 3). An intense electric field can be produced by applying a high voltage of 800–1000 V between the cathode and each individual anode. Charges (electron-hole pairs) that are separated in this electric field are produced by the semiconductor's absorption of the incident X-rays. Once at the anodes, the electrons cause brief current pulses (10−9 seconds) that are transformed into voltage pulses with a half-width (FWHM) of 10–15 nanoseconds by an electronic pulse shaping circuit. The voltage pulse height is directly proportional to the absorbed energy E of the X-rays during the projection's measurement time (Figure 6).[13,14]

TECHNICAL CHALLENGES OF PCDCT

The reduction or elimination of a number of parameters influencing detector performance and, consequently, the final quality of reconstructed pictures has allowed PCDs to be use in clinical CT. Commercial scanners have the potential to have photon fluences as high as 108 photons/(mm^2 s).[15] When compared to hit rates often found in, say, nuclear medicine applications, is several orders of magnitudes. In clinical PCDs, monolithic CdTe layers linked to pixelated, small-pitch application-specific integrated circuits (ASICs) can alleviate the need for high hit-rate-capable detectors (>106 counts per second).

By decreasing the active area per pixel, as well as the corresponding count rate needed, pulse pileup, count loss, and spectral distortion, this arrangement produces a very high spatial resolution when the detector is run without rebinning (i.e., 1×1 reading).

However, the charge sharing effect occurs when multiple nearby pixels feel the same secondary charge clouds due to a tighter detector pitch. A number of approaches, including winner-take-all circuits where the pixel with the most charge is allocated to the total charge detected in a 2×2 neighborhood, which can be used to rectify charge sharing. This tactic is used, for example, by the Medipix3 ASIC. In addition to secondary charge sharing, there are various forms of pixel crosstalk that can occur in PCDs. For example, pixel crosstalk resulting from the initial interaction's producing fluorescence X-ray radiation can be observed in adjacent pixels. The lower limit of detector pixel size in real-world applications is caused by this phenomenon.[16]

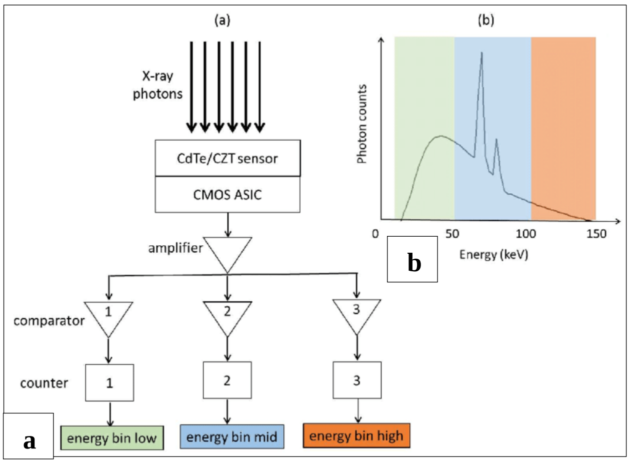

Figure 6: (a) A photon counting detector; and (b) the signals that are detected are sorted into various bins based on the photon energy.

Table 2: The distinctive characteristics of photon-counting detectors and their influence on CT images in clinical settings.

S. no.	Photon counting detector CT	Effects on medical pictures
1	X-ray coincidence directly converted to a signal proportional to photon energy	Lack of down weighting of lower energy photons results in an increased iodine signal. The voltage of an X-ray tube can be used to get multi-energy information. Virtual monoenergetic photos, non-contrast, non-calcium, and iodine maps are frequently accessible
2	Reduced pixel size in the detector	Enhanced spatial resolution
3	The lack of reflective septae for every detector element in traditional energy-integrated detectors leads to geometric dose inefficiency.	Reduced radiation dose and enhanced spatial resolution
4	2 and 3	Radiation dosage reduction attenuating filters are not necessary for ultra-high spatial resolution. There is no longer a radiation dose penalty for ultra-high spatial resolution, and it can be used in broader body regions.
5	The removal of electronic noise	There is only quantum noise.
6	X-ray beam shaping using tube potential selection, energy thresholds, and tin filters	Decrease in metal and blooming artifact - Decrease in radiation exposure

BENEFITS

Similar to spectral CT, photon-counting CT can distinguish between various tissues and contrast materials with ease. Similar to dual-energy CT, photon-counting CT may in the future provide greater virtual non-contrast imaging, better spatial resolution, higher signal-to-noise ratio, and spectral imaging data. It might lessen the quantity of radiation that is exposed to, the quantity of contrast agent that is required, and the quantity of CT artefacts. Additionally, it might make the use of several contrast agents—such as iodine, gadolinium, or gold nanoparticles—for simultaneous imaging possible. Photon-counting CT, in contrast to traditional CT, can easily determine the precise concentration of components within the voxel (e.g., calcium, iodine).

Imaging at high resolution without sacrificing dosage

There is a trade-off between radiation exposure and image resolution with traditional CT scanners. Two steps are involved in the indirect conversion process used by traditional EIDs. First, incident X-ray photons are converted by a scintillator into visible light. This scintillation light is then captured and transformed into electrical pulses by a sequence of photodiodes.[17] The detector loses the ability to store the energy information of individual X-ray photons when it integrates the energy of all photons over a given time interval.

This does not impose a limit on a detector's spatial resolution. In EIDs, individual detector cells are partitioned by optically opaque layers known as septa in order to prevent optical crosstalk. Nevertheless, these septa produce inactive regions on the detector surface and impact the geometric dose efficiency because of their thinness.[18]

It's critical to remember that PCDs function differently. They work with photons directly, converting them into electrical charges instead of visible light. They do not need septa as a result. Because of this design advantage, PCDs can use a larger pixel density without sacrificing the detector's geometric dose efficiency.[19]

PCDs with no electrical noise

The capacity to quantify the energy of each individual photon is essential in a PCD. Generally, electronic noise is seen in the spectrum below 25 keV. Since real photons have an energy level beyond this cutoff, PCDs are able to discriminate between real photon signals and electronic noise. Because of this distinction, electrical noise in the signal may be completely eliminated, guaranteeing a better and more accurate depiction of the scanned item.[20]

Decrease in beam-hardening

Low-energy photons are more attenuated than high-energy photons in CT because of the energy dependence of mass attenuation coefficients and the utilization of poly energetic beams.[21] This results in a phenomenon called beam hardening, which causes the X-ray beam's mean energy to shift toward the higher end of the spectrum.

The use of high-energy thresholds in PCDs provides the best immunity to beam-hardening effects. Beam hardening from a very dense target (cortical bone and metal implants) can result in characteristic artifacts such as cupping artifacts and streaking (dark bands) artifacts. These artifacts affect the image appearance and CT number accuracy for nearby soft tissues. In PCDs, constant weighting reduces the beam-hardening artifacts.[22]

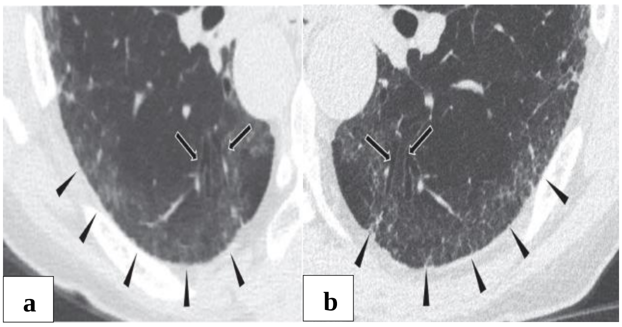

Figure 7: Using a clinical routine approach, patient is diagnosed with an idiopathic non-specific interstitial pneumonia, scanned on an exploratory PCD-CT and a conventional energy-integrating detector CT.

(a) and (b) in the right sub pleural right lower lobe, PCD-CT shows fine reticulations (arrowheads) in contrast to conventional CT, which seems to reveal ground glass opacities in this region (arrowheads). Compared to traditional CT, PCD-CT more clearly illustrates traction bronchiectasis (arrows).

Enhanced ratio of contrast to noise

By treating each photon's contribution more consistently, photon counting technology raises the contrast to noise (CNR). Low-energy photons are intrinsically down-weighted relative to high-energy photons in conventional CT. However, these low energy photons might provide important information about picture contrast, this bias has a negative effect on the CNR.[23]

In terms of contrast differentiation and image clarity, photon counting technology provides a notable advantage by guaranteeing a more consistent handling of all photons, irrespective of their energy. Quantum iterative reconstruction (QIR), which takes these benefits into account, improves imaging even further. Statistical optimization of spectral data is utilized in quantum iterative reconstruction to guarantee accurate artifact correction. Adaptive iterative regularization is used by QIR, in contrast to standard reconstruction techniques, to separate real information from noise. By taking use of the spatial and temporal coherence of spectral data gathering, this guarantees that all spectra are handled uniformly.[24,25]

DISCUSSION

PCDCT is an emerging imaging technology has been implemented. It is mostly used to treat the human's vascular diseases. It is the advance technology of computed, and could be the next big development in the discipline. PCDCT performs better than traditional computed tomography in terms of reducing artifacts and having a higher spatial and contrast resolution of soft tissues. This is made possible by the use of energy-resolving detectors, which allow scanning at many energies. With the least amount of radiation, photon counting CT is an advanced diagnostic tool for vascular diseases. PCDs have made it possible to employ CT in clinical settings by reducing or eliminating several factors that affect detector performance and, in turn, the ultimate quality of reconstructed images. Count rate performance was one of the main obstacles to the use of counting mode CdTe detectors in clinical CT. The introduction of iterative image reconstruction and reconstruction with deep convolutional neural networks will be useful via this article.

In this review article we compare the given dose and image quality of PCDCT and conventional CT. Relative to hit rates commonly observed in applications like radiation therapy, industrial scanners can produce photon dictates of more photons. This is an order of magnitude higher. ASICs with modest pitch which are connected to unitary CdTe layers may decrease the need of elevated hit-rate-capable detectors in clinical PCDs. The benefit of this approach involve improved CNR, better spatial resolution, and optimal the spectrum imaging. Computed Tomography converts visible light into electric signals through a two-step detection process, while PCDCT converts the energy in a single step.[26]

From the previous study for an improved awareness of the characteristics of photon-counting CT detectors and its clinical penalties, it is helpful to quickly examine the features of solid-state scintillation detectors, which form the basis of all modern medical CT scanners. Each detector cell, comprises of a scintillator of gadolinium oxide or gadolinium-oxy sulfide (GOS) and a photodiode glued to the backside to offer solid-state scintillation detectors.[27] Most of the low contrast information is transported by X-ray photons with lower energy which contribute less to the as a whole detector signal than do photons with greater energy. The energy-weighting causes a drop in the CNR in CT images. While the X-ray flux is low, electronic noise distorts the low-level analogue electric signal of the photodiodes. This noise grows in magnitude over the quantum noise of the x-ray photons and, if the flux is further dipped, results in an inappropriate increase in image noise and instability of low CT-numbers. A crucial component of the photon counting detectors we use today is cadmium telluride, or CdTe. CdTe is Zincblende, Sphalerite structured and crystallizes in the cubic $\bar{F}43m$ space group. We employ high density CdTe crystals as a compound semiconductor in our photon counting technique, which effectively absorbs light and transforms it into an electronic signal. The greater accuracy of CdTe detectors over typical X-ray detectors, which rely on scintillators for converting X-ray radiation to light, comes from CdTe's direct conversion of radiation into an electrical signal. In this paper we discussed about the crystalized structure of CdTe.[28]

Conventional EIDs flash light in proportion to the energy of every X-ray they detect, and they additionally generate a signal. Since iodine and other physiological tissues have the tendency to attenuate lower energy X-rays, PCDs equally weigh detected X-rays of different energies, providing higher signal to lower energy photons that may contribute significantly to parts of a CT image. PCDs can distinguish between various energies because each X-ray deposits an electrical signal that is commensurate to its energy.[29]

CONCLUSION

Photon-counting CT will become the wave of the future for workhorse CT imaging systems. These systems promise to help reduce dose, improve image quality and offer spectral imaging built into every exam. This enables views of datasets at any energy level, material decomposition imaging so things like calcium and iodine can be mapped and faded in or out of images to show perfusion defects and the ability to reduce or eliminate calcium blooming and see inside calcified arteries.

The creation of PCDs with energy discrimination capabilities has been made possible for medical X-ray CT and X-ray imaging. These PCDs have the potential to both boost the current CT and X-ray images, such as dose reduction, and open up revolutionary new applications, such as molecular CT and X-ray imaging, by measuring the material information of the object to be imaged and using detection mechanisms that are completely different from the current energy integrating detectors. Nevertheless, PCD performance is not flawless, and creating PCDs with almost ideal properties appears to be very challenging.

PCDs provide several advantages, including routine multi-energy imaging, increased dosage efficiency, electronic noise reduction, stronger contrast-to-noise ratio, and superior spatial resolution.

Funding: No funding sources
Conflict of interest: None declared
Ethical approval: Not required

REFERENCES

1. Rajiah P. Updates in Vascular Computed Tomography. Radiol Clin N Am. 2020;58:671-91.
2. Taguchi K, Iwanczyk JS. Vision 20/20: Single photon counting X-ray detectors in medical imaging. Med Phys. 2013;40:100901.

3. Willemink MJ, Persson M, Pourmorteza A, Pelc NJ, Fleischmann D. Photon-counting CT: Technical Principles and Clinical Prospects. Radiology. 2018;289:293-312.
4. Kreisler B. Photon counting Detectors: Concept, technical Challenges, and clinical outlook. Eur J Radiol. 2022;149:110229.
5. Carturan SM, Marchi T, Fanchini E, De Vita R, Finocchiaro P, Pappalardo A. Scintillator and solid-state neutron detectors and their applications. Eur Phys J Plus. 2014;129(10):212.
6. Roessl E, Proksa R. K-edge imaging in x-ray computed tomography using multi-bin photon counting detectors. Phys Med Biol. 2007;52:4679-96.
7. Gutjahr R, Halaweish AF, Yu Z, Leng S, Yu L, Li Z, et al. Human Imaging With Photon Counting-Based Computed Tomography at Clinical Dose Levels: Contrast-to-Noise Ratio and Cadaver Studies. Invest Radiol. 2016;51:421-9.
8. Rajagopal JR, Farhadi F, Solomon J, Sahbaee P, Saboury B, Pritchard WF, et al. Comparison of low dose performance of photon-counting and energy integrating CT. Acad Radiol. 2021;28(12):1754-60.
9. Tanguay J, Cunningham IA. Cascaded systems analysis of charge sharing in cadmium telluride photon-counting x-ray detectors. Med Phys. 2018;45(5):1926-41.
10. Yu Z, Leng S, Kappler S, Hahn K, Li Z, Halaweish AF, et al. Noise performance of low-dose CT: comparison between an energy integrating detector and a photon counting detector using a whole-body research photon counting CT scanner. J Med Imaging. 2016;3(4):043503.
11. Yu Z, Leng S, Kappler S, Hahn K, Li Z, Halaweish AF, et al. Noise performance of low-dose CT: comparison between an energy integrating detector and a photon counting detector using a whole-body research photon counting CT scanner. J Med Imaging. 2016;3(4):043503.
12. Rajagopal JR, Farhadi F, Solomon J, Sahbaee P, Saboury B, Pritchard WF, et al. Comparison of low dose performance of photon-counting and energy integrating CT. Acad Radiol. 2021;28(12):1754-60.
13. Kreisler B. Photon counting Detectors: Concept, technical Challenges, and clinical outlook. Eur J Radiol. 2022;149:110229.
14. Willemink MJ, Persson M, Pourmorteza A, Pelc NJ, Fleischmann D. Photon-counting CT: technical principles and clinical prospects. Radiology. 2018;289(2):293-312.
15. Kreisler B. Photon counting Detectors: Concept, technical Challenges, and clinical outlook. Eur J Radiol. 2022;149:110229.
16. Rajendran K, Petersilka M, Henning A, Shanblatt ER, Schmidt B, Flohr TG, et al. First clinical photon-counting detector CT system: technical evaluation. Radiology. 2022;303(1):130-8.
17. Leng S, Rajendran K, Gong H, Zhou W, Halaweish AF, Henning A, et al. 150-μm spatial resolution using photon-counting detector computed tomography technology: technical performance and first patient images. Investig Radiol. 2018;53(11):655-62.
18. Symons R, Cork TE, Sahbaee P, Fuld MK, Kappler S, Folio LR, et al. Low-dose lung cancer screening with photon-counting CT: a feasibility study. Phys Med Biol. 2016;62(1):202.
19. Willemink MJ, Persson M, Pourmorteza A, Pelc NJ, Fleischmann D. Photon-counting CT: technical principles and clinical prospects. Radiology. 2018;289(2):293-312.
20. Taguchi K, Iwanczyk JS. Vision 20/20: single photon counting x-ray detectors in medical imaging. Med Phys. 2013;40(10):100901.
21. Brooks RA, Di Chiro G. Beam hardening in x-ray reconstructive tomography. Phys Med Biol. 1976;21(3):390.
22. Barrett JF, Keat N. Artifacts in CT: Recognition and avoidance. Radiographics. 2004;24:1679-91.
23. Silkwood JD, Matthews KL, Shikhaliev PM. Photon counting spectral breast CT: effect of adaptive filtration on CT numbers, noise, and contrast to noise ratio. Med Phys. 2013;40(5):051905.
24. Swank RK. Absorption and noise in x-ray phosphors. J Appl Phys. 1973;44:4199-203.
25. Giersch J, Niederlöhner D, Anton G. The influence of energy weighting on X-ray imaging quality. Nucl Instrum Method Phys Res Sect A: Accel Spectrometers Detect Assoc Equip. 2004;531(1-2):68-74.
26. Schlomka JP, Roessl E, Dorscheid R, Dill S, Martens G, Istel T, et al. Experimental feasibility of multi-energy photon-counting K-edge imaging in pre-clinical computed tomography. Phys Med Biol. 2008;53(15):4031-47.
27. Carturan SM, Marchi T, Fanchini E, De Vita R, Finocchiaro P, Pappalardo A. Scintillator and solid-state neutron detectors and their applications. Eur Phys J Plus. 2014;129(10):212.
28. Christensen NE, Christensen OB. Electronic structure of ZnTe and CdTe under pressure. Phys Rev B. 1986;33(7):4739.
29. Leng S, Rajendran K, Gong H, Zhou W, Halaweish AF, Henning A, et al. 150-μm spatial resolution using photon-counting detector computed tomography technology: technical performance and first patient images. Investig Radiol. 2018;53(11):655-62.

Cite this article as: Singh P, Verma M. Technical challenges and benefits of photon counting detector computed tomography. Int J Community Med Public Health 2024;11:3287-94.