Characteristics of the new Photon counting CT detector

Petra Mimica[1], **Tatjana Matijaš**[1]

[1] University of Split, University Department of Health Studies, Split, Croatia

Corresponding author: **Tatjana Matijaš**, University of Split, University Department of Health Studies, Split, Croatia, email: tmatijas@ozs.unist.hr
DOI: https://doi.org/10.55378/rv.47.2.3

Abstract

Radiology is one of the fastest growing branches of medicine, and precisely one of the devices that follows this development the fastest is the computerized tomography device. Scientists are most focused on research into improving the parameters of the X-ray tube and detector as the most important parts of this imaging modality. In addition to the detector with energy integration, which is widely used today as part of everyday clinical practice, the subject of research is also a Photon countng detector. The numerous virtues that stand out in the new type of detector are responsible for the general opinion that in the future this new technology will dominate CT devices in clinical practice, improving the acquisition of numerous diagnostic and interventional procedures. It is believed that this technology ushers radiology into a new era precisely with its improved spatial and contrast resolution, noise reduction and increase of contrast to noise ratio, but also with new possibilities such as simultaneous imaging of multiple contrast agents and multi-energy imaging. Improvements in the visibility of iodine contrast agent provide additional diagnostic possibilities and more accurate differentiation of benign and malignant suspected lesions. The significantly smaller pixel size on this detector offers an almost two-fold increase in spatial resolution, and thus an increase in radiation dose efficiency. The use of a detector with a photon counter is therefore also suitable for pediatric patients, and there is also a wide range of low-dose imaging options for the purpose of preventive cancer screening. Processes such as the material decomposition and spectral imaging are still in the research process, but have great potential to optimize future daily clinical practice. More work and published research results are needed, but surely this technology will one day make the work of radiologic technologists and radiologists easier.

Keywords: computed tomography; energy integrating detector; photon counting detector

Abbreviations and acronyms: CNR (Contrast-to-noise ratio), CT (Computed tomography), EID (Energy integrating detector), FDA (Food and drug administration), HU (Hounsfield unit), MSCT (Multi-slice computed tomography), PCD (Photon counting detector), SNR (Signal-to-noise ratio), UHR (Ultra hard resolution)

Introduction

Computed tomography (CT) is a radiological imaging modality that represents the greatest progress in radiology after the discovery of X-rays in 1895 [1, 2]. CT represents the first digital radiological method. The development of CT is not directly related to the discovery of X-rays, but to the development of computer techniques [3]. By definition, CT is an imaging technique that uses a collimated beam of X-ray radiation to obtain axial layers of the imaged object [1, 2]. Regardless of radiation dose concerns, CT imaging has become an "indispensable tool" in many branches of medicine [3]. Constant technological progress improves the diagnostic value of CT and there is an increasing number of indications for this examination [4]. This is supported by data from 2021 that over 80 million CT scans are performed annually in the USA alone, which makes CT the most important and widespread imaging modality for obtaining diagnostic information for the purpose of treating patients [5]. The most important characteristics of this device are high scanning speed, wide availability of the device and software aspects, for example high spatial resolution [6].

Historical overview of the development of the CT device

Following the history of science, it is evident that at the turn of the 19th and 20th centuries, revolutionary discoveries were made in medicine and physics. The combination of engineering and knowledge in medicine enabled the development of diagnostic imaging as we know today [7]. In 1972, the first commercial CT device was put into use (Figure 1), and in 1979, Godfrey Hounsfield and Alex Cormack won the Nobel Prize for the discovery and construction of this device. Later, the unit used in the interpretation of images created by CT technology was named after Godfrey Hounsfield. With the beginning of the use of CT devices, we also associate the beginning of the use of computers in radiology and the first steps in the gradual digitization that followed in the coming decades [1, 2]. The first constructed CT devices were intended only for scanning the head and brain because the examination took a long time, and for a good image the patient was required to be completely still. Devices of the second generation, which began to develop in 1975, enable imaging of the head and the rest of the body.

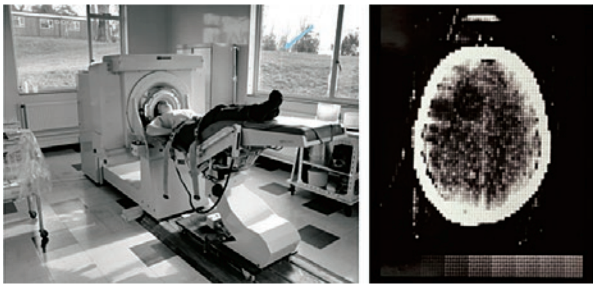

Figure 1. The first CT device and the first image taken with the CT device
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC8555965/pdf/JMI-008-052110.pdf

The development of CT took place in several stages, which are called generations of CT devices, and each generation differs from the previous one by the improved performance of the detector and the mutual relation and path of movement of the X-ray tube and the detector [2]. After the appereance of spiral CT technology, the division according to device generations was abandoned. Today, only spiral CT devices are used, which as a working principle have a continuous rotation of the X-ray tube and detector "banana" around the imaged object, and the table with the patient simultaneously moves linearly through the primary beam of radiation [1, 2]. The primary X-ray beam forms a spiral around the patient's body, so in that way it's scanning a large anatomical volume in a short exposure time [2]. Devices of the latest generation are called Multislice computed tomography (MSCT) [1, 2]. They have up to 320 rows of detectors in the detector banana, and each row contains several hundred (600-800) detector units. Then it became possible to scan a large volume of the body in one acquisition and shorten the examination time and radiation dose. Various image reconstructions are performed by processing raw data, and the thickness of the transverse layer can be as much as 0.5 mm, and high spatial resolution images are obtained with a very good view of fine structures [2]. Since its first use, i.e. since the introduction of CT devices into medicine, its importance and field of application have exceeded all expectations of the researchers responsible for this technology [3].

Working priciple of the CT device

In interaction with electrons, X-rays cause electrical and chemical reactions that are recognized on detectors as photons of visible light due to scintillation and contact with photodiodes [7]. The operating principle of the CT device is based on attenuation, i.e. the weakening of the X-rays passing through the imaged object. Attenuation occurs as a result of the passage of a useful beam of X-ray radiation through tissues of different atomic number, more precisely, of different density. Energy is absorbed and dissipated, and the differently attenuated X-rays fall on the radiation detector, which converts them into electrical signals proportional to the attenuation, and a CT image of different shades of gray scale is created, depending on the energy of the X-rays reaching the detector. The image is displayed on the screen as a matrix of pixels of different shades, and is created by data processing with complex mathematical algorithms.

Attenuation of X-rays can be quantitatively measured by Hounsfield units (HU) or so-called CT number [1, 2]. It is an agreed-upon scale that indicates the damping coefficient in the range from -1000 to +3000, where water is always 0, numbers in the negative part of the scale indicate tissues such as fat and air, and numbers in the positive part indicate fluids, parenchymal organs, bones and others structures [1, 2, 3]. With newer devices, the scale has been extended to values from -10,000 to +30,000, where the most negative value is the mark for air, and the most positive for bones and metals. The extended scale is used for a better display of implants, metals in the body or prostheses. CT devices work on the principle of recording one plane of the body, i.e. transverse or axial, and the display of the coronary and sagittal planes, and if necessary oblique projections, are created by reconstruction from the data obtained by recording axial layers. The contrast resolution of the CT device has been greatly improved, so when editing the image in postprocessing, it is possible to select different windows with different gray scale levels that improve the visibility of target tissues [1, 2].

The main parts of the CT device

The main parts of the device include the gantry, the table on which the patient lies, the high-frequency generator, the control unit for radiologic technologists and doctors, and the software for storing images. The gantry contains the most important parts, i.e. the X-ray tube and radiation detectors, as well as the tube and device cooling system [2]. In the rotating part of the gantry there are also slip rings, i.e. sliding rings that enable continuous rotation of the x-ray tube and detector banana, as well as accelerated rotation and reduction of the duration of the examination, without tangling the cables that supply the X-ray tube with electricity [8]. X-ray tubes on CT devices operate at a voltage of up to 140 kV and are made of materials with a high thermal capacity [2]. An X-ray tube primarily consists of a cathode and an anode located inside a vacuum glass tube, and X-rays are produced by accelerating electrons and their impact on the anode's focal point [3]. An important component of a CT device is a collimator that creates a narrow beam of X-rays at the exit of the X-ray tube. It is made of lead or other metals with a high atomic number. The collimator ensures the directing of the primary beam of X-rays to the patient and maximum collimation in order to reduce the radiation dose [2].

Detectors on the CT device

CT technology belongs to digital radiological methods, which means that, specifically in CT devices, the electrical pulses that arise due to the collision of the X-ray beam with the detector are digitized in an analog-to-digital converter. Then, using special software and computer programs, the signals are further converted into a visual image on the monitor of the control unit [2]. The transition to such detectors represents a complete transition of radiology to a digital way of working, and digital detectors enable a qualitative and quantitative representation of the absorption of X-rays through the recorded object [1, 2, 7]. The digital image displayed on the monitor is composed of basic image elements (pixels). Each pixel also forms a volume element of the image (voxel), the other two dimensions which indicate the thickness of the layer recorded by the CT device. These elements together form the image matrix, which determines the spatial resolution. The greater the number of pixels, or voxels in the matrix, the better is the spatial resolution. On today's devices, the usual matrix size is 1024 x 1024 pixels [1, 2].

In modern devices, the detector system is called a detector "banana", which means the placement of the detector in an arc, i.e. linear. There are scintillation and gas detectors. Scintillation detectors work on the principle of scintillation, i.e. the shimmering of the material. After the contact of X-rays with certain materials, they scintillate or it may be said that they sparkle, and the amount of visible light produced by scintillation is proportional to the amount of ionizing radiation that came into contact with the detector [2]. There are also gas detectors, and the noble gas xenon is used the most. In ionization sections that were filled with gas, X-rays ionized gas atoms and interacted with electrons. This technology was mostly abandoned in the eighties of the last century because solid-state detectors had better properties and enabled greater sensitivity and resolution [2, 7]. The radiation detector is a fundamental component for image creation and significantly affects image quality and dose efficiency [4].

Energy integrating CT detectors

CT devices that are in use today have the Energy integrating detectors (EID). Today's detectors are mostly scintillation, built of the ceramic material gadolinium oxysulfite, and scintillation occurs when the X-ray comes into contact with the detector, and the photodiode converts visible light into an electrical signal for creating an image [7, 9]. This type of conversion and signal processing is called indirect conversion [10]. Individual elements of these detectors are separated by thin septums (Figure 2), but this reduces the efficiency of X-ray detection and affects the spatial resolution [9]. Thin optical barriers are directing light towards the optical photon sensor [11]. The intensity of the resulting light is proportional to the energy and number of incident photons, however, it does not represent the energy of individual, but the integrated energy of several incident photons [7, 9]. The name "Energy integrating detectors" originates from this fact, because the electrical signal is ultimately proportional to the total X-ray energy integrated in the detector [9]. Quality control of these detectors is performed every 6 months, while calibration is performed daily to optimize imaging parameters [12].

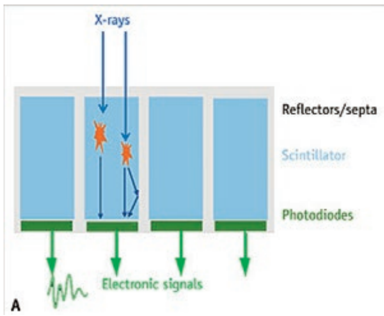

Figure 2. Schematic representation of the Energy integrating detector and its septums
Source: https://pubmed.ncbi.nlm.nih.gov/36047540/

The aim of the article

The aim of this article is to describe the new detector technology for CT imaging and highlight its technical characteristics and innovative solutions. The article is supported by literature published over the last 10 years in publicly available databases, especially scientific research that emphasizes the clinical implementation of the detector and the comparison of the Photon counting detector (PCD) in relation to the EID detector.

Discussion

Development of the Photon counting CT detector

The development of CT detectors reaches its peak with the development of the PCD detector, which registers every single incident photon and reads its energy [7]. The greatest development of this type of detector was in the past decade. Recent advances in semiconductor design have enabled the detection of a single photon and accelerated the development of these detectors [10]. Until recently, the PCD detector was used only for nuclear imaging because the amount of incident photons is much smaller than for CT, but constant innovations in physics and engineering bring changes and improvements to this system and enable new clinical applications of the PCD detector [4, 12]. Prototypes of CT devices with a PCD detector were set up in the research laboratories, and research was mostly performed on phantoms and small animals [13]. Phantom analyzes were, for example, performed for dedicated diagnosis of stroke, when it is crucial to observe small intracranial hemorrhages and occlusions of large arteries of the brain. The advantage of the PCD detector in the form of a reduced pixel size comes to the fore in these images. Research has shown an improvement in the visibility of millimeter lesions by 20% with PCD detectors [14]. The first PCD system that could perform diagnostic tests on humans was introduced in 2010 [15]. The first approval for use for clinical purposes was issued by the American Food and Drug Administration (FDA) on September 30, 2021 [16].

Characteristics of the Photon counting CT detector

PCD detectors have a two-layer structure (Figure 3). The upper layer consists of a sensor made of cadmium and tellurium, and sometimes zinc or silicon, which is hit by X-rays, and the lower layer is a bottom with a matrix of pixels [4, 7]. The bottom layer is called the layer of anodes that are connected to integrated circuits. Complex sensors are 2-3 mm thick and absorb X-rays in the range of 20-140 kV [12]. The absorption efficiency of incident photons depends on the sensor material and its thickness. Materials with a high atomic number, such as CdTe sensors, have higher absorption efficiency and are the most commonly used sensors in PCD detectors [17]. The matrix consists of approximately 250,000 pixels, each with an area of 0.05 mm2. In the upper layer, photons perform ionization, i.e. release electrons, and then the microelectronics from the lower layer registers the photons and their energy [7]. Between the two layers of the detector, traveling from the anode to the cathode, a negative bias voltage is created which is responsible for creating the electric field [12]. Quantitative measurement of the amount of energy of individual photons occurs in several phases, i.e. predetermined energy thresholds or containers [7]. Each detected output signal is compared with pre-calibrated values and incident photon energies are distributed into energy thresholds. The number of possible energy thresholds depends on the design of the detector, but most often it is between 2 and 8. Energy thresholds should be set before exposure, and energy threshold values are expressed in kV [18].

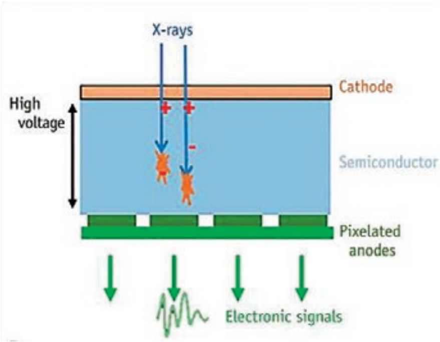

Figure 3. Schematic representation of the Photon counting detector
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC9434736/bin/kjr-23-854-g001.jpg

Working principle of the Photon counting CT detector

In order to develop PCD detectors, it is necessary to achieve high counting speed with high detection efficiency and optimal detector parameters [12]. The photon counting rate required by clinical CT systems is about one million counts per second per mm2 [19]. The PCD detector counts individual photons in each individual energy window. The energy of each incident photon creates electric charges that travel through the detector under the influence of an electric field. The charges induce a pulse signal that travels to the counter, one of the most important components of the detector [12]. These detectors give photons of different energies a different shade of gray scale in the final image, proportional to the energy of the incident photon. This enables a process called energy weighting, which means determining the exact energy value of each incident photon, despite the fact that X-rays produce photons of a wide range of energies. Energy weighting occurs inside the detector or during image processing, so the contrast-to-noise ratio (CNR) increases up to 15-57% compared to EID detectors [10]. Other processes also take place in the detector, such as signal processing in the digital-analog converter and calibration [12]. The output signal is processed in electrical comparators and counters [18].

Spatial resolution of the Photon counting CT detector

The PCD detector offers the potential to double the spatial resolution compared to current detectors. Since 1990, the spatial resolution has not improved significantly. That's why PCD detectors are the real sensation because they offer a significant increase in resolution with a minimal increase in radiation dose. The lack of a scintillation layer in these detectors greatly affects the improvement of spatial resolution and the better use of Ultra hard resolution (UHR) imaging with a low dose of radiation [4]. The lack of partitions, the so-called septa, between individual detector units also increases the geometric resolution of the PCD detector. Smaller pixel size eliminates the use of comb filters and grid, in that way increasing dose utilization rate. There is also a reduction of noise on the images [18]. The spatial resolution of the PCD detector is between 2.81 and 4.00 lp/mm compared to the EID detector, where it is about 2.08 lp/mm [4].

Some studies have compared the parameters of UHR images and the MACRO mode, i.e. normal high-resolution image. In UHR mode, the spatial resolution is 3.33 lp/mm, and the pixel size is 0.25 x 0.25 mm, while the pixel size in MACRO mode is 0.5 x 0.5 mm. Research has shown

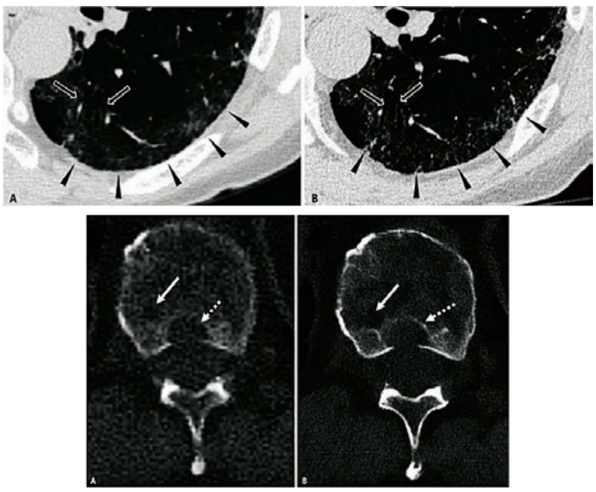

Figure 4. Comparison of the spatial resolution of lung parenhyma and vertebra on the images taken with EID detector (left) and PCD detector (right)
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC9434736/bin/kjr-23-854-g002.jpg

that the difference in pixel size and the use of certain reconstruction algorithms improves the spatial resolution by 87% and lowers the noise level by 15% [18, 20]. In general, the pixel size is smaller for PCD detectors and is 0.2-0.5 mm, while for EID detectors it is 1-1.4 mm. It is concluded that the spatial resolution is significantly better with the PCD system (Figure 4) [12]. By reducing the pixel size, there is also criticism that it will cause an increase in noise, but this possible problem is eliminated by using the soft and sharp reconstruction kernels [21]. Research shows the high superiority of the UHR mode in PCD detectors compared to EID detectors. This has been proven in studies that used images of phantoms, deceased patients and volunteer patients [20, 22, 23].

Contrast to noise ratio of the Photon counting CT detector

Noise on CT scans can be caused by two reasons, quantum and electronic noise. Quantum noise is determined by the number of detected photons and may depend on random interactions and the nature of the photons themselves, while electronic noise is not related to the number of detected photons, but originates from analog electronic circuits. The type of noise that prevails depends on the speed of photon flow to the detector, so that at high speed quantum noise prevails, and when the number of incident photons is low, the proportion of electronic noise is higher. There are various noise reduction solutions on today’s devices, and the noise is usually negligible when dealing with clinical imaging of average-sized patients where the average dose of radiation is used. Higher levels of noise appear in low-dose imaging and in overweight patients, where degradation of the imaging occurs. Higher levels of noise also appear along the longitudinal line of the body, for example along the line connecting the shoulders [24, 25]. Electronic noise is usually detected as a low energy signal, so the PCD detector classifies it in lower energy thresholds. For example, if the low energy threshold is set to 25 kV, low energy noise amplitudes can be excluded from the set of measured data, so they do not affect the image quality. It is unlikely that photons with such low energy will be of diagnostic significance, so such photons can be excluded from the obtained data in order to reduce electronic noise [18]. The key features of the PCD detector are better CNR and reduction of noise in the images, which enables the reduction of the radiation dose and the amount of contrast agent that is used (Figure 5) [26].

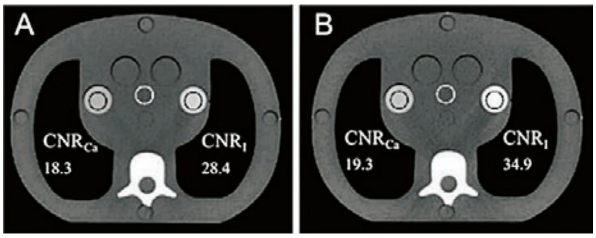

Figure 5. Comparison of the CNR on images taken with EID detector (left) and PCD detector (right) tested with anthropomorphic phantom
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC8409241/bin/nihms-1720806-f0002.jpg

Radiation dose efficiency

Dose efficiency is a measure that depends on the performance of the detector. The parameter that affects the most on the dose efficiency is the noise in the image. Noise and dose efficiency are inversely proportional. A reduction of noise also means an increase of dose efficiency, i.e. a decrease of the dose ultimately received by the patient. The assessment of these parameters is performed by measuring noise with an equal dose of radiation [10]. The conducted research showed that for imaging of small objects it is possible to increase the dose efficiency by 30% using the noise reduction method [27]. The dose efficiency is proportional to the CNR [28]. For appropriate dose efficiency, it is important to ensure the optimization of tube parameters, i.e. voltage and filtration, depending on the diagnostic test [29].

Radiation dose reduction

Several features of the PCD detector are crucial for reducing the radiation dose, especially for imaging pediatric patients. Higher spatial resolution and better CNR increase anatomical visibility in pediatric patients and decrease the radiation dose delivered to the patient [21]. With the UHR mode, the dose can be reduced by 20-30% without degrading the image quality. Tin filters are used to remove low energy photons, i.e. scattered radiation [30]. Lowering the radiation dose is also useful for repeated imaging in young patients with chronic diseases. For example, low-dose lung imaging stands out because it provides an excellent representation of anatomical and pathological structures with a low dose of radiation, and also reducing the long-term danger of repeated imaging [21].

Contrast agents

In radiology, different contrast agents are used for the purpose of better visualization of structures in the body, and their main working principle is changing the absorption of X-rays in the specific organ where it is applied. Positive contrast agents are made of chemical elements with a high atomic number and absorb X-rays stronger than the surrounding structures [1]. Iodine is one of the most commonly used contrast agents in CT examinations, but sometimes its differentiation is difficult, for example differentiation of the contrast agent from a calcified plaque in the lumen of a blood vessel [10]. Iodine belongs to the group of water-soluble contrast agents and is used to visualize most organs, organ systems, blood vessels and body cavities. It is mostly excreted by the kidneys, so it is called a urotropic contrast agent. When applying iodine contrast agents, the radiation dose increases by about 20%. When preparing the patient and injectioning the contrast agent, great attention should be given to possible side effects and their prevention [1].

Improved signal of the iodine contrast agent

The PCD detector offers improved iodine representation at the same tube potential as the EID detector (Figure 6). Routine tests with enhanced contrast are improved and the radiation dose received by the patient is reduced [21]. Low-energy photons contribute more to image contrast with PCD than with EID detectors, leading to improved CNR and reduced noise in iodine contrast imaging [12, 31-33]. In children, the improved CNR of iodine means a significant reduction in radiation dose. Imaging using a PCD detector can reduce the dose of radiation or the volume of applied contrast agent, which additionally protects the patient from the harmful effects of CT imaging. This is particularly important in patients with damaged renal function because the toxic effect of the contrast agent is reduced [18]. Increased signal of iodine contrast agent contributes to better acquisition of low-dose images. Contrast can be additionally corrected during postprocessing [21]. It was shown that the PCD detector enables more accurate quantification of iodine contrast agent in phantoms of different sizes, and a lapse of 0.5 mg of iodine per mL was shown during the measurement [18]. Specifically, when imaging the abdomen, a better delineation of neoplasms and metastases within the organ parenchyma is achieved. These are the so-called low-contrast lesions because their density is very similar to parenchyma, so they are difficult to differentiate using classic CT detectors. Using the PCD detector increases the visibility of hypervascularized and hypovascularized tumors of the liver and pancreas [11, 34, 35]. At higher tube voltages, 120-140 kV, the appearance of the iodine contrast agent is more visible, and it can be observed in all patients, from newborns to patients with increased body weight [28].

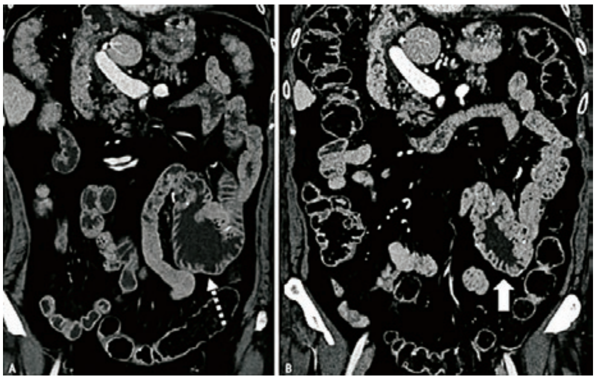

Figure 6. Comparison of the iodine contrast visibility in abdomen on images taken with EID detector (left) and PCD detector (right)
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC9434736/bin/kjr-23-854-g005.jpg

Spectral imaging

The incident photons, which are characterized by different energies, contain spectral information that enables imaging specific to a certain material, i.e. a certain substance [10]. This method of imaging transcends dual-energy imaging and moves into multi-energy imaging [36]. This type of imaging is the main initiator behind the development of the PCD detector and the desire to introduce this detector into clinical use to facilitate the acquisition of certain diagnostic tests. Multi-energy imaging is based on the ability of the PCD detector to differentiate photons of different energies at the same tube potential. In that way image is simplified and possible sources of artifacts are eliminated, spatial resolution is increased and better contrast is achieved [18]. Energy thresholds can be adjusted to provide optimal spectral contrast at any kV value or adapted to the size of the patient, which means that it is possible to set energy thresholds individually [35]. Multi-energy imaging offers a choice of several types of reconstruction that are particularly suitable for visualization of contrast agents within the organism, such as virtual mono-energy images, virtual non-contrast images and reconstructions with automatic bone removal [18]. Multi-energy reconstructions are, for example, of great importance in the musculoskeletal system in the assessment of gout and bone edema (Figure 7) [21].

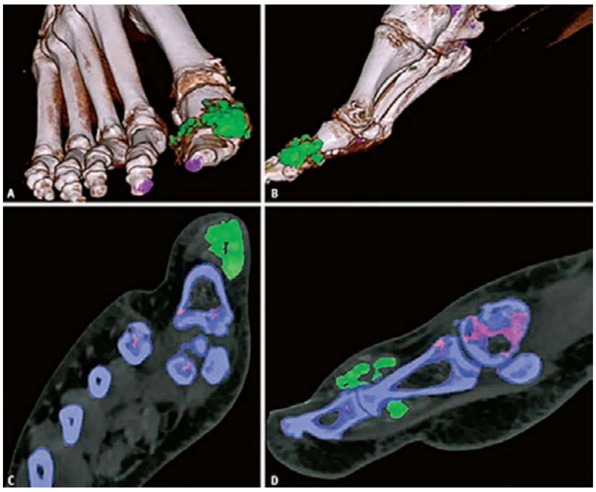

Figure 7. Image taken with PCD detector which represents accumulation of monosodium urate (green color) indicating the presence of gout
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC9434736/bin/kjr-23-854-g010.jpg

Artifacts

Another advantage of PCD detectors is the reduction of artifacts, especially streaking artifacts, beam hardening, artifacts created by the metal object in the organism or calcium “blooming” artifacts [21]. Even in patients with increased body weight, the PCD detector offers various advantages, i.e. it removes artifacts from high-attenuation image parts and removes electronic noise [37]. PCD detectors use certain mechanisms to reduce streaks, such as voltage modulation on the tube or the multidimensional filtering. Metal artifacts are common regardless of detector type, and cause artifacts through several mechanisms of action. Eliminating metal artifacts depends on the type of metal. For example, gold and mercury absorb almost all the photons they come into contact with, and a special software tool is needed to remove the bright areas in the images caused by this phenomenon. Artifacts created with light metals also require a software solution [35]. Reduction of metal artifacts is best achieved by taking high-energy images and using a tin filter to shape the radiation beam or using certain reconstruction techniques [37, 38].

The calcium blooming is an artifact that often occurs in cardiovascular system images, and the main reason for this phenomenon is insufficient spatial resolution (Figure 8). This artifact is particularly noticeable when imaging smaller blood vessels, for example coronary vessels, and structures that are filled with iodine contrast agent or when a stent is present. Then it is difficult for the radiologist to differentiate the contrast in the lumen of the

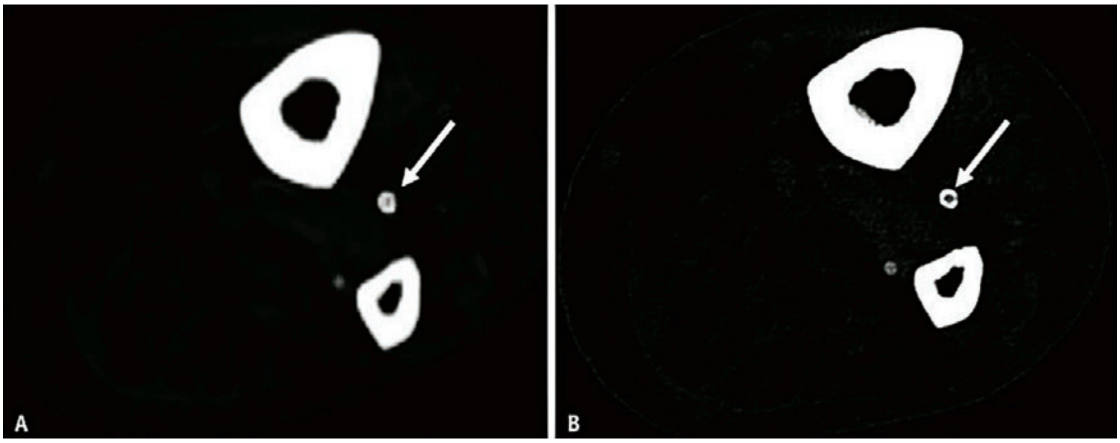

Figure 8. Blood vessel CT scan taken with EID detector (left) and PCD detector (right) that shows removal of the calcium blooming arifact
Source: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC9434736/bin/kjr-23-854-g012.jpg

blood vessel from the wall of the blood vessel. Calcified plaques or stents appear larger on the image than they are in reality, and this may even result in inappropriate treatment of the patient [21]. PCD detectors improve visibility of structures and remove blooming artifact due to high spatial resolution and smaller voxel size [39]. The beam hardening artifact appears as the dark areas, i.e. streaks on images next to objects with high attenuation, such as cortical bone or metal implants [18, 35]. This phenomenon causes the inaccuracy of the CT number of soft tissues and the irregular appearance of the image [18].

Advantages and disadvantages of Photon counting and Energy integrating detectors

PCD detectors face certain challenges in their work, and scientists have not yet found the right solution for certain parameters. This includes limitations independent of the photon flow, such as charge splitting and charge trapping, and effects related to the photon flow, such as pulse pileup. In almost all PCD systems, the pulse pileup appears and it depends on the photon counting rate and the dead time of the detector. When multiple pulses reach the detector, photons accumulate and the pulses are registered as one photon, and the main purpose of this detector is lost – the recording of each individual photon and its energy [12]. As a result of this phenomenon, there is an error in the counting speed, distortion of the spectrum and a decrease in image quality [10]. There is also a phenomenon called charge sharing, which affects detector performance and image quality. When an incident photon is detected by a PCD detector, a charge cloud is created inside the material. The charge cloud, due to the electric field inside the detector, tries to reach the appropriate anode and reach the appropriate energy threshold. As a result of this process, the charge cloud be divided and detected by two or more anodes, and its energy is detected as lower than it actually is, so the photon is detected twice, both times with the wrong energy value [12, 35].

PCD detectors have a higher energy detection potential because electrons are detected directly by the anode, while EID detectors work on the principle of scintillation and part of the energy is lost by scattering in the scintillator until it is registered by the photodiode. Due to the separation of energy thresholds, PCD detectors generate lower noise and have better Signal-to-noise ratio (SNR). This increases the dose efficiency and consequently reduces the radiation dose [7]. The PCD detector achieves a better spatial resolution due to the smaller pixel size and enables a sharper display of pathological changes, primarily in lung and bone imaging [7, 19].

Future development of the Photon counting detector

The technical possibilities of the PCD detector continue to improve, and the question arises as how the new systems can improve the already existing applications of this detector in clinical practice or create new imaging protocols [35]. As in most branches of radiology, the implementation of artificial intelligence is being researched in the use of PCD detectors, more precisely for the correction of its parameters. Learning methods using neural networks and machine learning are being researched [40-43]. Despite advanced algorithms, the amount of data produced by these detectors can be a challenge for artificial intelligence approaches [44].

Conclusion

The PCD detector technology is a part of modern CT imaging in the research process, but compared to the EID detector, it offers numerous advantages in the acquisition of diagnostic and interventional procedures. The key feature that sets this technology apart from conventional technology is the direct conversion of X-rays into an electrical pulse. High spatial resolution, noise reduction and increased dose efficiency are parameters that bring this technology to the door of implementation in everyday clinical practice. This process will be time-consuming, but it will bring CT imaging into a new era. Prototypes of a CT device with a PCD detector were made for the simulation and research of imaging on phantoms, animals and deceased patients, and later for in vivo imaging of volunteers. Scientists who comprehend the new practical and theoretical knowledge about this technology are making it closer to healthcare institutions and radiological technologists around the world. With the close cooperation of radiological technologists, radiologists and physicists, it will be possible to optimize imaging parameters adapted to each diagnostic task.

All data in this paper are part of the results of the undergraduate thesis " Photon counting CT detector" written at the University Department of Health Studies, University of Split [45]. ■

Karakteristike novog CT detektora s brojačem fotona

Sažetak

Radiologija je jedna od najbrže rastućih grana medicine, a upravo jedan od uređaja koji najbrže prati ovaj razvoj je i uređaj za CT. Znanstvenici su najviše fokusirani na istraživanje poboljšanja parametara rendgenske cijevi i detektora kao najvažnijih dijelova ovog modaliteta oslikavanja. Uz detektor s integriranjem energije koji je danas u širokoj uporabi u sklopu svakodnevne kliničke praske, predmet istraživanja je i detektor s brojačem fotona. Brojne vrline koje se ističu kod nove vrste detektora zaslužne su za opće mišljenje da će u budućnosti ova nova tehnologija zavladati CT uređajima u kliničkoj praksi poboljšavajući akviziciju brojnih dijagnostičkih i intervencijskih postupaka. Smatra se da ova tehnologija radiologiju uvodi u novu eru upravo svojom poboljšanom prostornom i kontrastnom rezolucijom, smanjenjem šuma i povećanjem omjera kontrasta i šuma, ali i novim mogućnostima kao što su oslikavanje više kontrastnih sredstava istovremeno te višeenergetsko oslikavanje. Poboljšanja vidljivost joda pruža dodatne dijagnostičke mogućnosti te točniju diferencijaciju benignih i malignih suspektnih lezija. Znatno manja veličina piksela na ovom detektoru nudi gotovo dvostruko povećanje prostorne rezolucije, a samim time i povećanje učinkovitosti doze zračenja. Korištenje detektora s brojačem fotona je zato pogodno i kod pedijatrijskih bolesnika, a javlja se i širok spektar mogućnosti *low-dose* oslikavanja u svrhu preventivnog *screeninga* karcinoma. Procesi poput tzv. razgradnje materijala i spektralnog oslikavanja su još u procesu istraživanja, ali imaju veliki potencijal da optimiziraju buduću svakodnevnu kliničku praksu. Potrebno je još rada i objavljenih rezultata istraživanja, no zasigurno će ova tehnologija jednog dana olakšati posao radiološkog tehnologa i radiologa.

Ključne riječi: CT; detektor s brojačem fotona; detektor s integriranjem energije

Literature

1. Janković S, Mihanović F. Uvod u radiologiju. Split: Sveučilište u Splitu; 2014.
2. Janković S, Mihanović F, Punda A, Radović D, Barić A, Hrepić D. Radiološki uređaji i oprema u radiologiji, radioterapiji i nuklearnoj medicini. Split: Sveučilište u Splitu; 2015.
3. Liguori C, Frauenfelder G, Massaroni C, Saccomandi P, Giurazza F, Pitocco F, Marano R, Schena E. Emerging clinical applications of computed tomography. Medical Devices: Evidence and Research. 2015 Jun 5:265-78.
4. Sartoretti T, Wildberger JE, Flohr T, Alkadhi H. Photon-counting detector CT: early clinical experience review. The British Journal of Radiology. 2023 Feb;95:20220544.
5. https://www.health.harvard.edu/cancer/radiation-risk-from-medical-imaging (accessed on 5.9.2023.)
6. McCollough CH, Bushberg JT, Fletcher JG, Eckel LJ. Answers to common questions about the use and safety of CT scans. InMayo Clinic Proceedings 2015 Oct 1 (Vol. 90, No. 10, pp. 1380-1392). Elsevier.
7. Schlemmer HP. The Eye of the CT Scanner: The story of learning to see the invisible or from the fluorescent screen to the photon-counting detector. InRöFo-Fortschritte auf dem Gebiet der Röntgenstrahlen und der bildgebenden Verfahren 2021 Sep (Vol. 193, No. 09, pp. 1034-1049). Georg Thieme Verlag KG.
8. Tóth T, Hudák R. Computed tomography-its development, principle and image artifacts. Acta Mechanica Slovaca. 2013;17(4):40-7.
9. Nakamura Y, Higaki T, Kondo S, Kawashita I, Takahashi I, Awai K. An introduction to photon-counting detector CT (PCD CT) for radiologists. Japanese Journal of Radiology. 2023 Mar;41(3):266-82.
10. Wang AS, Pelc NJ. Spectral photon counting CT: Imaging algorithms and performance assessment. IEEE transactions on radiation and plasma medical sciences. 2020 Jul 7;5(4):453-64.
11. Willemink MJ, Persson M, Pourmorteza A, Pelc NJ, Fleischmann D. Photon-counting CT: technical principles and clinical prospects. Radiology. 2018 Nov;289(2):293-312.
12. Taguchi K, Iwanczyk JS. Vision 20/20: single photon counting x-ray detectors in medical imaging. Medical physics. 2013 Oct;40(10):100901.
13. Rajendran K, Löbker C, Schon BS, Bateman CJ, Younis RA, de Ruiter NJ, Chernoglazov AI, Ramyar M, Hooper GJ, Butler AP, Woodfield TB. Quantitative imaging of excised osteoarthritic cartilage using spectral CT. European radiology. 2017 Jan;27:384-92.
14. Ji X, Zhang R, Chen GH, Li K. Task-driven optimization of the non-spectral mode of photon counting CT for intracranial hemorrhage assessment. Physics in Medicine & Biology. 2019 Oct 31;64(21):215014.
15. Yu Z, Leng S, Jorgensen SM, Li Z, Gutjahr R, Chen B, Halaweish AF, Kappler S, Yu L, Ritman EL, McCollough CH. Evaluation of conventional imaging performance in a research whole-body CT system with a photon-counting detector array. Physics in Medicine & Biology. 2016 Feb 2;61(4):1572.
16. https://www.fda.gov/news-events/press-announcements/fda-clears-first-major-imaging-device-advancement-computed-tomography-nearly-decade (accessed on 7.9.2023.)
17. Hamann E, Koenig T, Zuber M, Cecilia A, Tyazhev A, Tolbanov O, Procz S, Fauler A, Baumbach T, Fiederle M. Performance of a Medipix3RX spectroscopic pixel detector with a high resistivity gallium arsenide sensor. IEEE transactions on medical imaging. 2014 Apr 14;34(3):707-15.

18. Leng S, Bruesewitz M, Tao S, Rajendran K, Halaweish AF, Campeau NG, Fletcher JG, McCollough CH. Photon-counting detector CT: system design and clinical applications of an emerging technology. Radiographics. 2019 May;39(3):729-43.
19. Liu X, Grönberg F, Sjölin M, Karlsson S, Danielsson M. Count rate performance of a silicon-strip detector for photon-counting spectral CT. Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment. 2016 Aug 11;827:102-6.
20. Leng S, Rajendran K, Gong H, Zhou W, Halaweish AF, Henning A, Kappler S, Baer M, Fletcher JG, McCollough CH. 150-μm spatial resolution using photon-counting detector computed tomography technology: technical performance and first patient images. Investigative radiology. 2018 Nov 1;53(11):655-62.
21. Esquivel A, Ferrero A, Mileto A, Baffour F, Horst K, Rajiah PS, Inoue A, Leng S, McCollough C, Fletcher JG. Photon-Counting Detector CT: Key Points Radiologists Should Know. Korean Journal of Radiology. 2022 Sep 1;23(9):854-65.
22. Leng S, Yu Z, Halaweish A, Kappler S, Hahn K, Henning A, Li Z, Lane J, Levin DL, Jorgensen S, Ritman E. Dose-efficient ultrahigh-resolution scan mode using a photon counting detector computed tomography system. Journal of Medical Imaging. 2016 Oct 1;3(4):043504-.
23. Zhou W, Lane JI, Carlson ML, Bruesewitz MR, Witte RJ, Koeller KK, Eckel LJ, Carter RE, McCollough CH, Leng S. Comparison of a photon-counting-detector CT with an energy-integrating-detector CT for temporal bone imaging: a cadaveric study. American Journal of Neuroradiology. 2018 Sep 1;39(9):1733-8.
24. Duan X, Wang J, Leng S, Schmidt B, Allmendinger T, Grant K, Flohr T, McCollough CH. Electronic noise in CT detectors: impact on image noise and artifacts. American Journal of Roentgenology. 2013 Oct;201(4):W626-32.
25. Liu Y, Leng S, Michalak GJ, Vrieze TJ, Duan X, Qu M, Shiung MM, McCollough CH, Fletcher JG. Reducing image noise in computed tomography (CT) colonography: effect of an integrated circuit CT detector. Journal of computer assisted tomography. 2014 May 1;38(3):398-403.
26. Higashigaito K, Mergen V, Eberhard M, Jungblut L, Hebeisen M, Rätzer S, Zanini B, Kobe A, Martini K, Euler A, Alkadhi H. CT angiography of the aorta using photon-counting detector CT with reduced contrast media volume. Radiology: Cardiothoracic Imaging. 2023 Jan 26;5(1):e220140.
27. Baek J, Pineda AR, Pelc NJ. To bin or not to bin? The effect of CT system limiting resolution on noise and detectability. Physics in Medicine & Biology. 2013 Feb 11;58(5):1433.
28. Gutjahr R, Halaweish AF, Yu Z, Leng S, Yu L, Li Z, Jorgensen SM, Ritman EL, Kappler S, McCollough CH. Human imaging with photon-counting-based CT at clinical dose levels: Contrast-to-noise ratio and cadaver studies. Investigative radiology. 2016 Jul;51(7):421.
29. Chen H, Xu C, Persson M, Danielsson M. Optimization of beam quality for photon-counting spectral computed tomography in head imaging: simulation study. Journal of Medical Imaging. 2015 Oct 1;2(4):043504-.
30. Rajendran K, Voss BA, Zhou W, Tao S, DeLone DR, Lane JI, Weaver JM, Carlson ML, Fletcher JG, McCollough CH, Leng S. Dose reduction for sinus and temporal bone imaging using photon-counting detector CT with an additional tin filter. Investigative radiology. 2020 Feb;55(2):91.
31. Bennett JR, Opie AM, Xu Q, Yu H, Walsh M, Butler A, Butler P, Cao G, Mohs A, Wang G. Hybrid spectral micro-CT: system design, implementation, and preliminary results. IEEE transactions on biomedical engineering. 2013 Aug 26;61(2):246-53.
32. Shikhaliev PM. Soft tissue imaging with photon counting spectroscopic CT. Physics in Medicine & Biology. 2015 Mar 5;60(6):2453.
33. Silkwood JD, Matthews KL, Shikhaliev PM. Photon counting spectral breast CT: effect of adaptive filtration on CT numbers, noise, and contrast to noise ratio. Medical physics. 2013 May;40(5):051905.
34. Iyer VR, Ehman EC, Khandelwal A, Wells ML, Lee YS, Weber NM, Johnson MP, Yu L, McCollough CH, Fletcher JG. Image quality in abdominal CT using an iodine contrast reduction algorithm employing patient size and weight and low kV CT technique. Acta Radiologica. 2020 Sep;61(9):1186-95.
35. Hsieh SS, Leng S, Rajendran K, Tao S, McCollough CH. Photon counting CT: clinical applications and future developments. IEEE transactions on radiation and plasma medical sciences. 2020 Aug 28;5(4):441-52.
36. McCollough CH, Leng S, Yu L, Fletcher JG. Dual-and multi-energy CT: principles, technical approaches, and clinical applications. Radiology. 2015 Sep;276(3):637-53.
37. Kalisz K, Halliburton S, Abbara S, Leipsic JA, Albrecht MH, Schoepf UJ, Rajiah P. Update on cardiovascular applications of multienergy CT. Radiographics. 2017 Nov;37(7):1955-74.
38. Zhou W, Bartlett DJ, Diehn FE, Glazebrook KN, Kotsenas AL, Carter RE, Fletcher JG, McCollough CH, Leng S. Reduction of metal artifacts and improvement in dose efficiency using photon counting detector CT and tin filtration. Investigative radiology. 2019 Apr;54(4):204.
39. Koons E, VanMeter PD, Rajendran K, Yu L, McCollough C, Leng S. Improved assessment of coronary artery luminal stenosis with heavy calcifications using high-resolution photon-counting detector CT. InMedical Imaging 2022: Physics of Medical Imaging 2022 Apr 4 (Vol. 12031, pp. 319-323). SPIE.
40. Touch M, Clark DP, Barber W, Badea CT. A neural network-based method for spectral distortion correction in photon counting x-ray CT. Physics in Medicine & Biology. 2016 Jul 29;61(16):6132.
41. Jenkins P, Schmidt TG. Experimental study of neural network material decomposition to account for pulse-pileup effects in photon-counting spectral CT. InMedical Imaging 2019: Physics of Medical Imaging 2019 Mar 1 (Vol. 10948, pp. 498-503). SPIE.
42. Lu Y, Kowarschik M, Huang X, Chen S, Ren Q, Fahrig R, Hornegger J, Maier A. Material decomposition using ensemble learning for spectral X-ray imaging. IEEE Transactions on Radiation and Plasma Medical Sciences. 2018 Feb 12;2(3):194-204.
43. Hornegger J, Maier A. A learning-based material decomposition pipeline for multi-energy x-ray imaging. Medical physics. 2019 Feb;46(2):689-703.
44. Zimmerman KC, Sharma G, Parchur AK, Joshi A, Schmidt TG. Experimental investigation of neural network estimator and transfer learning techniques for K-edge spectral CT imaging. Medical physics. 2020 Feb;47(2):541-51.
45. Mimica P. CT detektor s brojačem fotona (Završni rad). Split: Sveučilište u Splitu, Sveučilišni odjel zdravstvenih studija; 2023 (pristupljeno 5.9.2023.) Available at: https://repo.ozs.unist.hr/islandora/object/ozs%3A1412