X-ray detectors in medical imaging

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A B S T R A C T
Healthcare systems are subject to continuous adaptation, following trends such as the change of demographic structures, the rise of life-style related and chronic diseases, and the need for efficient and outcome-oriented procedures. This also influences the design of new imaging systems as well as their components.

The applications of X-ray imaging in the medical field are manifold and have led to dedicated modalities supporting specific imaging requirements, for example in computed tomography (CT), radiography, angiography, surgery or mammography, delivering projection or volumetric imaging data. Depending on the clinical needs, some X-ray systems enable diagnostic imaging while others support interventional procedures. X-ray detector design requirements for the different medical applications can vary strongly with respect to size and shape, spatial resolution, frame rates and X-ray flux, among others.

Today, integrating X-ray detectors are in common use. They are predominantly based on scintillators (e.g. CsI or Gd₂O₂S) and arrays of photodiodes made from crystalline silicon (Si) or amorphous silicon (a-Si) or they employ semiconductors (e.g. Se) with active a-Si readout matrices.

Ongoing and future developments of X-ray detectors will include optimization of current state-of-the-art integrating detectors in terms of performance and cost, will enable the usage of large size CMOS-based detectors, and may facilitate photon counting techniques with the potential to further enhance performance characteristics and foster the prospect of new clinical applications.

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1. Drivers shaping healthcare and X-ray imaging

Several global trends are shaping current healthcare systems. Among them is the change of demographic structures in many developed countries where the population pyramids are either stationary or even contracting while life expectancy is increasing. This and other factors lead to an increasing occurrence of diseases such as cardiovascular diseases, cancer, stroke or diabetes which are among the leading causes of death. Another continuing trend relates to an ever increasing number of disease patterns which can be treated by applying minimally invasive techniques rather than by open surgery. Healthcare information technology has become an integral part within hospitals and networks as digital imaging modalities are connected to picture archiving systems (PACS) and hospital or radiology information systems (HIS/RIS) and as the whole clinical and administrative workflow in hospitals is increasingly supported by software solutions and processes. Emerging rural healthcare in developing countries facilitates a broader access of the population to the local healthcare systems. As the annual fraction of GDP spent on healthcare has increased in many countries over the years, solutions to control or reduce overall cost are required and have led to more outcome-oriented reimbursement schemes.

These general trends influence the design and performance of current and future X-ray systems such as computed tomography scanners, C-arm systems for vascular or surgical X-ray imaging, general radiography or mammography systems. Systems used for interventional procedures will furthermore need to support image fusion with other imaging modalities or enable integration of other equipment such as ablation, mapping or navigation devices.

Future developments of X-ray detectors will have to support the increasing clinical demands and new applications, assist improved workflows, reflect the increasing awareness of efficiently utilizing X-ray dose, and be cost-efficient.

2. Clinical applications

Shortly after the discovery of X-rays by W.C. Röntgen in 1895, X-ray imaging was established as the first medical imaging technique and has continuously evolved while new imaging modalities such as magnetic resonance, positron emission tomography (PET) or ultrasound were introduced. Today the medical applications of X-ray imaging as well as the X-ray systems enabling the related clinical tasks are manifold. In the following a broad but far from exhaustive overview of clinical applications...
for important diagnostic and interventional X-ray modalities is provided.

Radiography which covers chest, trauma, pediatric, orthopedic and general examinations has evolved in many radiology departments during the past two decades from a technique dominated by analog screen-film combinations to a digital technique based on flat detectors. Radiography has profited from that change in many ways. While not possible for film, the separation of acquisition medium (detector) and display medium (monitor) has permitted optimizing these parts of the image chain independently and introducing image processing as a means to further improve image quality substantially. Workflow has improved because of the instant image display, eradicating the time-consuming development of film. It has further profited from connecting the X-ray units with HIS/RIS, allowing the download of predetermined worklists including patient information and examination type. PACS has eliminated film loss and enabled instant and simultaneous image access by physicians from different locations. The introduction of flat detectors with high detective quantum efficiency (DQE) [1] has led to dose reduction in skeletal and chest radiography [2]. Portable, wireless detectors have further broadened the spectrum of usability, by providing free positioning. The introduction of digital radiography has facilitated techniques such as image-composing to cover the whole body, dual-energy subtraction methods capable of removing bony structures by an appropriate linear combination of two chest images taken with different X-ray spectra, or the alternative and more dose-efficient method of applying bone-suppression algorithms on a single standard chest image.

In mammography, the change from screen-film combinations to flat detectors has led to similar changes of improved image quality and simplified workflow. Digital mammography has also enabled new features such as computer-aided detection (CAD) of microcalcifications, integrated stereotactic biopsy, and tomosynthesis [3], a technique generating 3D image data which promises to reduce tissue overlap, improve the separation of lesions, and enhance the visualization of microcalcifications.

Floor-mounted, ceiling-mounted or robotic C-arm systems (Fig. 1) are used for diagnostic and interventional procedures in radiology, cardiology, oncology or surgery environments. Traditional endovascular imaging includes coronary angiography, general angiography and neuroangiography. During X-ray acquisition, iodine-based contrast medium is injected intra-arterially, improving visualization of the vascular structure and thus generating a series of images of diagnostic quality. Digital subtraction angiography (DSA) may be used whenever organ motion is negligible enhancing the visibility by subtracting an image with anatomical background from images containing contrast medium in the blood vessels. Fig. 2 depicts an example of a complete DSA sequence of the cerebral vessel system in a single color-coded image, providing the physician with a better understanding of the contrast flow within the pathology and a visualization of the success after treatment. 3D information may be acquired by a technique often referred to as “flat-panel cone-beam CT”. Here the C-arm rotates around the patient while up to several hundred images are acquired from which volumetric data is reconstructed [4]. Low dose fluoroscopy or roadmapping is applied to visualize devices such as guidewires, balloons, stents or coils used in interventional cases. Common endovascular procedures include treatment of coronary stenoses by balloon angioplasty, coiling of cerebral aneurysms, thrombolysis or treatment of arteriovenous malformations. Many other applications have become routine or are being established, among them radio frequency ablation in electrophysiology, tumor embolization, aortic stent grafting (Fig. 1), transcatheter aortic valve implantation (TAVI) or X-ray guidance during surgical procedures such as spinal fusion.

Other specialized X-ray systems include fluoroscopy systems for gastrointestinal studies, myelography, arthrography or the evaluation of urinary tract infections, mobile X-ray systems for bed-side usage, and mobile C-arm systems for intraoperative use in orthopedic, trauma and spine surgery.

Computed tomography, introduced in the early 1970s by G.N. Hounsfield, is an imaging technique based on the reconstruction of the linear X-ray attenuation coefficient as a function of spatial coordinates in the imaging plane. To perform that task, the attenuation of an X-ray beam crossing the object or patient has to be measured from a large number of different angles. Cross-sectional images are reconstructed by applying techniques such as filtered back projection. The extension to volumetric acquisition with multi-slice spiral CT scanners requires a cone-beam reconstruction by approximate algorithms such as the one first developed by Feldkamp, Davis and Kress [5]. More recently, iterative maximum-likelihood expectation-maximization methods taking the physical model of the scanner into account have been introduced [6]. Several rendering techniques such as surface rendering, volume rendering or image segmentation are used for 3D visualization of the data. State-of-the-art subsecond multi-slice spiral CT scanners enable a wide range of clinical applications including the visualization of tumors and lesions, cerebral perfusion imaging, i.e. visualizing the blood flow into the brain tissue following an acute stroke or aneurysm fracture, interventional puncture guidance for the extraction of biopsy samples or the detection of the degree of plaque calcification in coronary arteries (calcium scoring). Fig. 3 shows an example of a coronary CT angiogram (CTA) to evaluate the restenosis of an implanted stent. A particular system configuration, called dual-source CT, consists of two X-ray tubes and two detectors in the rotating gantry. This configuration can be used to cover a larger scan volume during one gantry rotation which, for instance, eliminates the need and risk of sedation during pediatric chest scans [7] or it can be used to simultaneously acquire data with different X-ray spectra for dual-energy imaging.

3. Current main-stream X-ray detector technologies

The various clinical X-ray applications are generating specific requirements for the respective medical systems and in turn for
the detectors used therein. Table 1 summarizes some of the typical design parameters of X-ray detectors and the conditions they are operated in for general X-ray, full-field mammography, angiography and multi-slice CT.

Table 1

<table>
<thead>
<tr>
<th></th>
<th>General X-ray</th>
<th>Full-field mammography</th>
<th>Angiography</th>
<th>Multi-slice CT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Detector area [cm²]</td>
<td>35 × 43</td>
<td>18 × 24</td>
<td>20 × 20</td>
<td>~1 × 100 (2 rows)</td>
</tr>
<tr>
<td></td>
<td>43 × 43</td>
<td>24 × 30</td>
<td>30 × 40</td>
<td>~13 × 100 (128 rows)</td>
</tr>
<tr>
<td>Detector structure</td>
<td>Flat</td>
<td>Flat</td>
<td>Flat</td>
<td>Polygonal shape</td>
</tr>
<tr>
<td>Pixel size [μm]</td>
<td>130–200</td>
<td>70–100</td>
<td>150–200</td>
<td>~1000</td>
</tr>
<tr>
<td>Number of pixels (approx.)</td>
<td>~5–10 M</td>
<td>~4–14 M</td>
<td>~1–5 M</td>
<td>~1500–120,000</td>
</tr>
<tr>
<td>Frame rate [1/s]</td>
<td>1 (single image)</td>
<td>1 (single image) 2 (tomosynthesis)</td>
<td>60–125</td>
<td>2000–5000</td>
</tr>
<tr>
<td>Max. photon energy [keV]</td>
<td>40–150</td>
<td>23–49</td>
<td>~10⁶</td>
<td>80–140</td>
</tr>
<tr>
<td>Max. photon flux [1/mm²s]</td>
<td>~10⁸</td>
<td>~10⁷</td>
<td>~10⁹</td>
<td>~10⁹</td>
</tr>
</tbody>
</table>

Fig. 2. Dynamic flow evaluation. The single color-coded image (right) contains the temporal information of the flow of contrast medium through the cerebral vessel tree and tissue, generated from a sequence of digital subtraction angiography (DSA) images (left). Courtesy of Dr. C. Strother, University of Wisconsin Hospital, Madison, WI, USA.

Fig. 3. Volume rendering of a coronary CT angiogram (CTA) to evaluate the in-stent restenosis (left). The calcifications (orange arrows) as well as the structure of the implanted stent are clearly visible in the multiplanar reconstruction (MPR) image (right). Courtesy of Deutsches Herzzentrum, Munich, Germany. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

Two types of integrating detection processes prevail: (i) direct conversion based on semi-conductor X-ray detector material, directly creating electric charge, and (ii) indirect conversion based on solid-state scintillators, first generating optical photons which are subsequently converted into electric charge in a photodiode. Scintillator-based, indirect conversion is the dominant detection principle used across all clinical applications. For mammography, however, direct conversion amorphous
selenium (a-Se) has established itself as an alternative X-ray detection material.

3.1. Amorphous silicon flat detectors

Flat detectors, first introduced in the late 1990s, have by now become the gold standard for digital radiography, full-field mammography, fluoroscopic and endovascular imaging as well as imaging in hybrid surgical environments.

The development of liquid crystal displays based on large-area amorphous silicon for the consumer market has made the technology available for medical X-ray imaging. Hydrogenated amorphous silicon (a-Si:H) has sufficiently good semiconductor properties so that photodiodes with high quantum efficiency in the green part of the visible spectrum as well as thin-film switching transistors (TFT) can be made. Furthermore, plasma deposition allows processing very large areas on a single glass substrate exceeding 40 × 40 cm². Another advantage is the high radiation hardness of a-Si:H. These properties are ideally suited to make large-area active-matrices, the central building block of flat detectors which are discussed in the following, starting with the indirect conversion flavor and followed by the direct conversion type.

Indirect conversion, scintillator-based flat detectors [8—11] are the main type of flat detectors in use, having replaced image intensifiers in angiography and enabled digital radiography with instant image display and high dose efficiency, not available with screen-film or computed radiography systems. In this detection process, an X-ray photon is absorbed in the scintillator, predominantly by the photoelectric effect, creating a large number of electron—hole pairs which are trapped in luminescence centers and recombine to produce scintillation photons in the visible range [12]. The light is converted into electric charge in a pixelized matrix of photodiodes. Fig. 4 shows a schematic view of this detector configuration, described below in more detail.

The scintillator of choice for flat detectors is thallium-doped cesium iodide (CsI:Tl) for several favorable properties. Using physical vapor deposition, CsI:Tl is grown in a columnar, “needle-like” structure with individual crystals measuring some 5–10 μm in diameter and up to 600 μm or more in length. This structure facilitates the collection of the emitted light in the photo-diode below. The emitted green light is well matched to the photoefficiency of the a-Si:H photodiode with a maximum around 550 nm. Due to the high atomic numbers of 55 and 53 for Cs and I, respectively, and an effective density of about 3.6 g/cm³, CsI has very good X-ray absorption properties and is therefore particularly well suited to enable high DQE(f) for the clinically relevant spatial frequencies f and the different X-ray energy spectra required for the wide range of imaging tasks performed with a vascular or radiographic X-ray system.

The acquisition and readout principle consists of the following steps. First, the inversely biased photodiodes are charged to a given voltage. Then, during X-ray acquisition, the impinging optical photons, created during the photoabsorption and scintillation processes, generate electron—hole pairs which gradually discharge the photodiodes. The readout process is accomplished by addressing all TFTs in a given line, measuring the charge required to recharge the respective photodiodes to the predefined bias voltage, and repeating that process line by line. Low-noise application-specific integrated circuits (ASIC) are used to provide charge-sensitive amplification and multiplexing means to feed analog-to-digital converters (ADC) with bit depths of typically 14 bits or more. At this point, the image data is available for calibration corrections and clinical image processing.

Direct conversion flat detectors based on photoconductors represent an alternative way of X-ray detection [13,14]. Here, the photoelectron, generated during the X-ray absorption process, directly creates electron—hole pairs which are collected by applying an electric field across the detector material. While many semiconductors have been investigated as X-ray detection materials such as PbO, PbI₂, HgI₂, or CdTe, only a-Se has thus far made its way into clinical routine where large detector sizes are required.

A-Se can be deposited directly onto an active matrix of a-Si where each pixel consists of a charge collecting electrode and readout TFT. With its low K-edge of 12.6 keV, a-Se is well matched to the relatively soft X-ray spectra used in mammography (Table 1). To reach efficient charge collection, electric fields of about 10 V/μm are applied. High voltages of several kV are therefore required for material thicknesses of about 250 μm (mammography) or 1000 μm (general radiography), needed to reach adequate DQE [1]. This fact necessitates careful design to protect the TFT from being damaged by high voltage discharge.

3.2. Solid-state detectors for multi-slice CT

State-of-the-art full-body CT scanners are multi-slice spiral CT scanners. They come with different detector sizes and exhibit a large variety of detector rows of between 2 and 128 or even up to 320 in the scanning direction. Pixel sizes of typically 1 × 1 mm² lead to a spatial resolution of about 0.5 × 0.5 mm² at the isocenter. Higher resolution may be reached by a technique called flying focal spot, featuring a periodic motion of the X-ray focal spot in the scanning and/or rotational direction, which effectively doubles the sampling frequency of the respective spatial dimension(s).

Besides high X-ray detection efficiency, which is mandatory for all medical X-ray detectors, very short afterglow time is a key requirement of CT detectors due to sampling rates of up to 5 kHz needed in high-end multi-slice CT scanners with gantry rotation times of as short as 0.3 s. High-Z inorganic scintillators which are consistent with the short decay times required in CT scanners are, for example, gadolinium oxysulfide (Gd₂O₂S) or lutetium-based garnets. The overall detector design furthermore has to fulfill the high requirements for linearity, electronic noise and temporal stability.

In contrast with the flat detector case the comparatively large pixel sizes in CT allow a more discrete build of the scintillator material which is divided by septa in axial and angular directions, optically separating the individual pixels and spatially matching the photodiode structure below. The photodiodes are connected to front-end ASICs including the ADC function which provide digital data with bit depths of typically 20 bits.
Many moderately sized individual detector modules are assembled in segments to form the ring-like detector structure in the gantry opposite of the X-ray tube. Basic processing still occurs on the data acquisition system inside the gantry before the data are transmitted via slip rings to an external computing unit which is responsible for the CT data reconstruction.

CT detectors support different slice widths realized by binning methods, i.e., the combination of signals from neighboring pixels, so that optimum scanning speed, spatial resolution in the scanning direction and image noise can be achieved in accordance with the needs required by a given clinical application.

4. Prospects for X-ray detectors

While it is difficult to predict which developments or technologies will shape clinical routine in the future, some trends can be observed in the field of X-ray detectors. These developments address the improvement of established X-ray detector technologies with respect to performance and cost, as well as the evaluation of new technological approaches with the primary goals of providing even better performance, enabling new features or facilitating new applications.

4.1. Optimization of current X-ray detector technology

The performance of CT detectors can be enhanced by moving front-end readout electronics closer to the scintillator/photodiode structure or combining photodiode and readout circuitry on a single chip, optimizing electronic noise, complexity, and power consumption.

While CsI/a-Si-based flat detector technology is subject to continued performance and cost improvements, it is bound to achieve further success in some clinical fields where it has not already replaced older technologies. The advancements address all components of the detector. Optimized CsI scintillators, in particular with respect to their thickness, and improved low-noise readout electronics will further enhance DQE(f) and extend the quantum-noise limited regime toward even lower dose values.

An approach to improve signal-to-noise electronics performance can be realized by stacking the photodiode on top of the TFT and thereby maximizing the optical fill factor of the photodiode [15]. Higher ADC resolution of 16 bits or beyond will improve cone-beam CT imaging, further probing into the domain of CT, which is the prime standard for low-contrast X-ray imaging. Portable, wireless X-ray detectors profit in terms of size and weight from even higher electronic and mechanical integration. Furthermore, reduced power consumption will help making liquid cooling obsolete, required today in many high frame rate applications.

4.2. CMOS-based integrating X-ray detectors

Advances in CMOS imaging device technology, driven by consumer products such as high-end digital SLR cameras or mobile phones, and the continued trend to larger wafer sizes have paved the way to adopt CMOS technology in medical imaging. The basic detection principle for such devices is the same as the one described above for CsI/a-Si-based active matrix integrating detectors, however here, the a-Si photodiode and readout structure is replaced by a CMOS sensor.

While not achievable with amorphous Si, the electrical properties of crystalline Si allow realizing high performance analog or digital circuitry. This, for instance, makes on-pixel amplification possible, reducing readout electronic noise, which helps to move towards the goal of quantum-noise-limited X-ray detectors in the low dose regime. Other advantages are the non-destructive readout, allowing multiple readings of the signal, and the low power consumption of these devices. Depending on wafer technology, several orders of magnitude smaller structure sizes can be achieved, so that even more complex functions can be integrated on-pixel, while leaving ample space for the photodiode.

Besides on-pixel amplifiers other structures are possible in CMOS technology such as an analog memory in each pixel to realize global shutter function and hence allowing pixel matrix readout while the next frame is irradiated. This technique helps to decouple the requirements for X-ray integration and sensor read-out time in view of high frame-rate applications. Bi-plane angiography could profit, because this function would allow reading out one plane while irradiating the other plane without accumulating scatter radiation, which is a source of image-quality degradation.

Also other functions such as on-chip driving circuitry or analog-to-digital converters can be realized in CMOS technology, making external components with these functions obsolete.

In order to be successful in the medical X-ray detector arena, solutions have to be implemented to overcome some of the limitations of CMOS. They include tiling techniques to cope with limited wafer sizes in order to build large-area flat detectors, appropriate design or software solutions to suppress direct hits in the Si bulk material, and cost efficient manufacturing.

4.3. Photon counting X-ray detectors

Photon counting X-ray detectors for medical applications have attracted increasing attention in recent years [16–19], not least fostered by the developments of pixelated detectors for the innermost tracking devices of multi-layer detectors at high energy physics research centers such as CERN [20] or for macromolecular crystallography at synchrotron radiation facilities [21].

Unlike integrating detectors which measure the overall charge created by all absorbed X-rays during a given time interval, photon counting detectors enumerate each individual X-ray photon whose energy is above a predefined threshold. Hence, here the signal is formed by the number of X-ray photons, i.e., applying equal weight irrespective of the individual photon energy. A step further can be taken by defining multiple thresholds conditions and separating the photons from the broad clinical X-ray spectra into several energy bins.

The general structure of a counting pixel, in this example with four discriminator thresholds, is shown in Fig. 5. After amplification and pulse-shaping, the signal is compared to the different threshold. Hence, here the signal is formed by the number of X-ray photons, i.e., applying equal weight irrespective of the individual photon energy. A step further can be taken by defining multiple thresholds conditions and separating the photons from the broad clinical X-ray spectra into several energy bins.

![Fig. 5. Exemplary pixel scheme of a counting pixel, including charge amplification and signal shaping, four thresholds, adjustable by digital-to-analog-converters (DACs), respective discriminators and counters, and readout and control logic.](image-url)
discriminator thresholds which are provided by individual digital-to-analog-converters (DACs). If the signal is above a given threshold, the respective counter is incremented by one count. For every pixel the counters are read out at a given sampling frequency and reset afterwards to be prepared for the next X-ray acquisition. The complexity of counting pixel structures, realized in ASICs, may require up to thousands of transistors [20].

CdTe and CdZnTe (CzT) are very promising semiconductor materials in the field of X- and gamma-ray detection. The high atomic numbers of Cd and Te of 48 and 52, respectively, and the high densities of around 5.8 g/cm³ are well suited to provide good X-ray absorption. Other favorable properties include (i) a high bandgap energy of above 1.5 eV, allowing operation under room temperature conditions, (ii) pair creation energies of about 4.5 eV, generating in the order of ten thousand electron—hole pairs for typical diagnostic X-ray energies, and (iii) a sufficiently large electron mobility-lifetime product $\mu e_T$ of several $10^{-3}$ cm²/V.

Among the various crystal-growing techniques, the low-temperature traveling heater method (THM) has been shown to include a relatively small number of impurities and defects [22]. However, only small crystal slices with areas of only several cm² can be cut from an ingot which requires tiling a large number of CdTe slices to build X-ray detectors with sizes appropriate for clinical use.

Since the counting ASICs also come in relatively small sizes, a detector with a clinically relevant size will have to be built from several modules. Fig. 6 shows a schematic view of a CdTe-based photon counting detector made from such modules which are 4-side buttable, requiring the electric connections with the peripheral photon counting detector made from such modules which are 4-module counting sensor based on semiconductor technology. The interest in photon counting X-ray detectors based on direct conversion semiconductors such as CdTe or CZT is motivated by several potential benefits. Firstly, near-optimum detective quantum efficiencies $DQE(f)$ for the clinically relevant spatial frequencies, $f$, are expected due to the detector material properties mentioned above and the fact that the generated charges are collected by an applied electric field, leading to no considerable degradation of the modulation transfer function (MTF) when the material thickness is increased. Therefore an optimum CdTe or CZT thickness can be chosen so that high DQE is achieved even for the hard X-ray spectra occurring for a given clinical application (Table 1). Recalling the definition of $DQE(f)$ as

$$SNR_{out}(f) = DQE(f) \cdot SNR_{in}(f)$$

improved DQE can be directly invested in a reduction of dose which is proportional to the square of the input signal-to-noise ratio, $SNR_{in}(f)$, or improved image quality described by the output signal-to-noise ratio, $SNR_{out}(f)$. Secondly, the equal weighting of X-ray quanta in a pure counting scenario will lead to improved object contrast-to-noise-ratios (CNR) in all imaging cases where the low-energy part of the detected X-ray spectrum carries more contrast information which is the case for most imaging tasks. In case of energy-discriminating photon counting detectors, further CNR improvements can be achieved by applying optimum weight factors to the counts measured in the different energy bins [23]. And thirdly, energy-discriminating counting detectors may enable new clinical applications via material discrimination [24] or K-edge techniques and may be used for beam hardening corrections in CT or flat-panel cone-beam CT imaging.

The challenges to master, however, are considerable and only some are mentioned in the following. Material performance limitations have to be overcome stemming from polarization effects generating temporal instabilities or from inhomogeneities due to dislocations or Te inclusions. Depending on pixel size, pulse splitting due to charge-sharing between pixels or K-fluorescence [25] may require next-neighbor anti-coincidence and energy summing schemes in particular for energy-discriminating photon counting detectors. Solutions to cope with pulse-pileup occurring at high X-ray fluxes have to be implemented. For some applications and detector designs, reliable butting and TSV techniques may be required while keeping the real estate needed for that purpose negligible [26]. Not least ambitious is a realization at an acceptable cost level. Despite these challenges, progress has been made in some application areas and initial results from various prototype detectors or systems have been reported [17,18,27].

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References


