13

Photon-Counting Detectors for X-ray Imaging

Michela Esposito

CONTENTS
13.1 Introduction .......................................................................................................................................................... 239
13.1.1 From High Energy Physics to Medical Imaging ......................................................................................... 240
13.1.2 Basic Principles of Photon Counting Detection ......................................................................................... 240
13.2 Semiconductor Materials for PCDs ...................................................................................................................... 241
13.2.1 Fluorescence ..................................................................................................................................................... 242
13.2.2 Charge Trapping .............................................................................................................................................. 243
13.2.3 Polarization ...................................................................................................................................................... 244
13.2.4 Technological Limitations ............................................................................................................................. 244
13.3 ASIC Design .......................................................................................................................................................... 244
13.3.1 General Principles for ASIC Designs ........................................................................................................... 244
13.3.2 Dead Time and Pile-Up .................................................................................................................................... 244
13.3.3 Charge Sharing ............................................................................................................................................... 246
13.4 Review of Hybrid Pixel Detectors for Photon Counting Imaging ................................................................. 247
13.4.1 Medipix ........................................................................................................................................................... 247
13.4.2 Medipix2 .......................................................................................................................................................... 247
13.4.3 Timepix ........................................................................................................................................................... 248
13.4.4 Medipix3 .......................................................................................................................................................... 250
13.4.5 Large Area Medipix-Based Imaging Systems ............................................................................................... 251
13.4.6 Pilatus .............................................................................................................................................................. 252
13.4.7 XPAD ............................................................................................................................................................. 254
13.5 Conclusions .......................................................................................................................................................... 254
References .................................................................................................................................................................. 255

13.1 Introduction

From the first radiograph ever produced of Frau Röntgen’s hand (Röntgen 1896a,b) (see Section II, Chapter 17 of this book), to most of the X-ray detectors commonly used today in medical imaging, X-ray detection devices are based on energy integrating technologies (Yaffe and Rowlands 1997; Rowlands and Yorkston 2000; Russo 2004; Esposito et al. 2011a; Esposito et al. 2014). When energy integrating detectors (EIDs) are irradiated with polychromatic X-ray sources, they record a signal proportional to the sum of the energy of every photon, thus the information carried by the energy of individual photons is lost. In energy integrated images, high energy photons weigh more than low energy ones, the latter creating higher contrast between tissues, and Poisson noise from high energy photons is enhanced. Additionally, dark current and noise sources (electronic noise, Swank noise, etc.) are added to the signal (see Section I, Chapter 14 of this book). As a result, EIDs are limited in terms of signal-to-noise ratio (SNR) and dynamic range.

Photon-counting detectors (PCDs) have the potential to overcome these issues by processing incoming X-ray photons individually. By means of dedicated electronic circuits, based on the use of one or more detection thresholds, PCDs can process the signal generated by individual photons, recording either deposited energy (spectroscopic systems) or an incremental count (simple photon counting systems). For the latter, each photon contributes with the same weight to the image formation, while, for the former, each photon can be weighted proportionally to its energy, thus resulting in contrast enhancement. Additionally, an appropriate choice of detection thresholds for PCDs allows for full noise rejection, leading to higher SNR, allowing for long acquisition times, and resulting in an extended dynamic range.

The improvement in SNR from integrated imaging to simple photon counting has been shown for a mammographic phantom in a prototype setup to be as high as 1.1 for high contrast details (breast tissue/calcifications) and 1.2 for low contrast details (breast tissue/adipose tissue) (Giersch et al. 2004). For spectroscopic detectors, the use of energy weighting factors, such as weighting the contribution of each photon to the final image by an energy dependent factor (Tapiovaara and Wagner 1985; Giersch et al. 2004; Karg et al. 2005), can lead to a further
improvement in SNR (an enhancement of a factor 1.2 and 1.4 for mammographic high and low contrast details, respectively) and has the potential to reduce dose (Giersch et al. 2004).

Nonetheless, it is of note that, for PCDs, the enhancement in SNR due to single quantum processing is strongly dependent on the assumption of having a PCD with a zero dead time, as an increase in dead time would lead to a decrease in SNR (Alvarez 2014). Additionally, PCDs can provide means to reject Compton events scattered at large angles (Pedersen et al. 1997), leading to a reduction in dose when compared to mammographic imaging systems using anti-scatter grids (Säbel and Aichinger 1996).

A further advantage of using PCDs for X-ray imaging, compared to EIDs, is related to the different conversion and signal detection mechanisms these two technologies rely on.

In fact, most EIDs are indirect detection systems, where incoming radiation is converted into visible light by means of a scintillating material and, subsequently, the generated optical photons are detected by photodiodes in the pixelated detector. PCDs, in the majority of cases, work as direct detection systems, although some examples of PCDs coupled to scintillators do exist (Dierickx et al. 2016). Direct detection is a single-stage process, where incoming radiation generates electron-hole (e-h) pairs in the semiconductor detector volume, which are then collected at the detector electrodes. When comparing these two approaches to image formation, it is of note that scintillating materials require a higher energy for generation of optical photons than commonly used semiconductors for e-h pairs generation, resulting in a weaker signal that, together with optical losses within the scintillator material and at the interface with the sensor, lead to a reduction in the achievable SNR. Furthermore, optical photons spread laterally as they cross the scintillator and the amount of spread depends on the depth of interaction, introducing an energy dependent blurring or loss of spatial resolution (Evans et al. 2002; Borasi et al. 2003; Menuin et al. 2005).

Spectroscopic capabilities in PCDs can also allow for new quantitative X-ray imaging modalities, such as K-edge X-ray imaging, where the dependence of X-ray attenuation on energy can lead to material identification (Roessl and Proksa 2007), K-edge imaging with simultaneous contrast media (Schlomka et al. 2008; Taguchi and Iwanczyk 2013), and molecular tomography as a means for functional computed tomography (CT) imaging, by using heavy metal nanoparticles bound to targeted molecules (Jaffer and Weissleder 2004; Hyafil et al. 2007).

In the last two decades, a number of PCDs have been developed for medical imaging applications and, although most of the PCDs currently available are benchtop experimental setups, some PCDs have been commercialized (Åslund et al. 2007) or tested in the clinical environment (Iwanczyk et al. 2009). The most common technological choice for photon counting imaging is hybrid pixel detectors, although some other options are available including edge-on silicon strip detectors (Beuvillé et al. 1998; Xu et al. 2013a,b). The rest of the chapter will focus on the design choices, performance, and limitations of hybrid pixel detectors as PCD for X-ray imaging. Properties and limitations of semiconductor sensors for PCDs will be reviewed in Section 13.2. The rationale behind electronics design for PCDs and a review of current designs will be presented in Section 13.3. An outlook for future developments and challenges related to PCDs will be discussed in Section 13.4.

13.1.1 From High Energy Physics to Medical Imaging

Hybrid pixel detectors were originally developed for High Energy Physics experiments, to provide two-dimensional position sensitive detectors, i.e., pixelated detectors, for collider experiments. In fact, linear detectors, such as silicon strip detectors, were limited by ambiguities in identifying event positions at high occupancy (Anzivino et al. 1988). When using linear detectors, at least two orthogonal layers are necessary to identify the position in space where an event has occurred. The need for unambiguous event identification at high luminosity triggered the development of pixelated detectors, which required a complete electronic chain in each pixel. The level of miniaturization required, together with the need to create an electronic contact between the semiconductor pixels and their own electronics, were the main technological challenges to be addressed in the development of hybrid pixel detectors. The first hybrid pixel detector was developed at CERN (Conseil Européen pour la Recherche Nucléaire, Geneva, Switzerland) in 1991 by the RD19 collaboration (Heijne et al. 1988, 1994), to be used for High Energy Physics experiments (WA97 (Alexeev et al. 1995) and DELPHI (Becks et al. 1997) experiments). Between 1998 and 2006, four experiments (ATLAS, CMS, ALICE, and LHCb) were installed on the Large Hadron Collider at CERN, and had hybrid pixel detectors, totaling more than 100 million pixels (Delpierre 2014). The advantages of this detector technology, such as full noise rejection, high readout speed, high detection efficiency, and being capable of energy selection, made hybrid pixel detectors attractive for medical and biological applications. Consequently, a number of hybrid pixel detectors dedicated to X-ray imaging were developed.

13.1.2 Basic Principles of Photon Counting Detection

A photon counting hybrid pixel detector is a two-dimensional semiconductor array of pixelated anodes, which act as microscopic sensitive elements, namely pixels. Each sensitive element is connected to its individual readout electronic chain, provided by an application-specific integrated circuit (ASIC). A hybrid pixel detector consists of two superimposed layers. The top layer is the detecting material or sensor, in which X-rays can interact and their interaction is detected. The bottom layer is the readout electronics, and defines the segmentation and pixel pitch of the PCD. Sensor material and readout electronics are processed on two different substrates, and are electrically connected via the so-called bump-bonding and flip-chip techniques, realized by placing a drop (with a diameter in the order of 10–20 µm) of solder material (e.g., In, PbSn, Au) between two metal pads attached to sensor and ASIC. Each pixel of a hybrid pixel detector needs individual bump-bonding. A cross-sectional view of a hybrid pixel detector is shown in Figure 13.1, comprising a pixelated semiconductor sensor and a readout ASIC (Medipix2
241

Photon-Counting Detectors for X-ray Imaging

chip). Each pixel cell of the readout ASIC features a complete signal processing cell. The readout of the processed digital signal from each individual pixel is driven by dedicated circuits placed at the chip periphery. When radiation quanta deposit their energy in a semiconductor detector material, e-h pairs are created as a result of ionization. Free electrons and holes are separated in the detection material by an externally-applied electric field. As a result of this, charge carriers drift and diffuse towards their respective pixel electrode and possibly their neighbors. This electric signal, proportional to the number of electron-hole pairs generated in the sensor volume, is then processed by the pixel readout electronics.

13.2 Semiconductor Materials for PCDs

Hybrid pixel detectors for High Energy Physics experiments have traditionally relied on the use of Silicon (Si) as detection material, because of its relatively low mass (Z = 14), uniformity across large wafers, reliability, and cost. However, the low energy of diagnostic X-ray spectra (20–120 keV) represents a challenge for thin (300 mm–1 mm) Si sensors. For this reason, a range of both elemental and compound room-temperature semiconductors have been proposed as detecting media for hybrid pixel detectors. These include, gallium arsenide (GaAs) (Hamann et al. 2015), cadmium telluride (CdTe) (Seller et al. 2011; Koenig et al. 2012), cadmium zinc telluride (Wilson et al. 2007; Barber et al. 2015), high purity germanium (HPGe) (Pennicard et al. 2013, 2014), and mercury iodide (HgI2) (Schieber et al. 1999; Liao et al. 2013).

Compound semiconductors in particular, first used in 1945 for the detection of alpha particles and X-rays with silver chloride (AgCl) crystals (Van Heerden and Milatz 1950), have the potential to be grown with properties, such as a high atomic number and a wide energy bandgap, to suit the specific application they are designed for. Nonetheless, growth of high performance compound semiconductors is still limited by difficulties in producing chemically pure and structurally perfect crystals (Del Sordo et al. 2009).

Table 13.1 lists the main physical properties of elemental and compound semiconductors used in PCDs.

<table>
<thead>
<tr>
<th>Material</th>
<th>Si</th>
<th>GaAs</th>
<th>CdTe</th>
<th>Cd0.9Zn0.1Te</th>
<th>HPGe</th>
<th>HgI2</th>
</tr>
</thead>
<tbody>
<tr>
<td>Atomic numbers</td>
<td>14</td>
<td>31, 33</td>
<td>48, 52</td>
<td>48, 30, 52</td>
<td>32</td>
<td>80, 53</td>
</tr>
<tr>
<td>Density (g/cm³)</td>
<td>2.33</td>
<td>5.32</td>
<td>6.20</td>
<td>5.78</td>
<td>5.33</td>
<td>6.4</td>
</tr>
<tr>
<td>Bandgap Eᵢ (eV)</td>
<td>1.12</td>
<td>1.43</td>
<td>1.44</td>
<td>1.57</td>
<td>0.67</td>
<td>2.13</td>
</tr>
<tr>
<td>Electron lifetime-mobility τₑμₑ (cm²/Vs)</td>
<td>&gt;1 10⁻⁵</td>
<td>10⁻³</td>
<td>10⁻³–10⁻²</td>
<td>&gt;1 10⁻⁴</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hole lifetime-mobility product τₕμₕ (cm²/Vs)</td>
<td>&lt;1 10⁻⁶</td>
<td>10⁻⁴</td>
<td>10⁻⁵</td>
<td>&gt;1 10⁻⁵</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Ionization energy W (eV/e-h pairs)</td>
<td>3.62</td>
<td>4.2</td>
<td>4.43</td>
<td>4.6</td>
<td>2.96</td>
<td>4.2</td>
</tr>
<tr>
<td>Resistivity (Ωcm)</td>
<td>10⁴</td>
<td>10⁷</td>
<td>10⁹</td>
<td>10¹⁰</td>
<td>50</td>
<td>10¹³</td>
</tr>
</tbody>
</table>

Image quality in X-ray imaging, and specifically detection quantum efficiency (DQE) (Dainty and Shaw 1974), largely depend on two parameters: intrinsic detection efficiency (DE) and charge collection efficiency (CCE). The former is a combination of absorption efficiency and thickness of the material used as the detection medium. The latter is defined as the ratio between the effective amount of charge collected, per interaction, at the readout electrodes and the amount of charge deposited by the radiation quantum interacting in the detector sensitive volume. To maximize the DQE, both DE and CCE need to be as close as possible to 1.
Figure 13.2a shows the intrinsic DE for a number of semiconductor materials of commonly used thicknesses at diagnostic energies (20–120 keV). For a 1-mm thick Si sensor, DE drops sharply for X-ray energies higher than 15 keV, with a DE of 40% at 25 keV (mammography) and below 5% at 80 keV (general radiology). High-Z semiconductors offer a much higher DE over the diagnostic range: all the high-Z materials reported in Figure 13.2a have a DE close to 100% at 25 keV, while still offering a relatively high DE at higher energies (83% for CdTe, 30% for Ge, 13% for GaAs at 80 keV), making them well suited in the whole diagnostic range.

In addition to their higher cross-section for photon absorption, high-Z sensors have the advantage of a low cross-section for Compton scattering in the diagnostic range. Figure 13.2b shows the cross-section for photoelectric effect and Compton scattering for CdTe and Si. For Si sensors, Compton scattering is the dominant mechanism of interaction from X-ray energies higher than 15 keV, while for CdTe sensors, the photoelectric effect is dominant up to ~250 keV. The Compton effect is an incoherent scattering process between a photon crossing a medium and quasi-free atomic electrons. Energy and momentum lost by the interacting photon is transferred to the electron, which is emitted at a given angle while the photon is scattered from its original trajectory. There is a maximum kinetic energy that can be transferred in a Compton collision (the so-called “Compton edge”), which corresponds to scattering events where the photon is backscattered and the recoil electron is emitted in the direction of the impinging photon. Compton scatter events deteriorate both spectroscopic and imaging performance of the detection system. In fact, photons undergoing inelastic scattering will deposit a variable amount of energy in the detector, leading to the creation of low energy tails (up to the Compton edge) in the measured spectrum. Additionally, Compton-scattered photons can undergo other interactions in the detector volume and be absorbed in a location different from where the initial interaction took place, resulting in decreased spatial resolution. Hence, high-Z detectors, for which the photoelectric effect is the dominant process at diagnostic energies, have the potential to offer better spectroscopic and imaging performance.

### 13.2.1 Fluorescence

Although a high cross-section for photoelectric effect has advantages in terms of imaging performance, fluorescence X-rays emitted as a result of photoelectric effect can be a limiting factor for high-Z sensors. When an X-ray of energy $h\nu$ undergoes the photoelectric effect, the interacting photon is absorbed in the interaction with the atom, which in turns emits an electron with kinetic energy $K_e = h\nu - B_e$, with $B_e$ being the binding energy of the atomic electron (see Section I, Chapter 1 of this book). However, when electrons are ejected from an atomic shell, a vacancy is created within that atomic shell, leaving the atom in an excited state. This triggers a readjustment of the electrons from other shells to fill the vacancy, leading either to the atom emitting one or more Auger electrons (non-radiative transition) or to the emission of characteristic fluorescence X-rays (radiative transition).

While Auger electrons can travel a very short path within the detector, fluorescence X-rays can have a much longer mean free path, resulting in considerable changes in the spatial distribution of the charge cloud, thus affecting spatial and spectroscopic performance, especially for fine pixelated detectors. The probability of non-radiative transition, with emission of Auger electrons, is high for low-Z materials and decreases with atomic number Z. At $Z \approx 30$, the probability of radiative and non-radiative transition is almost equal. The probability of emitting fluorescence photons per K-shell vacancy, namely the K-fluorescence yield, is

---

*Atomic electrons are defined as quasi-free when, to a first approximation, their binding energies do not affect the interaction, and can be neglected in calculations.*
shown in Figure 13.3 as a function of the atomic number, calculated by means of an empirical formula proposed by Hagedoorn and Wapstra (1960). Values for some semiconductors commonly employed in PCDs, are reported in Figure 13.3. While Si has a moderate K-fluorescence yield (4%), high-Z materials are strongly affected by fluorescence emission. Fluorescence yield is above 50% for GaAs and above 80% for CdTe. An additional parameter to consider when evaluating the effect of fluorescence X-rays in hybrid pixel detectors is the energy of the emitted photons.

Fluorescence photons are emitted with an energy corresponding to the difference between the outer and the inner atomic shells interested by the readjustment of the atomic electrons, as a consequence of a photon interacting via photoelectric effect. For high-Z materials, this energy difference, and thus the energy of the emitted photons, is relatively high (\( \sim 10 \text{ keV for Ga and As,} \sim 23 \text{ keV for Cd and} \sim 27 \text{ keV for Te} \)), compared to just 1.7 keV for Si. The relatively high energy of these characteristic X-rays results in a long mean free path: (\( \sim 40 \mu\text{m for Ga,} \sim 50 \mu\text{m for As,} \sim 110 \mu\text{m for Cd, and} \sim 60 \mu\text{m for Te} \)) (Tlustos 2005). Mean free paths for fluorescence X-rays in high-Z materials are comparable to commonly used pixel pitches (\( \sim 50 \mu\text{m} \)), resulting in an appreciable deterioration of spatial and spectroscopic imaging properties of PCDs.

### 13.2.2 Charge Trapping

Another significant effect limiting the performance of high-Z semiconductors in PCDs is charge trapping. Semiconductor defects can temporarily trap electrons and holes as they drift towards the electrodes, resulting in a reduction of CCE. Charge trapping also entails that X-rays interacting at different depths in the sensor produce a different amount of induced charge, due to the different fraction of charge trapped along the path to the electrodes.

![Figure 13.3](image_url)

**Figure 13.3** K-fluorescence yield as a function of the atomic number Z calculated by means of an empirical formula proposed by Hagedoorn and Wapstra (1960). Values for some semiconductors used in hybrid pixel detectors are also reported.

![Figure 13.4](image_url)

**Figure 13.4** CCE for several sensor materials (Si, CdTe, and GaAs) as a function of the normalized interaction position, calculated using the Hecht equation. Values of lifetime-mobility product are from Table 13.1. Typical values for electric field and detector thickness were used. An electric field E of 3000 V/cm was assumed for Si and CdTe, while for GaAs a value of 6000 V/cm was used. Detector thicknesses were 1 mm for Si and CdTe and 0.5 mm for GaAs.

High-Z semiconductors are largely affected by charge trapping, with a lifetime for both electrons and holes several orders of magnitude lower than Si. However, charge trapping also depends on the time needed for charge to be collected, and effectively the parameter that governs CCE is the mobility-lifetime product \( \mu \tau \). Values of mobility-lifetime product for several elemental and compound semiconductors are reported in Table 13.1.

For a planar detector of thickness, T, subject to a uniform electric field, E, and neglecting charge de-trapping, CCE, as the ratio between the collected charge, Q, and the photo-generate charge, \( Q_0 \), is given by the Hecht equation (Hecht 1932; Knoll 2010, p.480)

\[
\text{CCE} = \frac{Q}{Q_0} = \frac{\lambda_e}{T} \left[ 1 - e^{-\frac{x}{\lambda_e}} \right] + \frac{\lambda_h}{T} \left[ 1 - e^{-\frac{T-x}{\lambda_h}} \right],
\]

where \( \lambda_e = \mu_e \tau_e E \) and \( \lambda_h = \mu_h \tau_h E \) are the mean drift lengths for electrons and holes, respectively, and x is the interaction position, that is, the point in space where charge deposition occurs. Values of CCE for Si, CdTe, and GaAs are shown in Figure 13.4 as a function of the normalized interaction position, x. Typical values of detector thickness, T, and electric field, E, were used: a sensor thickness of 1 mm and an electric field of 3000 V/cm were used for Si and CdTe, while a thickness of 0.5 mm and an electric field of 6000 V/cm were assumed for GaAs. While Si shows a CCE very close to 100% with a mean drift length \( \lambda_e \sim \lambda_h \sim 3 \times 10^4 \text{ cm} \), CdTe (\( \lambda_e \sim 3 \text{ cm}, \lambda_h \sim 0.3 \text{ cm} \)) has a CCE close to 100% for charge deposited near the collection electrodes (\( x = 0 \)) which then decreases to \( \sim 85\% \) for \( x = T \). Among the materials compared in Figure 13.4, GaAs shows the lowest CCE, ranging between
~70% and ~15%, with a mean drift path of $\lambda_e \sim 0.06$ cm and $\lambda_h \sim 0.006$ cm for electron and holes, respectively.

Strategies to mitigate charge trapping by shortening the time needed for full charge collection include reducing the sensor thickness (which also results in a reduction of the DE), increasing the electric field within the sensor, and using semiconductors with high charge mobility (Pennicard et al. 2011). Pixel geometry also has the potential to mitigate the effects of charge trapping. In fact, when pixels are small compared to the sensor thickness, the weighting field that surrounds small pixels, conceptualized by the Shockley-Ramo theorem (Shockley 1938; Ramo 1939), is of considerable magnitude only close to the pixel defects. Thus, charge carriers will induce a significant signal to the electrodes when they are in close proximity. In a sensor irradiated from the cathode, the signal will be generated only in the first few micrometers, far from the pixel anodes. If a material has much lower electron trapping, which is the case in Cd(Zn)Te for example, the holes will drift to the cathode without inducing a significant signal at the anodic electrodes. Therefore, the signal will be almost exclusively induced by the electrons, effectively overcoming limitations arising from charge trapping in high-Z semiconductors. In addition to mitigating charge trapping, small pixel detectors also will have a fast signal rise-time, allowing for a fast-analog readout without ballistic deficit effects.* This effect is referred to as the “small pixel effect” (Sellin 1999; Wilson et al. 2007). However, while small pixels are desirable to mitigate a charge trapping mechanism, there is a trade-off between advantages related to the small pixel effect and disadvantages related to charge sharing in fine pixelated detectors (see Section 13.3.3).

13.2.3 Polarization

Polarization is a phenomenon occurring in semiconductor detectors leading to a time-dependent decrease in CCE, affecting both pulse height spectrum and count rate (Siffert et al. 1976; Szeles et al. 2007). It is thought to be related to trapping and re-trapping of the charge carrier, which produces alterations in the space-charge distribution and in the electric field profile across the detector (Malm and Martini 1974; Siffert et al. 1976; Niraula et al. 2002). It has, however, been reported that, with operation at high bias voltage and low temperature, it is possible to mitigate polarization effects (Niraula et al. 2002).

13.2.4 Technological Limitations

High-Z semiconductor materials, offering a high DE in the diagnostic energy range, could be the detector of choice for PCDs in medical imaging. However, several technological limitations represent a bottleneck in the development of large area and high count rate high-Z PCDs. Single crystals CdTe wafers are currently limited to 3’ wafers for a 1-mm thickness, and also suffer from poor uniformity due to tellurium inclusions and dislocations, and require a low temperature bump-bonding process (Szeles 2004). CdZnTe, despite offering a resistivity of one or two orders of magnitude higher than CdTe, thus a much reduced leakage current and a lower degree of polarization compared to CdTe (Del Sordo et al. 2009), is still limited by the presence of extended defects (including secondary phases, sub-grain boundaries, dislocations). This leads to degradation in charge transport properties and uniformity of response, resulting in a decreased energy and spatial resolution (Parker et al. 1999; Bolotnikov et al. 2007; Hossain et al. 2010).

GaAs has the advantage of delivering a low leakage current, because of its high bandgap energy, but it is significantly affected by a high concentration of defects leading to charge trapping and, consequently, to a reduction of the CCE (Ayzenshtat et al. 2001) (see Figure 13.4). Ge, conversely, can be produced in high purity crystals, but its operation requires cooling (~70°C) to limit leakage current due to the low bandgap energy. Additionally, pixilation and bump-bonding are procedures not fully developed for Ge sensors (Pennicard et al. 2013, 2014).

13.3 ASIC Design

13.3.1 General Principles for ASIC Designs

Although a number of different ASIC circuits exist for pixel readout of PCDs, some basic design principles can be highlighted. A schematic diagram of a typical pixel cell for a hybrid pixel detector is shown in Figure 13.5. The signal collected at the pixel anode is amplified by a charge sensitive preamplifier. The amplified signal reaches a shaper which shapes the signal with a given time constant and also works as a low-pass filter, to improve SNR. N adjustable discriminators and N counters are provided for signal discrimination. When the shaped signal exceeds the discriminator threshold, the relevant counter is increased by one. This information is stored in the digital section of the pixel cell until it is readout by the off-chip readout electronics. The number of counts in a given energy window, defined by the values of two consecutive discriminator thresholds, is obtained by subtraction of the counts registered by the two discriminators. If a double threshold is used per each energy window, such as a high and low threshold, the achievable energy windows are N/2, with N being the number of discriminators. Conversely, if only a low threshold is used for each energy bin, the achievable number of energy windows is N. In addition to global threshold values, a per-pixel fine tuning of thresholds is generally implemented by means of in-pixel digital-to-analog-converters (DACs), in order to minimize pixel-to-pixel variations, due to mismatches and non-uniformity in the manufacture process. The use of N thresholds is the most common choice and usually the fastest approach to energy discrimination. A different approach to spectroscopic imaging with PCDs is provided by Time-over-Threshold (ToT) measurements, which relies on the proportionality between energy and width of the amplified pulse. This architecture, implemented in the Timepix (Llopart et al. 2007), Timepix3 (Poikela et al. 2014), and Dosepix (Zang et al. 2015) detectors, will be further discussed in Section 13.4.3.

13.3.2 Dead Time and Pile-Up

Pulse pile-up represents a serious limitation for PCDs in the high flux conditions (10⁹ photons mm⁻² s⁻¹) required for X-ray imaging.
Photon-Counting Detectors for X-ray Imaging

imaging (Taguchi et al. 2011; Roessl et al. 2016). Given that pulse processing requires a finite time to be completed, pulse pile-up occurs when a second photon is detected in a pixel while the signal from the first photon is being processed. Depending on the time at which the second photon is detected with respect to the first one, peak pile-up or tail pile-up events can occur.

For a peak pile-up event, two photons are detected in quasi-coincidence, resulting in a single event been detected at a higher energy. When the second detected photon arrives on the tail of the preceding pulse (tail pile-up), it will distort the shape of the pulse, recording a lower energy in the case of bipolar-shaped pulses and higher energy for unipolar-shaped pulses (Wielopolski and Gardner 1976). Effectively, both peak pile-up and tail pile-up result in loss of counts (known as dead time losses), a deviation from the linear behavior for count rate, and deteriorated spectroscopic performance (Frojdh et al. 2014). Two analytical models have been proposed to describe the dead time losses based on the pulse processing electronics used, referred to as paralyzable and non-paralyzable models (Knoll 2010, p.119).

Response of so-called paralyzable and non-paralyzable systems to pile-up events are show in Figure 13.6a. Each of these two systems has a characteristic dead time ($\tau$), due to the amount of time needed for the electronics to process the detected pulse. Given a certain arrival time of incoming photons (shown in the top inset of Figure 13.6a), events occurring in the dead time of the previous event are not accounted for in paralyzable systems (middle inset of Figure 13.6a), and the dead time is extended by $k\tau$, with $k$ being the number of events occurring in the dead time of the first detected pulse. Non-paralyzable systems, on the other hand, still disregard pile-up events, but are not affected by these in terms of dead time (bottom inset of Figure 13.6a). For both systems, however, pile-up events produce dead time losses. In the
example of Figure 13.6a, only half of the events interacting in the detector can be detected by the two systems. Following from this model, dead time losses can be modeled as a function of the true event rate, $n$, and detector characteristic dead time, $\tau$ (Wielopolski and Gardner 1976; Yu and Fessler 2000; Knoll 2010, p. 119). The detected count rate, $m$, for a non-paralyzable system can be written as $m = n/(\tau + 1)$, while this quantity becomes $m = ne^{-\tau}$ for paralyzable systems. Detected count rates as a function of the true event rate, in the case of paralyzable and non-paralyzable systems, are displayed in Figure 13.6b for a dead time, $\tau = 600$ ns, and also compared with the behavior of an ideal system with zero dead time. For very low values of the true event rate, small differences are visible among the three systems, but, as the true event rate increases, both paralyzable and non-paralyzable systems show a dead time loss, with an underestimate of the event rate. Non-paralyzable systems approach an asymptotic value for true event rate, $n = 1/\tau$, at which the detector just finishes one dead period, due to an event detection, before the next dead period starts. For paralyzable systems, on the other hand, the detected count rate increases up to a certain maximum and then decreases, due to the multiple extensions of the dead time following an initially detected event. Techniques to correct for dead time losses, both hardware and software-implemented, have been developed (Upp et al. 2001; Westphal 2008).

The dead time of a readout chain is strictly linked to the so-called peaking time, that is the time needed for a shaped signal to go from baseline to peak, which in turn depends on the timing constant of the shaper and the charge carrier drift time. Assuming a symmetrical shape for the shaper output, the theoretical minimum dead time ($\tau$) is twice the peaking time; however, this value tends to be larger in practice. A simple strategy to reduce dead time and, thus, increase the efficiency of the detector at high fluxes, is reducing the peaking time. However, peaking time cannot be reduced arbitrarily. In fact, peaking time should include the time required to collect all the charge generated by ionization of primary and secondary X-rays, including fluorescence X-rays. Another simple approach to reduce the dead time is reducing the pixel size, as every pixel will be exposed to a reduced event rate. However, this approach could easily lead to degradation of the spectroscopic performance due to charge sharing (see Section 13.3.3) (Ballabrigo et al. 2016).

Other options to increase the counting efficiency at high fluxes include specific ASIC design solutions. The Pilatus3 ASIC (see Section 13.4.6) features an instant retriggersing technology (Loeliger et al. 2012), developed to overcome the dead time losses due to paralyzation observed in its predecessor Pilatus2 (Trueb et al. 2012). The instant retriggering scheme implements the idea of overcoming paralyzation due to pile-up by retroactively partitioning the pulse into ideal single photon pulses. A more detailed description of this architecture can be found in Section 13.4.6. It is of note, however, that the instant retriggering scheme is applicable to monochromatic sources only (synchrotron sources), as it requires a fixed time interval for resampling (or dead time), which is given by the width of a single monochromatic photon pulse.

A different approach to overcome paralyzation due to high fluxes is related to the combined use of low and high thresholds (Kraft et al. 2012), based on the observation that high fluxes paralyze high level threshold. The method involves combining information from conventional thresholds with measurements from a last threshold placed at an energy higher than the end-point of the X-ray spectrum, which will paralyze only at very high count rates. This approach has the advantage of providing full resolution spectroscopic data in low flux regions and count rate information for high fluxes. A further solution, based on the use of clocked discriminators (Gustavsson et al. 2012), was proposed for use in edge-on silicon strip detectors (Liu et al. 2015). After detection of a hit by a low threshold, the system is sampled every clock cycle for a certain amount of time. If the signal exceeds any threshold, a digital register is set and the counter corresponding to the highest threshold exceeded is recorded. Following reset, the system is ready to accept a new pulse. This architecture effectively works as a peak finder. When high fluxes are expected, once the peak of the first incoming photon has been found, the shaper can act as a filter and reset its own signal to the baseline so that there is no reminiscence of the previous events, and successive events can be detected within a single readout cycle. The dead time in this case is given by the number of system clocks occurring between event detection and pulse reset, proportionally to the height of the shaped pulse and, thus, detected energy. This pixel architecture, effectively, makes the system non-paralyzable, with a reduced dead time compared to conventional architectures: a 60-ns dead time has been reported with a 40-ns peaking time (Gustavsson et al. 2012).

### 13.3.3 Charge Sharing

As X-rays interact in the semiconductor volume, a charge cloud is generated and drifts towards the collection electrodes in the electric field produced by the applied voltage. As the charge cloud migrates, it increases its size due to diffusion and effects of the Coulomb force. When a photon interacts closer to the edges of a pixel, charge can be split (shared) across multiple adjacent pixels, each of which detects a signal lower than the original one. In addition to diffusion, other physical effects contributing to charge sharing are: range of primary and secondary particles, fluorescence X-rays, and bremsstrahlung radiation. Although charge sharing can be an advantage for particle detection, as it can improve spatial resolution (Esposito et al. 2011b,c) and allow for tracking and particle identification (Bouchami et al. 2011; Urbal et al. 2011; Gwosch et al. 2013), in the field of X-ray imaging, charge sharing has been shown to produce a degradation in energy resolution (Norlin et al. 2006; Ponchut 2008; Russo et al. 2009), and it has been considered a limiting factor in developing energy sensitive X-ray imaging detectors with high spatial resolution (Nilsson et al. 2007; Marchal et al. 2010). Several offline correction schemes have been proposed to mitigate the effect of charge sharing (McMullan et al. 2007; Nilsson et al. 2007; Jakubek et al. 2008a; Esposito et al. 2011b,c). Hardware implementations of charge sharing correction algorithms are also available, and they are based on charge summation schemes, which are able to perform summation of charge spread across adjacent pixels. A charge summation scheme was implemented for the first time in the Medipix3 chip (Ballabrigo et al. 2011), and subsequently implemented for the PIXIEIII (Bellazzini et al. 2015), X-Counter PC (Ullberg et al. 2013), and the AGH_Fermilab (Maj et al. 2015) detectors. The charge summation pixel architecture of Medipix3 will be further discussed in Section 13.4.4.
13.4 Review of Hybrid Pixel Detectors for Photon Counting Imaging

A review of some hybrid pixel detectors for photon counting X-ray imaging is provided in this Section. This list should not be considered exhaustive, as the purpose of this review is to highlight challenges, design solutions, and, ultimately, the technological development path, which can lead to achieve a clinically usable photon counting imaging system. For a more comprehensive list of the PCDs available, the reader is referred to Table 13.2.

13.4.1 Medipix1

The first Medipix collaboration (Medipix1) was established in 1997 between CERN, the University of Glasgow, the University of Freiburg, and the University and Istituto Nazionale di Fisica Nucleare (INFN) of Naples and Pisa, to develop the first hybrid pixel detector dedicated to medical imaging. The resulting circuit was a small prototype detector consisting of $64 \times 64$ pixels with a 170 µm pitch. Each pixel was equipped with a single discriminator for energy thresholding with a 3-bit adjustment threshold, and a 15-bit counter (Campbell et al. 1998). The chip was read out by a custom I/O interface (MUROS-1) (Bardelloni et al. 2000). Several proofs of concept for use in medical and biological applications were provided by assembling the Medipix1 circuit on both GaAs (Schwarz et al. 1999; Abate et al. 2001; Amendolia et al. 2001) and Si sensors (Fornaini et al. 2001; Ponchut et al. 2002; Tlustos et al. 2003).

13.4.2 Medipix2

Following the successful proof of concept of Medipix1, the Medipix2 collaboration was formed in 1999 to build a high contrast and high resolution PCD for medical imaging. The Medipix2 chip was built on a $256 \times 256$-pixel matrix with a 55 µm pitch. Benefiting from the fast progress in Complementary Metal-Oxide Semiconductor (CMOS) technology, which made it possible to achieve enhanced pixel functionalities together with a smaller feature size, the Medipix2 chip, fabricated using the IBM 0.25 µm technology, was able to incorporate a more complex electronic chain in a smaller pixel cell, compared to its predecessor (Llopart et al. 2002). A schematic diagram of the Medipix2 pixel cell is shown in Figure 13.7.

The first stage of the pixel cell is a charge preamplifier, which integrates the signal generated in the bump-bonded sensor by interacting X-rays. The resulting voltage is then compared with two thresholds, whose value is set at the circuit periphery by two 10-bit DACs. The two thresholds can be further adjusted pixel-by-pixel by using two in-pixel 3-bit adjustment DACs, to minimize pixel-to-pixel variations due to transistor mismatches.

Table 13.2: Detector Specification Comparison for a Number of PCDs

<table>
<thead>
<tr>
<th>ASIC</th>
<th>Reference</th>
<th>Pixel Pitch (µm)</th>
<th>Matrix</th>
<th>No. of Thresholds</th>
<th>Tile-Up Capability</th>
<th>Noise (e− rms)</th>
<th>Count Rate (Mcps/Pixel)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Medipix1</td>
<td>Campbell et al. (1998)</td>
<td>170</td>
<td>$64 \times 64$</td>
<td>1</td>
<td>NA</td>
<td>170</td>
<td>2</td>
</tr>
<tr>
<td>Medipix2</td>
<td>Llopart et al. (2002)</td>
<td>55</td>
<td>$256 \times 256$</td>
<td>2</td>
<td>3</td>
<td>140</td>
<td>1</td>
</tr>
<tr>
<td>Medipix3</td>
<td>Ballabriga et al. (2007)</td>
<td>55 (110)</td>
<td>$256 \times 256$</td>
<td>2</td>
<td>3</td>
<td>80</td>
<td>2.5</td>
</tr>
<tr>
<td>Medipix3</td>
<td>Ballabriga et al. (2007)</td>
<td>55 (110)</td>
<td>$256 \times 256$</td>
<td>2</td>
<td>3</td>
<td>174</td>
<td>0.5</td>
</tr>
<tr>
<td>Timepix</td>
<td>Llopart et al. (2007)</td>
<td>55</td>
<td>$256 \times 256$</td>
<td>ToT</td>
<td>3</td>
<td>75</td>
<td>1</td>
</tr>
<tr>
<td>Timepix3</td>
<td>Poikela et al. (2014)</td>
<td>55</td>
<td>$256 \times 256$</td>
<td>ToT</td>
<td>3</td>
<td>62</td>
<td>$10^{-3}$</td>
</tr>
<tr>
<td>PIXIE II</td>
<td>Bellazzini et al. (2013)</td>
<td>60</td>
<td>$512 \times 476$</td>
<td>2</td>
<td>2</td>
<td>50</td>
<td>0.5</td>
</tr>
<tr>
<td>PIXIE III</td>
<td>Bellazzini et al. (2015)</td>
<td>62</td>
<td>$512 \times 402$</td>
<td>2</td>
<td>2</td>
<td>50 (SPM)</td>
<td>100 (CSM)</td>
</tr>
<tr>
<td>Pilatus I</td>
<td>Brönnimann et al. (2003)</td>
<td>217</td>
<td>$44 \times 78$</td>
<td>1</td>
<td>3</td>
<td>75</td>
<td>NA</td>
</tr>
<tr>
<td>Pilatus II</td>
<td>Kraft et al. (2009)</td>
<td>172</td>
<td>$60 \times 90$</td>
<td>1</td>
<td>3</td>
<td>123</td>
<td>4</td>
</tr>
<tr>
<td>Pilatus III</td>
<td>Loeliger et al. (2012)</td>
<td>172</td>
<td>$60 \times 90$</td>
<td>1</td>
<td>3</td>
<td>123</td>
<td>12</td>
</tr>
<tr>
<td>XPAD1</td>
<td>Pangua et al. (2010)</td>
<td>330</td>
<td>$25 \times 24$</td>
<td>1</td>
<td>3</td>
<td>350</td>
<td>1.5</td>
</tr>
<tr>
<td>XPAD2</td>
<td>Pangua et al. (2007)</td>
<td>330</td>
<td>$25 \times 24$</td>
<td>1</td>
<td>3</td>
<td>350</td>
<td>1</td>
</tr>
<tr>
<td>XPAD3</td>
<td>Pangua et al. (2008)</td>
<td>130</td>
<td>$80 \times 120$</td>
<td>1</td>
<td>3</td>
<td>174</td>
<td>2</td>
</tr>
<tr>
<td>Eiger</td>
<td>Radici et al. (2012)</td>
<td>75</td>
<td>$256 \times 256$</td>
<td>1</td>
<td>3</td>
<td>160</td>
<td>4.2</td>
</tr>
<tr>
<td>SiemensPC</td>
<td>Kappeler et al. (2010)</td>
<td>64</td>
<td>225</td>
<td>2</td>
<td>NA</td>
<td>NA</td>
<td>40</td>
</tr>
<tr>
<td>SamsungPC</td>
<td>Kim et al. (2012)</td>
<td>60</td>
<td>$128 \times 128$</td>
<td>3</td>
<td>0</td>
<td>50</td>
<td>NA</td>
</tr>
<tr>
<td>AGH Fermilab</td>
<td>Maj et al. (2015)</td>
<td>100</td>
<td>$18 \times 24$</td>
<td>2</td>
<td>0</td>
<td>84 (SPM)</td>
<td>NA</td>
</tr>
<tr>
<td>Philips Chromaix</td>
<td>Steadman (2010)</td>
<td>300</td>
<td>$16 \times 16$</td>
<td>4</td>
<td>2</td>
<td>400</td>
<td>38</td>
</tr>
<tr>
<td>Hexitec</td>
<td>Jones et al. (2009)</td>
<td>80</td>
<td>$250 \times 250$</td>
<td>NA</td>
<td>3</td>
<td>NA</td>
<td>$10^{-3}$</td>
</tr>
</tbody>
</table>

Note: SPM, single pixel mode; CSM, charge summing mode; ToT, working in time over threshold.

a Possibility of binning 2× binning for increased spectroscopic performance.

b Edge-on silicon strip sensors.

c Digitalization of pulse amplitude is conducted off-chip.
arising from manufacturing processes. If the collected signal falls between those two thresholds, a pulse is sent to a double discrimination logic. The chip is driven by an enabling signal, referred to as a "shutter" in Figure 13.7. If the shutter signal is active, the pulse produced by the double discrimination logic increments the clock of a 13-bit pseudo-random counter by one. If the shutter signal is low, the pseudo-random counter works as a shift register and all the pixels in one column are connected for readout. The front end electronics were designed to collect either electrons or holes, so that sensors collecting holes, such as CdTe or CdZnTe (Chmeissani and Mikulec 2001; Pellegrini et al. 2006; Pennicard et al. 2014), could be read out by this chip in addition to electron collecting sensors (e.g., Si, GaAs) (Schwarz et al. 1999; Llopart et al. 2002; Mikulec et al. 2003; Campbell 2011). The minimum detection threshold was reported to be as low as \(1000 \text{ e}^{-}\) for both hole and electron collection (Llopart et al. 2002). With a maximum readout frequency of 200 MHz, this chip produced a power consumption in the order of 500 mW, with a supply voltage of 2.2 V. (Llopart et al. 2002).

Several readout interfaces were developed, based on a number of technologies including: commercial PCI cards (San Segundo Bello et al. 2003; Mettivier et al. 2006), USB interface (Holy et al. 2006; Vykydal et al. 2006), parallel readout (Maiorino et al. 2006; Ponchut et al. 2007), Gigabit optical links (Fanti et al. 2007), and Low Voltage Differential Signaling (LVDS) drivers (Vykydal et al. 2008).

The Medipix2 chip, as a fine pixelated detector, is affected by charge sharing, that is charge spread across adjacent pixels (see Section 13.3.3). Several studies have been conducted to model the effect of charge sharing in the Medipix2 chip when bonded to Si sensors (Quarati et al. 2005; Norlin et al. 2006) or CdTe sensors (Pellegrini et al. 2006, 2007). Charge sharing is expected to deteriorate imaging performance in terms of SNR, MTF (Modulation Transfer Function), and DQE (Bisogni et al. 2003; Bartl et al. 2008). Nevertheless, it is possible to limit charge sharing by increasing the low threshold value of the Medipix2 chip, resulting in improved imaging performance (Michel et al. 2006; Durst et al. 2007). Although raising the discrimination threshold allows improvement in imaging performance, this would result in decreased detection efficiency, thus increased dose to the patient (Chmeissani and Mikulec 2001). Limitations in imaging performance arising from charge sharing, as well as the availability of algorithms for suppression of charge sharing (Nilsson et al. 2006; Ponchut 2008), triggered the development of the Medipix3 chip (see Section 13.4.4).

Although the Medipix2 detector was developed as a counting detector, the combined use of low and high discrimination threshold allows for energy discriminating imaging with a narrow discrimination window (1.4 keV) (Tlustos et al. 2006). Energy windowing has the potential to improve SNR (Rosso et al. 2007) and allows for energy weighting (Karg et al. 2005; Butler et al. 2008) and spectral imaging (Butler et al. 2008). Due to the limited imaging area of the Medipix2 detector (1.4 cm \( \times \) 1.4 cm), X-ray imaging applications of this detector are restricted to small animal imaging (Belcari et al. 2007). Alternatively, they require mosaic-tiling of either sensors (Blanchot et al. 2006) or of images, acquired with detectors stepping across the sample (Bellazzini et al. 2013). In the clinical field, a prototype large area mammographic system (25 cm \( \times \) 19 cm) was developed by tiling a number of Medipix2 sensors. This imaging system showed a significantly higher MTF when compared to conventional mammography detectors (Blanchot et al. 2006).

### 13.4.3 Timepix

Within the Medipix2 collaboration, a new sensor (Timepix) was designed in 2005 and produced in 2008 (Llopart et al. 2007; Campbell 2011) to offer time of arrival measurements when combined with gaseous detectors (van der Graaf 2007). However, it appeared evident in the design phase that the time measurement capabilities could be used for direct spectroscopic measurements when the chip was bonded to a semiconductor sensor. The Timepix chip features the same number of pixels and pixel pitch as Medipix2. A schematic of the Timepix pixel...
Photon-Counting Detectors for X-ray Imaging

readout is shown in Figure 13.8a. Each pixel is provided with an analog section with a charge sensitive preamplifier and a single threshold discriminator, and a digital section with a time-based synchronization logic and a 14-bit counter with overflow control logic. An external reference clock, diffused on the whole matrix in less than 50 ns, is used to generate the clock signal that increments the in-pixel counter. Each pixel can work in one of three different modes. Counting mode: the number of incoming X-rays is counted, similarly to Medipix2; Arrival time mode: the counter works as timer and measures the X-ray arrival time; Time-over-Threshold (ToT) mode: the counter works as Wilkinson-type analog-to-digital converter (ADC), allowing direct energy measurements in each pixel. Figure 13.8b shows the ToT mode. If the preamplifier generates a signal above the discrimination threshold (triangles), the discriminator produces a logic signal during the time the signal is above the threshold. The counter records a hit for each clock during the time when the discriminator output is high (black squares). Since the preamplifier signal height is related to the amount of collected charge and its width is related to the height by the characteristic time of the signal shaper, a longer signal at the output of the preamplifier corresponds to a large amount of collected charge. Practically, in ToT mode, Timepix is able to measure the amount of charge deposited in the sensor by measuring the length of the preamplifier pulse. Timepix has the same dimensions, readout architecture, and floor plan as Medipix2, allowing almost full backward compatibility with all the existing Medipix2 readout systems. The total power consumption is comparable with Medipix2 (∼425 mW), with a supply voltage of 2.2 V. The noise is ∼100 e− rms, and the minimum detectable signal is ∼650 e− (Tlustos 2005). For PCDs based on discrimination thresholds (e.g., Medipix2), the minimum dead time achievable is estimated to be twice the peaking time (see Section 13.3.2). For ToT PCDs, dead time can be significantly higher, as preamplifier output signals are required to return to the baseline for ToT measurements. ToT PCDs are intrinsically limited in terms of their capability to deal with high flux applications.

Several calibration procedures have been proposed for energy calibration of ToT measurements (Jakubek 2009, 2011; Ponchut et al. 2013; Schioppa et al. 2014). The relation between ToT and energy is non-linear and largely affected by pixel-by-pixel variations and mismatches, requiring a per-pixel calibration, and charge sharing correction. Regardless of the practical complications of this procedure, excellent spectroscopic capabilities have
been reported for Timepix with an energy resolution (rms) of 5% at 60 keV (Jakubek et al. 2009) and 0.6% at 5.5 MeV (Jakubek et al. 2008a,b) when bonded to Si sensors, and 10.6% at 27 keV and 3.9% at 77 keV (Maneuski et al. 2012) when bonded to a CdTe sensor. Although the capability of spectroscopic imaging (Dudak et al. 2015) and material discrimination (Jakubek 2009) has been demonstrated for Timepix, the major drawbacks arising from the relatively high dead time have limited its applicability to X-ray imaging. A newer version, Timepix3 (Poikela et al. 2014), has been designed to provide more accurate timing information and is focused on particle tracking applications in High Energy Physics experiments.

13.4.4 Medipix3

Fine pixelated detectors are affected and strongly limited in their imaging and spectroscopic performance by charge sharing. To address this, the Medipix3 collaboration was formed in 2005 to develop a fine pitch hybrid pixel detector with inter-pixel communication to mitigate the effect of charge sharing at the pixel level.

The Medipix3 chip was designed as a fine pixelated matrix (256 × 256 pixels with a 55 μm pitch) (Ballabriga et al. 2007, 2011, 2013) using a commercial 0.13 μm CMOS process. It can be configured to work in fine pitch mode, where each 55 μm pixel is read out by the corresponding pixel element of the ASIC, or in spectroscopic mode, where four pixels are binned. In spectroscopic mode, four pixels in the semiconductor sensor array are connected to only one ASIC pixel, effectively increasing the pixel pitch to 110 μm. With this modality, up to eight thresholds and eight counters are available for each 110 μm pixel, leading to improved capabilities in energy selection and reduction of dead time, respectively.

The Medipix3 chip can either work in single pixel mode, where each pixel works in a simple photon counting regime independently from its neighbors, or in charge summing mode, where charge summing circuits and arbitration logic are used to correct charge sharing effects. Furthermore, the chip can work either in high gain mode, corresponding to a reduction in noise and linearity, but at the expense of the maximum count rate achievable due to larger pulses, or low gain mode, which entails a higher noise level but improved linearity and a shorter dead time. These two gain modalities benefit from detection of low and high energy signals, respectively.

A schematic diagram of the charge summing algorithm is shown in Figure 13.9a. When a charge cloud is generated in the sensor, due to X-ray interaction, the readout ASIC compares each pixel output with a low discrimination threshold. If one or more pixels exceed this threshold, the pixel that detected the highest signal is kept as the spatial location of the event, while the other pixels are suppressed to preserve spatial resolution. To assign a signal height to the event, the signals from each group of four neighboring pixels are summed at the so-called “sensing nodes.” The summed signal is compared with a summation threshold. If at least one of the four nodes surrounding the pixel assigned to the event exceeds the summation threshold, the counter is incremented by one. It is of note that, for this charge summation algorithm to work effectively, hit pixels must be adjacent. This reflects the condition where charge sharing is produced by the dimensions of the charge cloud or low energy fluorescence X-rays, but would be partially ineffective in the case of charge sharing due to high energy fluorescence X-rays that can be detected in a non-adjacent pixel compared to the primary event. This is of particular concern for high-Z sensors, where the K-fluorescence yield is relatively high and the mean free path of the produced X-rays larger than the pixel pitch (Pennicard et al. 2011).

A schematic of the Medipix3 pixel cell is in Figure 13.9b. Each pixel is provided a charge sensitive preamplifier (based on the Krummenacher architecture (Krummenacher 1991) that can handle both positive and negative input charges), and a shaping amplifier that converts the input voltage into a number of current copies, which are then sent to the summing nodes, common to neighbor pixels, and effectively placed at the pixel corner. Two discriminators, provided with a 5-bit adjustment DAC to minimize threshold dispersion across the detector matrix, and two counters are placed in each pixel. Counters can be configured to work as 1-bit, 4-bit, or 12-bit counters. Additionally, it is possible to benefit from a 24-bit counter, and thus an extended dynamic range, if using only a single threshold. An arbitration circuit is provided for implementing the charge summing modality. The arbitration circuit of the pixel detecting the highest charge sum compared to its neighbors performs the charge arbitration decision. When the signal coming from a given discriminator wins the arbitration decision, a pulse increments the corresponding pixel counter. Data acquisition can be performed either sequentially or continuously. When working in sequential mode, both discriminators and counters are available for each readout cycle. Conversely, in sequential mode only one threshold and counter can be used in any readout cycle, while the discriminator and counter are read out. With about 1600 transistors per pixel, the power consumption of the Medipix3 chip is 900 mW in charge summing mode and 600 mW in single pixel mode, the latter being comparable to Medipix2. A noise level of ~70 e− rms is reported for this sensor when working in single pixel mode (Ballabriga et al. 2013).

The effectiveness of the charge summation algorithm of Medipix3, based on inter-pixel communication, is shown in Figure 13.10. X-ray spectra (120 kVp, W target) were measured with the Medipix3 chip bump-bonded to a 2 mm thick CdTe sensor for single pixel mode (SPM) and charge summing mode (CSM). Measured spectra are also compared with simulated spectra, one of which includes a low-pass filter whose parameters were empirically set to reproduce the measured spectra (Koenig et al. 2014).

The spectrum obtained in SPM over-estimate the simulated spectrum (Monte Carlo simulation, MC) and the spectrum obtained in CSM at low energy and underestimate these two at high energies. This effect arises from charge sharing. The effect of charge sharing is strongly mitigated in CSM. Additionally, the Cd- Kα fluorescence peak at 23 keV is visible in both measured spectra. For the CSM spectrum the peak appears more clearly while, in SPM, it is visible as a shoulder at low energies, due to charge sharing. Tungsten fluorescence peaks (W-Kα and W-Kβ), clearly highlighted in the simulated spectrum (MC), are visible, although not resolved, in CSM mode, and are absent in SPM.

Capabilities of spectral imaging have been demonstrated for the Medipix3 detector (Walsh et al. 2011, 2013; Hamann et al. 2015), as well as material identification of up to three tissue types (Ronaldson et al. 2012).
13.4.5 Large Area Medipix-Based Imaging Systems

The limited imaging area for all the chips of the Medipix family (1.4 cm × 1.4 cm) represents an important limitation for these sensors to be used in clinical applications that require significantly larger areas (e.g., 40 cm × 40 cm for chest radiography, 25 cm × 30 cm for mammography). The imaging area of PCDs is limited by the ASIC dimensions, which in turn is limited by the reticle used for the lithographic CMOS processing. A possible approach to produce larger ASICs would be using reticle stitching, based on the use of a mask reticle that is stepped and repeated, in whole or in part, across a silicon wafer, to create modularly different sectors of a large circuit (Turchetta et al. 2011). However, reticle stitching limits the CMOS processes available, and often entails a low yield, inversely proportional to the chip area, due to density of defect points and impurities. Currently the Medipix and Timepix ASICs are designed to be 3-side buttable (i.e., chips can be abutted on only three sides) and bump-bonded to a monolithic semiconductor sensor of the requested size. The fourth side, that cannot be used for mosaic-tiling, is used for periphery circuits and wire bonding for I/O connections.

FIGURE 13.9 (a) Schematic diagram of the charge summing scheme for the Medipix3 chip. A hit is detected across the 4-pixel boundaries and the signal recorded for each pixel is shown. The pixel containing the highest signal is chosen as the location of the event. Copies of the signal are delivered to the sensing nodes. The summation signal is compared with a summing threshold, and the highest signal in the summing nodes is associated to the event. (Reproduced with permission from Talla, P.T. et al. 2011. Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment 633:S128–30.) (b) Schematic representation of the Medipix3 pixel cell. (Reproduced with permission from Ballabriga, R. et al. 2011. Nuclear Instruments and Methods in Physics Research Section A: Accelerators, Spectrometers, Detectors and Associated Equipment 633:S15–S18.)
limiting the tiling possibilities of the device to one double row of chips (a $2 \times N$ configuration) (Fornaini et al. 2003). At the interface between two adjacent abutted chips, sensor pixels are stretched so the detection efficiency is preserved at the expense of the spatial resolution (Campbell et al. 2016). Several examples of large area detectors resulting from assemblies of multiple Medipix chips have been reported: MAXIPIX, featuring 5 × 1 or 4 × 4 arrays of Medipix2 chips (Ponchut et al. 2011); Widepix, using a 10 × 10 array of Timepix chips (Jakubek et al. 2014); Lambda, with a 2 × 6 array of Medipix3, notably bump-bonded to high-Z semiconductor including Ge (Pennicard et al. 2013); and Excalibur, consisting of three modules of $4 \times 4$ Medipix3 chips (Marchal et al. 2013). To overcome the limitations in imaging area arising from having a dedicate side of the chip for I/O pads, significant effort has been put into the development of the Through Silicon Via (TSV) process to produce edge-less chip modules that can be 4-side tillable with minimum dead space (Vykydal et al. 2008; Henry et al. 2013).

13.4.6 Pilatus

In 1998, the Swiss Light Source (SLS) synchrotron and the Paul Scherer Institute (PSI) joined efforts to develop a large area photon counting chip for X-ray experiments at SLS. Pilatus I was prototyped in 2002, featuring a 80 mm × 36 mm monolithic Si sensor (300 µm thick) bump-bonded to a mosaic-tiling of $2 \times 8$ Pilatus I ASICs (Brönnimann et al. 2003), and a subsequent larger version offering a 21 cm × 24 cm imaging area (Broennimann et al. 2006). However, to overcome a number of shortcomings related to this design, a second version was developed. The Pilatus II chip (Kraft et al. 2009) was a 60 × 97 pixel matrix with a 172 µm pitch, using a commercial 0.25 µm CMOS process. The individual chips can be tiled in a $2 \times 8$ configuration to form a so-called Pilatus II module. These modules can be further tiled to achieve larger areas. The largest Pilatus II assembly features 6 million pixels, covering a 43 cm × 45 cm imaging area (Kraft et al. 2009). The chip is bump-bonded to a 320-µm thick Si sensor. Each pixel cell of the Pilatus II chip comprises a preamplifier, with high and low gain setting, a shaper, and a single threshold discriminator, followed by a 20-bit counter. The single threshold of the discriminator can be adjusted by a 6-bit in-pixel ADC, reducing the threshold dispersion from 343 eV to ∼50 eV over 10² pixels. The high count rate requirements for synchrotron applications made it necessary to develop a count rate correction algorithm to reduce the dead time losses at high fluxes (Trueb et al. 2012).

To provide a more robust solution to the dead time losses, a design solution to avoid paralyzation due to pile-up was included in Pilatus III. The Pilatus III readout ASIC is based on Pilatus II, with the addition of an instant retriggering scheme and a better handling of the counter overflow, in order to increase the count rate (Loeliger et al. 2012). The instant retriggering scheme is based on the idea that it is possible to avoid paralyzation by partitioning the actual signal pulse into a sequence of nominal single photon pulses. In this scheme, the pulse signal is re-evaluated several times during a read out cycle, and, if a pile-up event is detected, counters are retriggered and able to detect successive photons arriving within the same readout cycle. The time period between consecutive re-evaluations is effectively the dead time of this architecture, and has to be chosen to match the width of a single photon pulse. The basic principles of the instant retriggering technology are shown in Figure 13.11a. The first diagram shows the signal pulses generated by incoming photons, including a single photon event, a pile-up of two events, and a pile-up of multiple events, which exceed a detection threshold. The output of the discriminator and counter for a conventional PCD are shown in the second and third diagrams. The effect of the
paralyzation is clear: the dead time extension, following the detected photon, paralyzes the system, and results in the following photons not being detected. The output of the discriminator and counter for a PCD with instant retriggering technology is reported in the fourth and fifth diagram.

A predefined dead time interval is started after the first photon has been detected and repeated for as long as the signal pulse is above threshold. The counter is retriggered for each new dead time interval, and multiple pulses within the same readout cycle can be separated. The improvement in terms of sustainable count rate is shown in Figure 13.11b. A Pilatus III, tested with and without the retriggering capability, and a Pilatus II detector were exposed to 10-keV X-rays at the Australian synchrotron with a flux up to $10^8$ photons pixel$^{-1}$ second$^{-1}$. The three systems show comparable behavior at very low fluxes, but, as flux increases, the differences become appreciable. Both Pilatus II and Pilatus III without the retriggering readout scheme behave as paralyzable systems, with an observed versus true count rate curve featuring a maximum (see for comparison Figure 13.6b). The use of the retriggering scheme makes the system non-paralyzable (see Figure 13.6b) and greatly improves counting efficiency, with a maximum observed count rate of $12 \times 10^6$ and $4 \times 10^6$, with and without retriggering, respectively.

The Pilatus detector has been mainly designed for synchrotron applications; however, proof of principle for the use of these detectors in X-ray imaging has been reported (Bech et al. 2008).
13.4.7 XPAD

Development of the XPAD (X-ray Pixel Chip with Adaptable Dynamics) started in 1990 at the Center de Physique des Particules de Marseille (CPPM, France) by the electronic design team who worked on the hybrid pixel detectors for the DELPHI experiment at LEP (CERN) (Pangaud et al. 2010). XPAD1 was released in 1998 and featured $25 \times 24$ pixels with a $330 \mu m$ pitch. Each pixel cell consisted of a charge sensitive preamplifier, a differential discriminator with a 4-bit DAC for fine tuning of the threshold level for each individual pixel, a 16-bit overflow digital counter, and a readout system (Blanquart et al. 2000). The XPAD1 was affected by a large dispersion at the input of the discriminators ($\sim3500 e^-$), that could be reduced to ($\sim1100 e^-$) using the 4-bit threshold tuning. A second version, XPAD2, was manufactured in 2002 to mitigate the large dispersion of the pixel offset, by increasing the number of in-pixel DAC bits (from 4 to 6) and other issues occurring at the counter level (Delpierre et al. 2002). XPAD2 was designed to be mosaic-tiled and up to eight sensors, bump-bonded to a $500 \mu m$ thick Si sensor, could be read out together, covering an imaging area of $68 mm \times 68 mm$ (Delpierre et al. 2007a). A CT scanner based on the XPAD2 chip (PIXISCAN) was built at CPPM (Delpierre et al. 2007b) for small animal imaging, showing the capability to detect in-vivo millimetric cancerous lesions at low dose (Debarbieux et al. 2010), although limited by the large pixel size ($330 \mu m$ pitch).

The XPAD3 chip represented a major redesign, abandoning the obsolete AMS-0.8 $\mu m$ CMOS process in favor of the more modern IBM-0.25 $\mu m$. The pixel pitch was smaller than the one used in its predecessors (130 $\mu m$ over a pixel matrix of 120 $\times$ 80), to provide higher spatial resolution (Pangaud et al. 2008). Two versions were designed: one that could accept positive carriers (XPAD-S), from Si or GaAs for example, and another for electron collection (XPAD3-C). The energy range at which those two chips work is also different: from 5 up to 25 keV for XPAD3-S, and across a 60 keV window for XPAD3-C (Berar et al. 2009), with the lower limit being set by the electronics noise ($<140 e^-$) and the higher limit set by the absorption efficiency of the sensor. For example, for a $500 \mu m$ thick Si sensor, the thickness chosen for XPAD3-S, the detection efficiency at 25 keV, is only 25%. The input charge is amplified by a sensitive preamplifier and, subsequently, integrated in an operational transconductance amplifier (OTA) that converts the voltage into a current, which is then passed to the current mode discriminator.

The discriminator, at the selection stage, receives the output of the OTA, together with currents for the global threshold and in-pixel adjustment values. Incremental counts are then sent to the 12-bit counter with overflow management, and then transferred to an in-pixel local memory for on-flight readout. The local memory can be accessed while the main pixel counter is in operation. The frame rate achievable with this sensor is 500 frames per second, while the maximum flux it can handle has been reported to be as high as $2 \times 10^6$ photons/ pixel/s (Pangaud et al. 2008). Although XPAD3 has been mainly designed for synchrotron applications, the capability for use in K-edge CT has been reported (Cassol Brunner et al. 2013).

13.5 Conclusions

Over the last two decades, significant advances in microelectronics, sensors’ manufacturing, and interconnection techniques for small pixels have made the realization of direct-conversion photon-counting detectors for X-ray imaging possible. Detectors with fine pitch pixels, high detection efficiency, and on-pixel intelligence can now be realized. The advantages of such detectors in the field on X-ray imaging have been discussed in this chapter. Noise rejection, low leakage current, and energy resolving capabilities make PCDs a potential improvement over conventional X-ray imaging modalities, by delivering higher image quality with lower dose. Furthermore, PCDs pave the way for the introduction of new imaging modalities, such as K-edge imaging for tissue identification and functional CT. The success of several prototype PCDs has attracted commercial interest, leading, for example, to the commercialization of Medipix2 through technological transfers to several companies including PANalytical and X-ray Imatek, and the creation of spin-off companies, including Detciris for the PSI detectors, and Pixirad for the PCDs developed at the University of Pisa and INFN. Major manufacturers of X-ray imaging equipment are currently developing CT systems based on PCDs, including Siemens (Kappler et al. 2012) and Philips (Steadman 2010).

Nonetheless, there are still several areas of improvement for PCDs needed to translate into clinical practice. Semiconductor sensors for clinical use require a large area, good uniformity, and low concentration of defects to mitigate charge trapping, increase charge carrier mobility, and, consequently, reduce peaking time in order to sustain high fluxes.

The capability of sustaining high count rates is an essential feature for clinically usable PCDs, as any counting inefficiency will lead to an increased dose to the patient. Several strategies to deal with high fluxes have been reviewed, but there is still a need to develop faster PCDs. For these detectors to be successfully translated into clinical practice, they need to provide large imaging areas required by medical applications. Current PCDs are limited by the dimension of their ASICs and by having only three sides available for mosaic-tiling. There is increasing attention towards the development of TSV technology for fine pixelated detectors, with the aim of manufacturing ASICs that can be tiled seamlessly on all four sides to build large area detectors with full area coverage.

To have large area PCDs, efforts should be made to reduce power consumption, especially for high-Z sensors, to avoid deterioration of detector performance arising from variations in leakage current, saturation of the front end electronics, and increased pixel-to-pixel variations. Further developments in CMOS processing and reduction of the feature size could lead either to smaller pixels or to the integration of more in-pixel complex circuitry.

Medipix3 has set an example of the achievements that can derive from on-pixel intelligence. Further development of on-pixel intelligence with functionalities for on-the-fly image pre-processing at pixel level, online dose monitoring and adaptive modalities could be greatly beneficial to a number of X-ray imaging applications.

---

‡ Detciris, Baden-Daettwil, Switzerland. https://www.dectris.com
§ Pixirad, Pisa, Italy. http://www.pixirad.com
REFERENCES


Bellazzini, R. et al. 2013. Chromatic X-ray imaging with a fine pitch CdTe sensor coupled to a large area photon counting pixel ASIC. *Journal of Instrumentation* 8:C02028.


Photon-Counting Detectors for X-ray Imaging


