A Segmented Silicon Strip Detector for Photon-Counting Spectral Computed Tomography

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Abstract

Spectral computed tomography with energy-resolving detectors has a potential to improve the detectability of images and correspondingly reduce the radiation dose to patients by extracting and properly using the energy information in the broad x-ray spectrum. A silicon photon-counting detector has been developed for spectral CT and it has successfully solved the problem of high photon flux in clinical CT applications by adopting the segmented detector structure and operating the detector in edge-on geometry. The detector was evaluated by both the simulation and measurements.

The effects of energy loss and charge sharing on the energy response of this segmented silicon strip detector with different pixel sizes were investigated by Monte Carlo simulation and a comparison to pixelated CdTe detectors is presented. The validity of spherical approximations of initial charge cloud shape in silicon detectors was evaluated and a more accurate statistical model has been proposed.

A photon-counting energy-resolving application specific integrated circuit (ASIC) developed for spectral CT was characterized extensively by electrical pulses, pulsed laser and real x-ray photons from both the synchrotron and an x-ray tube. It has been demonstrated that the ASIC performs as designed. A noise level of 1.09 keV RMS has been measured and a threshold dispersion of 0.89 keV RMS has been determined. The count rate performance of the ASIC in terms of count loss and energy resolution was evaluated by real x-rays and promising results have been obtained.

The segmented silicon strip detector was evaluated using synchrotron radiation. An energy resolution of 16.1% has been determined with 22 keV photons in the lowest flux limit, which deteriorates to 21.5% at an input count rate of 100 Mcps mm\(^{-2}\). The fraction of charge shared events has been estimated and found to be 11.1% for 22 keV and 15.3% for 30 keV. A lower fraction of charge shared events and an improved energy resolution can be expected by applying a higher bias voltage to the detector.

Key words: photon counting, spectral computed tomography, silicon strip detector, ASIC, energy resolution, cadmium telluride, charge sharing, Monte Carlo simulation, synchrotron
Publications

This thesis is based on the following papers:


Reprints were made with permission from the publishers.
The author has contributed to the following publications, which are to some extent related to the thesis but not included.


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Chapter 1

Introduction

1.1 Computed Tomography

Computed tomography (CT), a widely used diagnostic imaging modality, produces digital format images which represent the discrete slices of an imaged object by measuring the attenuated x-ray intensity and reconstructing the spatial distribution of linear attenuation coefficients from a sufficient number of x-ray projections taken at different angles [1,2]. The invention of CT successfully overcomes the fundamental limitation of conventional projection radiography, i.e. the superposition of different anatomical structures, thus offering a much better visualization.

The first clinical CT image was acquired in 1972 [3]. Over the past four decades, CT has experienced an enormous improvement [2,4]. The CT scanner has evolved from the pencil-beam geometry in the 1970s to today’s multi-slice spiral CT. The corresponding single slice image acquisition time has been reduced from 5 minutes to less than 1 second. CT has become one of the most important diagnostic imaging modalities. In addition to a wide range of routine applications, various new applications have been exploited [1,2]. The integration of CT with other functional imaging modalities, e.g. positron emission tomography (PET) and single photon emission computed tomography (SPECT), offers more possibilities [5,6].

The use of CT has increased markedly since its introduction, around 70 million CT scans were performed in the United States alone in 2007 [7], which also means a noticeable increase in radiation dose to patients [8,9], in particular for children who are more sensitive to radiation than adults [10–13]. Although the estimated risks to individuals are small, the small risks combined with a large exposed population could result in a potential public health risk and a remarkable number of cancers; around 1.5 to 2.0% of all cancers in the United States might be caused by the radiation from CT scans [7,8]. Simply replacing CT with other imaging modalities, such as magnetic resonance imaging (MRI), is not realistic for many common imaging scenarios [14–16]. Several methods and strategies have been proposed to reduce or optimize the radiation dose from CT scanning [17–19], for example by
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automatic exposure control [20]. The radiation dose can also be reduced by correspondingly improving the image quality with the development of new technology. The research interest of this thesis, photon-counting spectral CT, is one of these new technologies [21].

1.2 Spectral Computed Tomography

In clinical CT systems, the x-ray energy spectrum after passing through an object is polychromatic and x-ray photons of different energy have different contrast information as a result of the energy- and material-dependent linear attenuation coefficients. The state-of-the-art CT with energy-integrating detectors simply integrates the broad energy spectrum reaching the detector regardless of the contrast information contained in different photon energies. The potential benefits of using x-ray spectral information in medical imaging was realized early [22,23], and the corresponding applications in clinical CT has received increasing attenuation in recent years [24–27]. There are two main methods to fully exploit the spectral information of the transmitted x-ray energy spectrum through an examined object: energy weighting and material decomposition, both of which can improve the detectability of CT images.

Energy weighting requires a photon-counting energy-resolving detector which is capable of detecting the individual photons and measuring the correct photon energies accordingly. An improved detectability can then be obtained by optimally weighting the detected photons [23,28,29]. Different weighting methods have been proposed (projection-, image- and task-based) and shown a significant performance improvement compared to the energy-integrating method [30–32].

The goal of material decomposition is to differentiate and characterize different materials and tissues in an examined object by decomposing the energy-dependent linear attenuation coefficients into a linear combination of energy-dependent basis functions and the corresponding basis set coefficients [22,24,26,27,33]. The beam hardening artifacts can also be eliminated as a result of having the entire energy dependence of the linear attenuation coefficients. Two basis functions are enough to approximate the linear attenuation coefficients of low atomic number materials with high accuracy [22,33]. K-edge imaging is feasible if more basis functions are included to quantify the contrast agents (typically iodine and gadolinium with relatively high k-edge energies) [26,27].

Several approaches have been developed to extract the x-ray spectral information in CT applications. Two dual-energy CT systems are commercially available and have been used in clinical applications: the dual-source CT provided by Siemens [34–38] and the fast-kVp switching CT supplied by General Electric [39–44]. Both systems exhibit promising potential in material decomposition, e.g. the differentiation of the distribution of contrast agent and characterization of renal stones and calcified plaques in vessels [34,35,40]. However, the energy-integrating detectors are still used in these dual-energy CT systems and two energy spectra are
not entirely separated. The overlap in the energy spectra (i.e., an extra radiation dose), limited number of energy levels, scatter radiation between two detector systems for dual-source CT [36], the mis-registration problem for fast-kVp switching CT [45] are the corresponding drawbacks, which affect the effectiveness of spectral imaging and result in lower signal-difference-to-noise ratios (SDNR). The dual-layer detector technique also suffers from the disadvantage of overlapped spectra [46–48]. Another candidate technique for spectral CT, which attracts more attention recently, is to use photon-counting energy-resolving detectors with multi-energy bins and excellent energy-discrimination capability [Paper III, IV, V]. See next section for a detailed description of photon-counting energy-resolving detectors.

1.3 Detector Technology

The detector is one of the most important components of a CT system. Two main detector types are commonly used by almost all modern CT scanners: xenon filled ionization chambers and scintillation detectors [1,2]. The xenon ionization chambers are only installed in some low-end single-slice CT scanners due to the low detection efficiency and the difficulty to manufacture multiple detector rows. Most of the CT scanners are equipped with scintillation detectors which consist of scintillators coupled to photodiodes. Gadolinium oxysulfide (GOS) ceramic, ultra fast ceramic (UFC) and Gemstone are three scintillation materials used by the latest high-end CT scanners [38,44,49,50]. In scintillation detectors, an interacting x-ray photon is first converted to scintillation lights in scintillators, then electron-hole pairs are generated through the absorption of scintillation lights in photodiodes. All the detectors mentioned above integrate the energy deposited by the interacting photons over a certain exposure time and output a signal proportional to the total deposited energy, thus being called energy-integrating detectors. The energy information involved in the incident x-ray energy spectrum is simply lost as a result of the energy-integrating process. Additionally, the electronic noise produced by detector sensors and readout electronics is also integrated into the output signal, which degrades the SDNR of CT images.

The advances in semiconductor sensors and application specific integrated circuits (ASICs) have made photon-counting spectral CT feasible. The x-ray photons interacting with the semiconductor sensors can be converted directly to electron-hole pairs without any inefficient intermediate processes, ensuring the superior intrinsic energy resolution [51,52]. The high charge mobility and thus the short charge collection time in semiconductor detectors is obviously another advantage for clinical CT systems which are usually running under extremely high photon fluxes, up to several million photons s$^{-1}$ mm$^{-2}$. In addition, an ultra-fast photon-counting ASIC with energy-discrimination capability is needed to process the photon-converted pulses individually and retain good energy resolution [Paper III, IV]. The electronic noise can be removed by setting a threshold above noise floor and the energy-discrimination capability can be accomplished by applying multiple thresholds.
Cadmium telluride (CdTe) and cadmium zinc telluride (CZT) are two candidate semiconductor materials for photon-counting energy-resolving detectors [53, 54]. Several evaluations including preclinical research have been performed to investigate the spectral imaging capabilities of CT scanners based on CdTe/CZT detectors [55, 56]. However, the photon flux in the evaluations is far below the requirements of a CT system in clinical practice. Currently, the main method to mitigate the high photon flux problem for CdTe/CZT detectors is to decrease the pixel size [57, 58], which also means an increased spectrum distortion as a result of charge sharing and fluorescence escape [Paper I]. In addition, a serious effect limiting the use of CdTe/CZT detectors under high photon flux irradiation is polarization, i.e. the build-up of space charge in detectors, as a result of low hole mobility and hole trapping, which collapses the electric field, severely reduces the amplitude of the detected pulses and leads to a rapid decline of count rate at a critical photon flux [59]. A few other technologies for CdTe/CZT detectors, such as parallel drift structure [60] and multiple-layers arrangement [61], are under investigation but a range of technical difficulties need to be solved [53].

The feasibility of using silicon as the sensor material for photon-counting spectral CT has been investigated previously [24, 62]. The detector is a photon-counting segmented silicon strip detector [Paper V]. The mature manufacturing technology and low cost are the advantages of silicon detectors. The high mobility for both electrons and holes means less pulse pileup. The low detection efficiency of silicon for high energy x-rays can be compensated for by operating the silicon strip detector in edge-on geometry. The detector strips are segmented along the photon incident direction with each segment connected to an individual ASIC channel. As a result the count rate in each detector segment and the subsequent ASIC channel is reduced. The only possible limitation of using silicon for CT is the high fraction of Compton scattering for high energy photons. Previous studies have shown that the effect of Compton scattering is manageable [24, 62]. The detector and the corresponding evaluations are described in Chapter 4.

1.4 Outline of the Thesis

The project to develop a novel segmented silicon strip detector for photon-counting spectral CT was started in 2008, and I have been involved in the project ever since. My research focuses mainly on the evaluation of the detector by simulation and measurements. After four years of work, five peer-reviewed articles have been written based on the main research results, which are the basis of this thesis.

CdTe/CZT is the main competitor to silicon as a candidate detector material for photon-counting spectral CT. A simulation study was performed to investigate the energy response of pixelated CdTe detectors as a function of pixel size and the results are presented in Paper I. For comparison, a simulation of the segmented silicon strip detector was also carried out and the results are presented in Chapter 2. The purpose of such comparison between silicon and CdTe/CZT is to answer
1.5. CONTRIBUTION BY OTHERS

a question in this field “Why use silicon? Why not CdTe or CZT” from the perspective of detector energy response by showing that the pixelated CdTe detector also suffers from severe spectrum distortion for small pixel sizes. As a by-product of the simulation work, Paper II evaluates the validity of spherical approximations of initial charge clouds produced by x-ray photons in silicon detectors and proposes a more accurate statistical model. The characterizations of the first and second generations of the ASIC are presented in Paper III and IV, respectively. Paper V describes the evaluation of the first-version segmented silicon strip detector developed for photon-counting spectral CT.

1.5 Contribution by Others

The author is the principal contributor to all the results presented in this thesis. However, it should be recognized that some of the measurements depended on the work carried out by others. The ASIC was designed by Christer Svensson and collaborators at Linköping University. The mechanical components used in the measurements were fabricated by Staffan Karlsson. The photon-counting CT image presented in Paper III was reconstructed by Mats Persson and the image acquisition program was provided by Moa Yveborg. In Paper IV, the synchrotron measurements of the ASIC at the Diamond Light Source were carried out by Mats Persson, Staffan Karlsson, Hans Bornefalk and the author. In Paper V, the measurements of the detector at the Elettra synchrotron were performed by Mats Persson, Staffan Karlsson, Han Chen and the author.
Chapter 2

Detector Simulation

2.1 Introduction

In a photon-counting energy-resolving detector, several physical effects lead to detected events with reduced energies, including Compton scattering, fluorescence emission, charge diffusion, trapping of charge carriers and slow-hole-motion-induced incomplete charge collection. Charge sharing is the result of the lost energy being collected by adjacent pixels, and the shared charge may give rise to double counts in adjacent pixels.

Improvement in image quality of a photon-counting x-ray imaging system can be accomplished by weighting the detected photons to an optimal weighting scheme [28, 29]. However, the magnitude of the improvement will be reduced as a result of the deteriorated energy resolution of the detector caused by the various physical effects mentioned above [63] and pulse pileup [64]. The distorted energy spectrum caused by energy loss and charge sharing produces cupping artifacts and inaccurate CT numbers when applying optimal energy weighting schemes, also reduces the benefits of contrast-to-noise ratio (CNR) improvement and shifts the estimated attenuation coefficients of the investigated object [63]. Consequently, there is a need to better understand and investigate the energy loss and charge sharing effects in energy-resolving detectors for photon-counting spectral CT.

The effects of energy loss and charge sharing on the response of pixelated CdTe detectors developed for photon-counting spectral CT were evaluated in Paper II by simulating the photon transport and the charge-collection process with a Monte Carlo-based simulation. An evaluation on the segmented silicon strip detector is presented in the thesis for comparison.

Spherical approximations have been used extensively in low-energy x-ray imaging to represent the initial charge cloud produced by photon interactions in silicon detectors [65–68]. However, for high-energy x-rays, where the initial charge distribution is as important as the charge diffusion process, the spherical approximations will not result in a realistic detector response. Paper I evaluates the validity of the
spherical approximations of initial charge cloud shape in silicon detectors, then
presents a statistical model (bubble-line model) for photons in the upper energy
range of medical imaging.

2.2 Simulation Methods

The details of the simulation procedure are presented in Paper II. The interactions
of the incident photons with the detector volume and the transport of all the sec-
secondary particles were traced by a Monte Carlo simulation code PENELLOPE [69],
and the deposited energies and corresponding coordinates along the tracks were
recorded. The energy tracks were then converted into electron-hole pairs by di-
viding the deposited energies along the tracks by the ionization energy, 4.4 eV for
CdTe and 3.6 eV for silicon.

The charge carriers get separated immediately after generation and drift in the
electric field. The currents induced on corresponding electrodes by the movement
of charge carriers were calculated by the Shockley-Ramo theorem [70,71]

\[ i(t) = -q \sum_{j=1}^{N} \exp(-t/\tau) E_{wj}(\vec{r}_j(t)) \vec{v}_j(t) \]  \hspace{1cm} (2.1)

where \( q \) is the elementary charge, \( N \) is the number of charge carriers, \( t \) is the drift
time, \( \vec{r}_j \) determines the trajectory of a charge carrier, \( \tau \) is the mean lifetime of the
charge carriers, \( E_{wj} \) is the weighting field that describes the electrostatic coupling
between a charge carrier and the investigated electrode, and \( \vec{v}_j \) is the velocity of
the carrier. The velocity of charge carriers is induced by both the drift under
the influence of electric field and the thermal diffusion of charge carriers, where
the latter effect is the main cause of charge sharing during the charge collection
process. The diffusion effect was analyzed by the diffusion equation

\[ P(dr) = \frac{1}{\sqrt{4\pi D dt}} \exp\left(-\frac{dr^2}{4D dt}\right) \]  \hspace{1cm} (2.2)

where \( dr \) is the diffusing distance of a charge carrier in one direction randomly
chosen from the Gaussian probability distribution \( P(dr) \) after a time step \( dt \), and
\( D \) is the diffusion coefficient given by \( D = \mu kT/q \), with \( \mu \) being the mobility of a
charge carrier, \( k \) the Boltzmann constant and \( T \) the absolute temperature. Due to
the high purity of silicon detectors, the exponential component which represents the
trapping effect can be removed. The charge collected on an electrode was obtained
by integrating the induced current over a certain measurement time.

The electric potential in the detector volume was calculated numerically with
the Poisson equation

\[ \nabla^2 \varphi = -\frac{qN}{\varepsilon} \]  \hspace{1cm} (2.3)
on a three-dimensional grid using the Successive Over-Relaxation method, where \( \varepsilon \) is the permittivity of the material, \( N \) is the acceptor concentration for p-type CdTe or the donor concentration for n-type silicon, and \( \phi \) is the electric potential. The weighting field was calculated from Laplace’s equation by setting the potential of the investigated electrode biased at unit voltage and by grounding the other electrodes.

2.3 Detector Geometries

Three different pixel sizes were characterized for both the pixelated CdTe and segmented silicon strip detectors, which are 1.0 \( \times \) 1.0 mm\(^2\), 0.5 \( \times \) 0.4 mm\(^2\) and 0.3 \( \times \) 0.2 mm\(^2\), respectively. The detector thicknesses of the silicon detectors are 1.0 mm, 0.5 mm and 0.3 mm, and the pixel pitches are 1.0 mm, 0.4 mm and 0.2 mm for three different pixel geometries. By operating the silicon detectors in edge-on geometry, a 1.4 mm long center segment was chosen as the investigated segment around which to model charge sharing to all eight neighboring detector segments. A bias voltage of 600 V was applied to the backside of the segmented silicon strip detector with a mid-sized pixel (i.e. 0.5 \( \times \) 0.4 mm\(^2\)) and the bias voltage was adjusted to keep the average intensity of electric field identical for different pixel geometries, i.e. the bias voltages to the detectors with large and small pixel sizes were 1200 and 360 V, respectively. The CdTe detectors were irradiated through the cathode to reduce the drift distance of holes. The distance between anode and cathode was 3.0 mm for all three pixel geometries and the bias voltage was –1000 V.

2.4 Results and Discussion

2.4.1 Pixelated CdTe Detector

The trapping of charge carriers and slow-hole-motion-induced incomplete charge collection were evaluated with an analytical model, assuming that interacting photons deposit all energy as charges at points along the central axis of a pixel. A geometrical consequence of this simplification is that it avoids charge sharing as a result of diffusion, the initial charge cloud extent and fluorescence. The ratio of lost energy to deposited energy as a function of the interaction depth in the detector was obtained and shown in Fig. 2.1(a). The corresponding values of the weighting potential along the central axis of the investigated pixels are shown in Fig. 2.1(b). The weighting potentials concentrate around the detector anodes and the value of weighting potential is lower in the smaller pixel. The energy loss increases when the photon interacts with the detector at a deeper level. The smaller pixel results in less energy loss at the same interaction depth as a result of the lower value of the weighting potential. This effect is called “small pixel effect” [72], which means that the small ratio of pixel size to thickness helps improve the charge-collection efficiency and reduce the effect of poor hole collection.
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Figure 2.1: (a) The ratio of lost energy to deposited energy in a pixelated CdTe detector for three pixel sizes as a function of interaction depth along the central axis of the investigated pixels. (b) The distributions of weighting potential along the central axis.

For the following evaluations, the incident photon energy was sampled from a 120 kVp tungsten anode x-ray spectrum [73] filtered by 2 mm aluminum and 0.1 mm copper and passed through a model of human head. Fig. 2.2(a) shows the percentages of double counts induced by fluorescence escape photons in the CdTe detectors. The percentages decrease with increasing threshold and become zero at around 30 keV. Fig. 2.2(b) shows the percentages of double counts as a function of the electronic threshold, taking into account all the physical effects in CdTe. The total double-counting percentages are high for all the pixel geometries at low thresholds, and it is more than 100% for small pixels, indicating that the leaked charge of a single interacting photon has the possibility of being shared by more than one adjacent pixel as a result of fluorescence or of the diffusion to multiple adjacent pixels. The results explain why a standard threshold level in photon-counting CdTe detectors is relatively high. However, setting the threshold too high will reduce the count efficiency as a result of energy loss of the primary photons.

The results of the energy-loss evaluation in CdTe are shown in Fig. 2.3(a). Nearly all events lose at least 1 keV and about 20%, 34%, and 52% of events lose more than 10 keV for large, mid-sized, and small pixels, respectively. All three curves have inflexions at around 24 keV, and the percentages of energy loss decrease more at these inflexions as a result of the sudden disappearance of fluorescence in this region. Fig. 2.3(b) shows the count efficiencies in CdTe detectors. The figure shows how the count efficiency is decreased with increased threshold and also how the count efficiency varies with pixel size; the larger the pixel size, the higher the count efficiency.

The total proportions of escaped fluorescence photons are 31.80%, 21.89% and
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Figure 2.2: (a) Percentages of double counts induced by only escaped fluorescence photons in a pixelated CdTe detector for three pixel geometries with an incident x-ray spectrum of 120 kVp. (b) Amounts of double counts including all the effects in CdTe with an incident x-ray spectrum of 120 kVp.

Figure 2.3: (a) Percentages of events whose lost energy is larger than a certain energy in pixelated CdTe detectors with primary x-ray energies from a 120 kVp spectrum. (b) Count efficiencies in pixelated CdTe detectors for the 120 kVp tungsten spectrum.

13.46% of the number of primary interacting photons for small, mid-sized, and large pixels, respectively, and 26.54%, 16.46% and 7.94% of the escaped fluorescence photons are reabsorbed by the surrounding pixels and contribute to charge sharing. The distributions of the positions of reabsorbed fluorescence photons in the surrounding pixels of the detectors are shown in Fig. 2.4. The range of escaped
fluorescence for smaller pixels is more than one pixel. Fig. 2.5 shows the distributions of collected energies in a CdTe detector for the three pixel geometries with the corresponding incident monochromatic x-rays of 60 keV and 100 keV. The energy of most of the collected events deviates from the actual energy of the incident photons, most notably in the photo peaks of all the spectra, which are shifted toward the lower-energy side, an effect caused mainly by incomplete charge collection. The amount of double counting is inversely related to pixel size; the photo peak is higher in a large pixel than in a smaller one. The photo peaks appear at around 35 keV and 75 keV for different incident energies produced by the escape of fluorescence photons between 20 keV and 32 keV. The asymmetric distortion of photo peaks is caused by incomplete charge collection, a phenomenon known as “hole tailing”. The tailing effect is also observed below the escape peaks. The tailing effect is more evident for 100 keV photons than for 60 keV photons as a result of the longer drift time of holes arising from the deeper interaction positions of high-energy photons. The same phenomenon explains why the photo peaks from 100-keV photons are lower than those from 60-keV photons.

Fig. 2.6 shows the spectra of collected energies from the incident 120 kVp x-ray spectrum. Compared to the actual photon energy, the collected energies shift toward the lower-energy side for all three pixel geometries. The energy shift is more severe for the smaller pixels as a result of greater energy loss from charge sharing. A high number of double counts from other pixels was imported and located at the low-energy band, mainly below 60 keV on top of the collected energy spectra produced by primary photons in the investigated center pixel. The peaks near 25 keV represent the fluorescence photons from surrounding pixels. Charge sharing is important in the loss of energy resolution. For small and mid-sized pixels in Fig. 2.6, the events in the low-energy range introduced by charge sharing of neighboring pixels are similar in frequency to those of primary events.
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Figure 2.5: Simulated collected energy spectra from CdTe detectors for incident monochromatic x-ray of 60 keV and 100 keV. The energy spectra of charge-shared events from neighboring pixels are shown separately, as dashed curves. (a) 60 keV, small pixel; (b) 60 keV, mid-sized pixel; (c) 60 keV, large pixel; (d) 100 keV, small pixel; (e) 100 keV, mid-sized pixel; and (f) 100 keV, large pixel.

Figure 2.6: Simulated energy spectra for an incident x-ray spectrum of 120 kVp in a CdTe detector for three pixel sizes: (a) small, (b) mid-sized and (c) large. Solid curves A: the distributions of actual photon energies; dashed curves B: the collected energy spectra; dash-dot curves C: the energy spectra of double counts from neighboring pixels; dotted curves D: the superimposition of B and C.
2.4.2 Segmented Silicon Strip Detector

The fractions of double counts caused by diffusion and the initial charge carrier distribution in silicon for three pixel geometries assuming an incident 120 kVp x-ray spectrum are shown in Fig. 2.7(a). The amount of double counting decreases rapidly with increases in the electronic threshold and increases as pixel size is reduced. The value is around 4.5% for the smallest pixel at a threshold of 3 keV and well below 4% for all the geometries at the 5 keV threshold. The charge sharing induced by Compton scattering shows a higher fraction of double counts for larger pixel sizes (Fig. 2.7(b)). The reason is that the thicker silicon wafer for larger pixel size absorbs more scattered photons; however, most of the scattered photons escape from the detector wafer for all three geometries.

The total percentages of double counting are shown in Fig. 2.8(a). Because of the opposing signs of the effects from diffusion and Compton scattering, there are no obvious differences in the total amounts of double counting among the three geometries. The count efficiencies of the segmented silicon strip detectors for the three geometries do not differ noticeably (Fig. 2.8(b)). However, the count efficiencies decrease markedly when increasing the threshold slightly in the low-energy region, implying that the threshold in the segmented silicon strip detector in CT applications should be set as low as possible.

The above results show that for segmented silicon strip detector under the investigated bias voltages, the amount of energy loss and charge sharing is almost independent of the pixel size. Therefore, only the energy spectra of mid-sized pixel are shown in Fig. 2.9 for 60 keV monochromatic and 120 kVp polychromatic x-rays, respectively. In Fig. 2.9(a), the photo peak is very narrow and the peak width is limited by the energy bin size in the simulation (0.5 keV) and by the statistical fluctuation in the number of charge carriers initially released. The amount of events shared from neighboring segments is low and does not noticeably affect the energy resolution. In Fig. 2.9(b), only a small amount of the high-energy charge shared events are mixed with the useful counts. Most of the charge shared events are located in the low-energy region and can be eliminated by setting a proper threshold. As seen in Fig. 2.9(b), the collected energy spectrum deviates from the incident energy spectrum as a result of a high fraction of Compton scattering; more than 60% of the interacting photons experienced Compton scattering for the 120 kVp x-ray spectrum with the specific filtration and deposited only a small amount of the photon energy.

2.4.3 Shape of Initial Charge Cloud in Silicon

Fig. 2.10(a) shows the energy distributions of 1000 photons with 60-keV energy being photo-absorbed 4 μm from the border between two neighboring pixels in a silicon strip detector [62]. For the same interaction position of incident photons, the spherical approximations always result in the similar collected energy in the investigated pixel, with an uncertainty induced by the statistical fluctuation of the
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Figure 2.7: (a) Percentages of double counts introduced by the initial charge distribution and charge diffusion in the silicon detector for three pixel geometries with an incident x-ray spectrum of 120 kVp. (b) Percentages of double counts introduced by Compton scattering in the silicon detector for three pixel geometries with 120 kVp x-rays.

Figure 2.8: (a) The total amount of double counts in the segmented silicon strip detector with an incident x-ray spectrum of 120 kVp. (b) Count efficiencies in a segmented silicon strip detector as a function of threshold for three pixel geometries with 120 kVp x-rays.

number of initially released charge carriers and the diffusion process. For Monte Carlo simulation, the direction of Monte Carlo simulated charged tracks is crucial and determines the energy of detected events. In addition to a peak located at the energy of incident photons of 60 keV, other events were detected with variable
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Figure 2.9: Simulated collected energy spectra in silicon detectors with mid-sized pixel for (a) 60 keV monochromatic energy and (b) 120 kVp x-ray beam. Solid curve A: the distribution of actual photon energies; dashed curve B: the collected energy spectrum; dash-dot curve C: the energy spectrum of double counts from neighboring pixels; dotted curve D: the superimposition of B and C.

Figure 2.10: (a) Distributions of collected energy inside a pixel when interactions occurred 4 µm from the neighboring pixel. All photons were assumed to deposit 60 keV energy. (b) Probability of double counting as a function of interaction distance from the boundary of two neighboring pixels.

energy over the whole abscissa due to random distributions of individual tracks. Fig. 2.10(b) shows the probability of double counting as a function of interaction distance from the border of two pixels. The deposited energy was assumed to be 60 keV and the threshold is 5 keV, i.e. if the detected energy is less than 5 keV, the event will not be registered. For spherical approximations, all the events will be
detected by the neighboring pixel if the interaction site is close to the border of two pixels and no events will be detected if the distance is increased further. As can be seen from the Monte Carlo simulation in Fig. 2.10(b), the charge sharing affects the detector response up to 25 µm to the border in this case, which is underestimated by spherical approximations.

A statistical model has been proposed to simulate the distribution of initial charge cloud in silicon detectors for the photons in the energy range of medical imaging. An initial charge cloud can be generated by sampling the center of gravity and the track size from statistical distributions derived from Monte Carlo generated tracks and by distributing a certain fraction (68%) of the photon energy into a bubble uniformly and by distributing the remaining energy uniformly along a line [Paper II]. Compared to the spherical approximations of initial charge cloud, the statistical model can simulate the detector response accurately and corresponds to the Monte Carlo simulation results.
Chapter 3

ASIC Characterization

3.1 Introduction

A clinical CT system is usually running under very high photon fluxes, up to several million photons s$^{-1}$ mm$^{-2}$, where the expected benefits available for photon-counting spectral CT would be removed by the deterioration of energy resolution as a result of pulse pileup [64]. Therefore, an ultra-fast application specific integrated circuit (ASIC) is needed to mitigate pileup effect, and retain count rate linearity and energy resolution even under high photon flux. Two generations of an ultra-fast ASIC have been developed for photon-counting spectral CT. Paper III and IV give a detailed description of these two generations of the ASIC and the corresponding characterizations.

The ASIC is designed to be used with the segmented silicon strip detector [Paper V]. The ASIC chip has 160 individual channels. Each channel consists of an analog channel, 8 comparators, each with a threshold generated by a global 8-bit digital-to-analog converter (DAC) and a digital channel. The analog channel, illustrated in Fig. 3.1, includes a charge sensitive amplifier (CSA), a pole-zero cancellation circuit and amplifier, a pulse shaping filter implemented as two Gm-C filters and an offset calibration circuit. The pulse shaping filter can be set to 4 different peaking times, 10, 20, 40 and 60 ns. A previous work showed that the 40 ns peaking time is needed to reject cross talk induced by charge collection in neighboring detector diodes [62]. Therefore, only the measurements with this peaking time are discussed in this thesis.

A switched offset calibration mechanism was employed in the first generation of the ASIC to calibrate the analog channel before each measurement cycle to reduce threshold dispersion and to compensate for temperature variations. It is implemented as two feedback loops around each Gm-C filter, each creating a correction voltage on a capacitor, which is stable for a few hundred microseconds. The first generation of the ASIC was evaluated in Paper III and the results showed that with the switched offset calibration the channel-to-channel threshold dispersion was not
small enough and a rate-dependent DC offset was created. Therefore, in the second generation of the ASIC [Paper IV], the switched offset calibration is replaced by a continuous-time nonlinear filter circuit, which is similar to the baseline holder (BLH) circuit [74].

The digital channel contains eight 8-bit counters associated with respective comparators and some timing circuits to manage measurement sequence and filter reset [75]. In addition, there is a global digital part managing overall control, including DAC control and various configurations, and the readout process.

### 3.2 Dead Time Behavior

The count rate performance and dead time behavior of the detector are mainly determined by the timing circuits of the ASIC. After pulse shaping, any incoming pulse with amplitude higher than the lowest threshold is sampled at negative clock edges during a programmable time period (sample time $T_S$) and the pulse height (i.e. the photon deposited energy) is detected by comparing with eight comparators. After the sample time, a filter reset function can be enabled to reduce the risk of pulse pileup by forcing the remaining pulse to zero during another programmable time period (reset time $T_R$), which is achieved by shorting the capacitors in the two Gm-C filters. If the filter reset function is disabled the ASIC is simply halted for $T_R$ and a 2 clock-cycle internal delay ($T_D$) until the pulse tail is skipped (below the lowest threshold) to avoid double counting of the same single pulse. Totally the ASIC needs a detection period of $T_C = T_S + T_R + T_D$ to record a count for both the reset enabled and disabled modes.

When the filter reset function of the ASIC is enabled, the dead time behavior of the ASIC approximates a nonparalyzable model [75]. However, when the filter reset function is disabled, none of the existing models can satisfactorily describe the dead time behavior of the ASIC. A semi-nonparalyzable dead time model has been developed based on the principles of the ASIC timing for the filter reset disabled
3.3 Characterization Methods

The characterization of the ASIC was performed on a test board (Fig. 3.3) with 64 ASIC channels wire bonded to the central strips of a silicon strip sensor with 0.5 mm thickness, 50 µm strip width and 11.3 mm strip length [76]. A bias voltage of 100 V was applied to the backside of the detector for the measurements in this thesis. The detector was mounted in a light-tight box to prevent the disturbances from natural light. The commands input, data output and the clock of the ASIC
Bias input
Sensor ASIC
Test MUX 
Vddd Ground V dda
Electrical pulse input

Figure 3.3: A photograph of the test board with 64 channels of the ASIC wire bonded to a 50 µm strip width silicon strip sensor.

were managed by a field-programmable gate array (FPGA) card via a low-voltage differential signaling (LVDS) interface.

The first generation of the ASIC was evaluated in Paper III with electrical pulses and a pulsed laser mainly to verify the functionality of the ASIC. For the second generation of the ASIC, a comprehensive characterization was carried out by synchrotron radiation at beamline B16 of the Diamond Light Source with different monochromatic energies and various photon fluxes [Paper IV]. The count rate performance of the ASIC was also investigated with 120 kVp polychromatic x-rays produced by a tungsten anode tube at different fluxes. During the evaluation with x-ray photons, the sensor was irradiated from the strip side to facilitate the ASIC protection and setup alignment.

The performance of the ASIC was characterized by threshold scanning. For a certain amount of the energy injected into the input of an ASIC channel, either through electrical pulses or laser or x-ray photons, the lowest threshold was fixed at a low value above noise floor to start sampling of the incoming pulses, the second threshold was scanned and all higher thresholds were evenly distributed above the scan range of the second threshold. The threshold offsets and noise levels (or RMS energy resolution) of the ASIC were derived from the resulting S-curves. After obtaining the threshold offsets for different photon energies, the gain and threshold offset at zero keV of a certain ASIC channel were estimated by fitting a line to the threshold offsets corresponding to different photon energies. For the sensor with 50 µm strip width, the significant charge sharing effect makes the width of the measured energy spectra much larger than the internal noise level of the ASIC. Therefore, a deconvolution procedure was adopted to remove the charge sharing induced spectrum distortion and obtain the internal noise level of the ASIC by deconvolving the measured spectrum with the simulated energy spectrum.
3.4. RESULTS AND DISCUSSION

Fig. 3.4 shows the detector response produced by electrical pulses with the eight thresholds spread quasi-evenly across the whole DAC range and the intensity of electrical pulses swept from low to high, which clearly demonstrates the energy resolving capability of the ASIC.

Fig. 3.5(a) shows the gain value of the ASIC as a function of detector strip number. The average gain over 64 connected channels is 1.65 mV/keV, and the standard deviation of gain values is 0.03 mV/keV, which gives a gain variation of 2.0%. The central detector strips (also the central ASIC channels) show smaller gain than the outer ones. A possible reason for this is a slightly lower transconductance of the transistors in the filters of the central channels, which will influence the gain through a mismatch in the time constants of the amplifier and the filters. The transconductances are controlled by a common bias voltage, generated by a central current mirror. However, because of voltage drops in the ground distribution network (which carries quite large currents from chip edges to the middle of the chip), we can expect a systematic error in the local bias voltages (measured relative to local ground), causing the observed gain variation. Fig. 3.5(b) shows the distribution of threshold offsets at zero keV. The standard deviation of threshold offsets at zero keV is 1.47 mV (0.89 keV) which shows an improvement from the first generation of the ASIC with a value of 3.31 mV [Paper III].

Fig. 3.6(a) shows the noise level of the ASIC in mV as a function of strip number for two different CSA feedback resistance values measured with 15 keV photons at a
CHAPTER 3. ASIC CHARACTERIZATION

Figure 3.5: (a) Gain variation of the ASIC showing the gain values for different strips. (b) Threshold offset at zero keV as a function of detector strip number. The dashed line shows the average value and the dash-dot lines illustrate the dispersion (± standard deviation).

Figure 3.6: (a) RMS noise level in mV as a function of strip number measured with 15 keV photon energy. (b) RMS energy resolution of individual ASIC channels as a function of output count rate for 15 keV photons.

flux of 40 kcps where the pulse pileup effect is nearly negligible. The average noise level over 64 channels is 1.81 mV (1.09 keV) for 3.0 MΩ feedback resistance and 2.11 mV (1.28 keV) for 0.7 MΩ. Measurements done at different photon energies indicate that the noise level of the ASIC are almost independent of the incident photon energy. Fig. 3.6(b) shows the values of RMS energy resolution of individual ASIC
3.4. RESULTS AND DISCUSSION

Figure 3.7: (a) Count rate linearity of an individual ASIC channel for 120 kVp x-rays with $T_C = 100$ ns. (b) The energy spectra measured with the ASIC connected to a 50 $\mu$m strip-width sensor for different photon interaction rates from 120 kVp x-rays.

channels as a function of output count rate. The energy resolution is deteriorated due to the effect of pulse pileup when the count rate is increased. A linear fitting $\sigma = \Delta \sigma \cdot m + \sigma_0$ was used to fit the RMS energy resolution of the ASIC as a function of output count rate ($m$) for both 3.0 and 0.7 M$\Omega$ CSA feedback resistances, where $\sigma_0$ is the RMS energy resolution of the ASIC at zero output count rate and $\Delta \sigma$ is the deterioration rate of the RMS energy resolution as a function of output count rate. The RMS energy resolution is 1.09 keV at zero output count rate for 3.0 M$\Omega$ CSA feedback resistance, and it deteriorates as a function of output count rate at a rate of 0.29 keV/Mcps. The RMS energy resolution at zero output count rate and the deterioration rate are 1.27 keV and 0.32 keV/Mcps, respectively, for 0.7 M$\Omega$ CSA feedback resistance.

Fig. 3.7(a) shows the output count rate of an ASIC channel as a function of input count rate with $T_C = 100$ ns and the fitting with the semi-nonparalyzable dead time model (i.e. Eq. (3.1)). The average nonparalyzable dead time $\tau$ over all 64 evaluated channels is $-10.15 \pm 3.62$ ns for $T_C = 100$ ns and $32.96 \pm 5.24$ ns for $T_C = 120$ ns. A possible reason for the negative value of $\tau$ with $T_C = 100$ ns is that for polychromatic incident energy spectrum at high photon flux the events with energy below the lowest threshold might pile up on top of each other, activate a detection period and lead to an additional detectable count. There is also a possibility of double counting of the same single pulse for a short detection period of $T_C$.

Fig. 3.7(b) shows the measured energy spectra (average of 64 channels) under different photon interaction rates in each silicon sensor strip from 50 kcps to 5 Mcps (equivalent to photon fluxes from 5 to 500 Mphotons s$^{-1}$ mm$^{-2}$ in the segmented
silicon strip detector (Paper V)) with $T_C = 100$ ns. For low photon fluxes without the influence of pulse pileup, the distribution of deposited energies is highly skewed towards low energies due to the short photon interaction length of 500 $\mu$m and the severe charge sharing effect in the sensor with 50 $\mu$m strips. For higher photon fluxes pileup skews the distribution towards higher energies, which is consistent with the increased energy uncertainty with higher count rates of Fig. 3.6(b). However, for fluxes up to 150 Mphotons s$^{-1}$ mm$^{-2}$ the distortion is not severe, indicating that the energy resolution is retained to a large extent. The experiment was repeated for $T_C = 120$ ns. The deterioration of the energy spectrum is a little faster than that with $T_C = 100$ ns due to the longer detection period to detect each incoming event, but for fluxes up to 150 Mphotons s$^{-1}$ mm$^{-2}$ the spectrum distortion is still small.

A photon-counting CT image obtained with the first generation of the ASIC and 50 $\mu$m strip sensor is shown in Fig. 3.8. The nearly ring free image demonstrates that, at least in pure photon-counting mode, the relatively large threshold dispersion in the first generation of the ASIC can be calibrated away. A better image quality is expected for the second generation of the ASIC which has less channel-to-channel threshold dispersion.
Chapter 4

Segmented Silicon Strip Detector

4.1 Introduction

Paper V describes a silicon photon-counting energy-resolving detector developed for spectral CT. The detector modules are fabricated on high-resistivity n-type silicon substrate with p-type electrodes implanted. With the silicon sensor as base substrate, a Multi-Chip Module (MCM) technology is employed to integrate the ASICs and electric routing.

The detector width of 20 mm is divided into 50 strips with a strip pitch of 0.4 mm. Like most of the other silicon strip detectors applied in high-energy physics and particle physics, an edge-on geometry of the detector is adopted to provide a long absorption path of 30 mm for high energy photons encountered in clinical CT (up to 120 keV). The detector thickness is 0.5 mm. Consequently, an x-ray entrance (i.e. pixel size) of $0.5 \times 0.4 \text{ mm}^2$ is constituted for each detector strip. In order to mitigate the problem of high photon flux, each detector strip is divided into 16 segments with exponentially increasing lengths along the x-ray incident direction ensuring similar count rates in all 16 segments, which is achieved by implanting 16 individual p-type electrodes instead of a whole strip electrode onto the silicon substrate, and each electrode is connected to an individual ASIC channel. The count rate in each segment and the corresponding ASIC channel is thus reduced by a factor of 80 compared to the incident photon flux per square millimeter per second because of the segmented structure of the detector strips and the pixel size of $0.5 \times 0.4 \text{ mm}^2$. Fig. 4.1 shows a photograph of the detector module together with a schematic illustration of two detector segments. Five 160-channel ASICs are stud bonded to each detector module using a flip-chip technology to manage 800 detector elements on each detector module.

The first version of the detector module has been manufactured and the evaluation has been carried out at the beamline SYRMEP of the Elettra synchrotron light source. The results presented in Paper V shows how we solved the problem of high photon flux for a clinical CT system with the segmented silicon strip detector.
4.2 Characterization Methods

Paper V shows the detector evaluation performed at the beamline SYRMEP at the Elettra synchrotron light source. The detector was aligned in edge-on geometry and an ionization chamber was positioned in front of the detector to monitor the time-dependent variation of photon flux. A thermocouple was attached to the detector to record the temperature variation during the measurement. A bias voltage of 160 V was applied to the backside of the detector. The gain, threshold offset, energy resolution and the amount of charge shared events were estimated using the threshold scanning method.

4.3 Results and Discussion

The gain and threshold offset at zero keV were calibrated with four different photon energies of 22, 26, 30 and 34.9 keV. A narrow beam of 6 mm width struck several strips of the detector edge-on. Since the low-energy photon beam only penetrated a few segments of the detector, the response of the first 4 segments were investigated based on the distribution of interacting photons in the detector. The average
gain over 44 evaluated detector elements is 1.65 mV/keV and the gain variation calculated as the standard deviation divided by the mean is 3.7%. The standard deviation of threshold offsets at zero keV is 2.36 mV (1.43 keV). The gain variation and threshold dispersion are larger than the results measured on the test board with only one ASIC mounted [Paper IV], probably caused by the non-uniformity of the power distribution to different ASICs on MCM.

Fig. 4.2 shows the output count rate as a function of input count rate for individual detector elements measured with 22 keV photons. Fitting the semi-nonparalyzable model (Eq. (3.1)) to the measured relationship between input and output count rates gives an estimated nonparalyzable dead time of $79.8 \pm 1.9$ ns. Although the reset function of the ASIC was not enabled, the count rate performance of the detector is still very promising. The count rate performance of the detector is a function of photon-deposited energy, setting of thresholds and, most importantly, the timing of the ASIC, and a significant improvement is expected with 120 kVp polychromatic x-rays [Paper IV].

Fig. 4.3 shows the RMS energy resolution ($\sigma$) in keV for individual detector elements as a function of input count rate. The energy resolution of the detector gradually deteriorates as the increase of photon flux due to the effect of pulse pileup. All the evaluated detector elements in the first segment, two in the third segment and one in the fourth segment have higher noise levels than the other detector elements, because part of the routing wires connecting these detector elements and the ASICs is run in a metal layer with an insulator thickness of about 1.5 $\mu$m to
the silicon substrate, smaller than the 10 \( \mu \text{m} \) insulator thickness for the wires of the other elements running in another metal layer, thus having higher capacitances and easier to couple the noise from the diffused ground plane in the MCM. A linear fit \( \sigma = \Delta \sigma \cdot n + \sigma_0 \) was applied to the data set of each segment, with \( n \) being the input count rate, \( \sigma_0 \) the RMS energy resolution at zero flux and \( \Delta \sigma \) the deterioration rate of the energy resolution for a single detector segment in units of keV/Mcps. The values of \( \sigma_0 \) and \( \Delta \sigma \) for the first 4 segments and the average values of the four segments are shown in Table 4.1. An RMS energy resolution of 1.50 keV has been determined with 22 keV photons at zero flux giving an energy resolution (FWHM/\( E \)) of 16.1%. Previous tests showed that the RMS energy resolution was quite independent of the incident photon energy [77], thus a superior energy resolution is expected for higher photon energies. The average deterioration rate of the energy resolution for each detector segment as a function of input count rate for 22 keV photon energy is 0.41 keV/Mcps, which translates to a deterioration rate of 5.13 eV mm\(^2\)/Mcps for the studied detector layout with 16 depth segments and a pixel size of 0.5 \( \times \) 0.4 mm\(^2\). The energy resolution of the detector would be 21.5% for 22 keV photon energy at an input count rate of 100 Mcps mm\(^{-2}\).

The fraction of charge shared events was calculated and the results are presented in Table 4.2. A fraction of 11.1% of the interacting photons experienced charge sharing for the evaluated detector, and increased to 15.3% for 30 keV. As a result of energy loss, these charge shared events distort the energy spectrum collected by the detector, but they are still useful events if the system is operated in pure photon-counting mode. The energy leaked into the neighboring detector element

Figure 4.3: RMS energy resolution versus input count rate for each detector segment with an incident photon energy of 22 keV and the linear fits.
might be detected and result in noise counts if the leaked energy is higher than the lowest threshold of the neighboring detector element. Properly setting the threshold could help improve the system performance. Various effects result in charge shared events, including diffusion of charge carriers, distribution of initial charge cloud and Compton scattering. The diffusion effect was found to dominate the amount of charge shared events, thus the fraction of charge shared events is expected to be reduced by applying a higher bias voltage to the detector.

Table 4.1: The RMS energy resolution at zero flux ($\sigma_0$) and the deterioration rate of the energy resolution ($\Delta \sigma$) for each detector segment for 22 keV photon energy.

<table>
<thead>
<tr>
<th></th>
<th>$\Delta \sigma$ (keV/Mcps)</th>
<th>$\sigma_0$ (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Segment-1</td>
<td>0.48 ± 0.02</td>
<td>1.85 ± 0.01</td>
</tr>
<tr>
<td>Segment-2</td>
<td>0.31 ± 0.02</td>
<td>1.36 ± 0.01</td>
</tr>
<tr>
<td>Segment-3</td>
<td>0.37 ± 0.07</td>
<td>1.40 ± 0.03</td>
</tr>
<tr>
<td>Segment-4</td>
<td>0.48 ± 0.09</td>
<td>1.39 ± 0.02</td>
</tr>
<tr>
<td>Mean</td>
<td>0.41 ± 0.03</td>
<td>1.50 ± 0.01</td>
</tr>
</tbody>
</table>

Table 4.2: Fraction of charge shared events (%) for two photon energies and the first two detector segments.

<table>
<thead>
<tr>
<th>Energy (keV)</th>
<th>Segment-1</th>
<th>Segment-2</th>
<th>Mean</th>
</tr>
</thead>
<tbody>
<tr>
<td>22</td>
<td>9.8 ± 1.0</td>
<td>12.6 ± 1.0</td>
<td>11.1 ± 0.7</td>
</tr>
<tr>
<td>30</td>
<td>14.4 ± 0.8</td>
<td>16.2 ± 0.7</td>
<td>15.3 ± 0.5</td>
</tr>
</tbody>
</table>
Chapter 5

Conclusions

“Why use silicon? Why not CdTe or CZT?” This is a most frequently asked question from the researchers who are developing photon-counting spectral CT based on CdTe/CZT detectors. I am not surprised at all to hear this due to the well-known high-fraction Compton interactions in silicon in the energy range of x-ray CT. The answer to this question is obvious according to the evaluations and analyses presented in this thesis.

Actually silicon has properties which make it more attractive than CdTe/CZT as a candidate detector material for photon-counting spectral CT. The problem of extremely high photon flux, what prevents a photon-counting energy-resolving detector from being used in a clinical CT system, can be properly solved by adopting a segmented detector structure and operating the detector in edge-on geometry. The low photon detection efficiency in silicon can be compensated for by increasing the detector length. A previous study showed that the segmented silicon strip detector could perform comparably to the ideal energy-integrating detector for routine CT imaging tasks [24]. In addition, the energies deposited by photoelectric and Compton interactions are well separated without an overlap, indicating that statistical methods can be applied to make maximum use of the energy information remained in Compton events. The results presented in Chapter 2 further demonstrated that the pixel size of the investigated silicon detector could be reduced without the influence of charge sharing by simply keeping the intensity of electric field for different pixel sizes constant, which could help improve the spatial resolution of the system in the future.

For CdTe/CZT detectors, the primary challenge is how to handle the high photon flux encountered in clinical CT applications. Currently, no other way than decreasing pixel size has been put into practice [57, 58, 78], which is not effective enough to fulfill the high flux requirements and also means an increased spectrum distortion as a result of charge sharing and fluorescence escape, as demonstrated in Paper I. An edge-on multiple-layers arrangement which is similar to the detector structure presented in this thesis has been proposed in simulation [61], whereas
the real implementation is much more difficult than that with silicon. Polarization, another serious limitation of CdTe/CZT detectors, caused by low hole mobility and hole trapping at high photon flux, is also difficult to be addressed even with edge-on multiple-layers structure.

Both a novel detector structure and an ultra-fast ASIC are needed to develop a CT scanner with photon-counting energy-resolving capabilities. A segmented silicon strip detector has been designed. By operating the detector in edge-on geometry, the count rate in each detector segment and the corresponding ASIC channel can be reduced by a factor of 80 compared to the incident photon flux per square millimeter per second with the 16 detector segments along the x-ray incident direction and a pixel size of $0.5 \times 0.4 \text{ mm}^2$.

An ultra-fast photon-counting energy-resolving ASIC has been developed to process the photon-converted pulses in each individual detector segment. The synchrotron measurement demonstrated that the ASIC performed as expected. A noise level of 1.81 mV RMS (1.09 keV) has been measured, which is close to the simulated value of 1.0 mV. A channel-to-channel threshold dispersion of 0.89 keV RMS can be calibrated away by using a novel compensation method [79]. The dead time behavior of the ASIC approximates a nonparalyzable model in the filter reset enabled mode and follows a semi-nonparalyzable model when the filter reset is disabled. A promising count rate performance has been achieved in terms of the count rate linearity and energy resolution, i.e. the count loss and deterioration of energy resolution with the increase of photon flux are acceptable for the operational range of photon flux in clinical CT.

The first-version segmented silicon strip detector with 5 ASICs stud bonded was characterized using synchrotron radiation. An energy resolution of 16.1% has been determined with 22 keV photons at zero flux, which deteriorates to 21.5% at an input count rate of 100 Mcps mm$^{-2}$ according to the deterioration rate of RMS energy resolution of 5.13 eV mm$^2$/Mcps. The noise level of this version of the detector is higher than expected as a result of the imperfect ASIC power connection and detector layout. Lowering the noise of the detector is an ongoing task which is crucial for the performance of the whole CT system. The fraction of charge shared events has been estimated and found to be 11.1% for 22 keV and 15.3% for 30 keV. Since the diffusion of charge carriers dominates the amount of charge sharing, a lower fraction of charge shared events and an improved energy resolution can be expected by applying a higher bias voltage to the detector.
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Bibliography


