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Characterisation and application of photon counting X-ray detector systems

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To my sons, Johan and William, and to my wonderful wife Monica
ABSTRACT

This thesis concerns the development and characterisation of X-ray imaging systems based on single photon processing. “Colour” X-ray imaging opens up new perspectives within the fields of medical X-ray diagnosis and also in industrial X-ray quality control. The difference in absorption for different “colours” can be used to discern materials in the object. For instance, this information might be used to identify diseases such as brittle-bone disease. The “colour” of the X-rays can be identified if the detector system can process each X-ray photon individually. Such a detector system is called a “single photon processing” system or, less precise, a “photon counting system”.

With modern technology it is possible to construct photon counting detector systems that can resolve details to a level of approximately 50 µm. However with such small pixels a problem will occur. In a semiconductor detector each absorbed X-ray photon creates a cloud of charge which contributes to the image. For high photon energies the size of the charge cloud is comparable to 50 µm and might be distributed between several pixels in the image. Charge sharing is a key problem since, not only is the resolution degenerated, but it also destroys the “colour” information in the image.

This thesis presents characterisation and simulations to provide a detailed understanding of the physical processes concerning charge sharing in detectors from the MEDIPIX collaboration. Charge summing schemes utilising pixel to pixel communications are proposed. Charge sharing can also be suppressed by introducing 3D-detector structures. In the next generation of the MEDIPIX system, Medipix3, charge summing will be implemented. This system, equipped with a 3D-silicon detector, or a thin planar high-Z detector of good quality, has the potential to become a commercial product for medical imaging. This would be beneficial to the public health within the entire European Union.
SAMMANDRAG

Denna avhandling berör utveckling och karaktärisering av fotonräknande röntgensystem. ”Färgröntgen” öppnar nya perspektiv för medicinsk röntgendiagnostik och även för materialröntgen inom industrin. Skillnaden i absorption av olika ”färger” kan användas för att särskilja olika material i ett objekt. Färginformationen kan till exempel användas i sjukvården för att identifiera benskörhet. Färgen på röntgenfotonen kan identifieras om detektorsystemet kan detektera varje foton individuellt. Sådana detektorsystem kallas ”fotonräknande” system.


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As a child, my parents read physics books to me, and I looked at the fascinating pictures of the Bohr model and of the solar system. At the age of 11, I became active against nuclear power and decided to work in the field of radiation in the future. After six years of accident simulations for the nuclear industry I was more comfortable with nuclear power, since I found no design error of significance in the Swedish nuclear power plants. My friend Dr. Jan-Olov Andersson persuaded me to move north to Sundsvall and to the academic world, which has made it possible for me to explore the area of X-ray imaging.

I want to express my gratitude to my supervisors. Christer Fröjdh has shown large emotional involvement in my work, I felt it especially during my hospital stay when I also learned that Christers wife Maj is one of the best nurses in the children department at Sundsvall hospital. Hans-Erik Nilsson has been deeply involved in my work when he has had the chance to develop advanced simulations. For the published work in this thesis, my co-authors are; Ervin Dubarić, Anatoliy Manuilskiy, Heinz Graafsma, Cyril Ponchut, Vedran Vonk and Lukas Tlustos.

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Without the understanding from my wife Monica this thesis would probably not have been finished. Thanks to my parents in law, Anna and Tage, who stayed with the children several times during writing and research. My sons Johan (6 years) and William (3 years) also promised to be good, but only until the book is finished, as Johan expressed it. William wondered if the tooth felt pain, while Johan said he was lucky to have this book to read when he gets older.
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LIST OF PAPERS

This thesis is based on the following papers, herein referred to by their Roman numerals:

Paper I  Monte Carlo simulation of the response of a pixellated 3D photodetector in silicon

Paper II  Spectroscopy applications for the Medipix photon counting X-ray system

Paper III  Material recognition with the Medipix photon counting colour X-ray system

Paper IV  Energy dependence in dental imaging with Medipix2

Paper V  Spectral Response of Pixelated Semiconductor X-ray Detectors

Paper VI  Characterization of a pixellated CdTe detector with single-photon processing readout

Paper VII  Characterisation of the charge sharing in pixellated Si detectors with single-photon processing readout
Paper VIII  Spectral performance of a pixellated X-ray imaging detector with suppressed charge sharing

Paper IX  Charge sharing suppression using pixel to pixel communication in photon counting X-ray imaging systems

Related papers not included in the thesis:

Monte Carlo simulation of the imaging properties of scintillator-coated X-ray pixel detectors

Metallized and Oxidized Silicon Macropore Arrays Filled With a Scintillator for CCD-Based X-ray Imaging Detectors

Performance of Scintillating Waveguides for CCD-based X-ray Detectors

Evaluation of the charge-sharing effects on spot intensity in XRD setup using photon-counting pixel detectors

An Area Efficient Readout Architecture for Photon Counting Color Imaging
1 INTRODUCTION

1.1 THESIS OUTLINE

This thesis studies photon counting X-ray imaging systems. Photon counting, or single photon processing, is a method to improve the performance of X-ray imaging systems as compared to standard charge integrating systems. The first section offers the arguments for research in X-ray imaging and introduces some basic concepts for detector systems. Section 2 introduces the Medipix X-ray imaging system, which is used in most of the experimental work in this thesis. Section 3 is a detailed description of the charge sharing phenomenon [1]. The section presents simulations of absorption and charge transport in small pixellated detectors together with measurements on the Medipix2 system. Methods for image correction are also considered. Section 4 describes the research concerning the improvement of image quality and new possibilities for X-ray imaging applications. The problem associated with charge sharing is also discussed. Section 5 summarizes the work covered by all the papers included in the appendix and section 6 contains the conclusions of this work.

1.2 THE DISCOVERY OF X-RAYS

X-rays were discovered in 1895 by W. K. Röntgen. X-rays are high-energy photons invisible to the human eye. Due to the high energy X-ray photons can penetrate through matter. Mono-energetic X-ray photons are released in the decay chain for radioactive atoms. X-rays are typically generated in an X-ray tube where electrons are accelerated by a high voltage (several thousand volts) and collide with a metal target. In the collisions “bremsstralung” is generated when the paths of the electrons are bent by the interaction with the nucleus of the metal atoms. The momentum energy lost by the electrons is emitted as X-ray photons. The process generates a spectrum with energies distributed between “low” and “maximum”, where the maximum occurs when the full energy of the electron is converted to an X-ray photon, see Figure 1. It is convenient to measure the energy of the photons in electron volts (eV), for a tube voltage of 60 kV the maximum energy of the photons will be 60 keV. The majority of the photons will however receive about half this energy.
1.3 APPLICATIONS FOR X-RAYS

X-rays started to be used for medical diagnosis immediately after they were discovered using a photographic film as the detector. The first medical use of X-rays was to view the skeleton of a human since the X-ray absorption is much stronger in bone than in soft tissue. X-ray tubes are commonly used as sources for clinical imaging. Through means of dental X-rays imaging and mammography almost the entire population is affected by X-ray imaging today. X-rays are of course important not only in medicine but in many research areas such as astronomy, spectroscopy, and crystallography and of course in nuclear physics.

Soon after the discovery of X-rays, negative effects such as eye damage, skin damage and loss of hair were observed due to radiation exposure. Reproducing cells are more sensitive to radiation, so X-rays can be used in the treatment of growing cancer tumours. Cancer cells are more significantly damaged by radiation than are the surrounding normal cells.

1.4 NEGATIVE EFFECTS OF X-RAYS

X-rays have enough energy to ionize atoms in the material they hit and are therefore counted as ionizing radiation. This has several negative influences on the body, but the main concern is the increased risk of developing cancer. It is proven that ionizing radiation induces cancer to human tissue, and that the influence is
linear.\textsuperscript{1} For people who are exposed to radiation at work, a maximum lifetime dose and yearly dose is defined by law. This dose is considered to give a reasonable statistical reduction of the individual’s lifespan, which is considered to be small relative to other negative factors such as stress, depression, lack of physical training, poor eating habits and normal consumption of poisons such as alcohol and nicotine. A statistical reduction in lifespan is a reasonable measure of the risk, since cancer generation is a slow process.

When the risk of a specific radiation exposure has to be estimated, a complication occurs. No lower level has been found for cancer induced by radiation, which is not the same as toxic influences, where a minimum level of the poison is required before the body is affected. An interesting conclusion can be drawn; a lowering of the dose will always result in less cancer development, no matter how low the risk already is. This decrease in risk might not be crucial for the individual, but when the whole population, of for example the European Union is considered, the decreased cost for society due to the cases of induced cancer which are thus avoided is significant.

1.4.1 X-ray dose and mammography

When the cancer-inducing effects of X-rays became known, ways to reduce the dose given to the patient began to be investigated. This involves both the quality of the diagnosis and ways to verify the actual radiation field exposing the patient [3]. For diagnosis, the number of images, the image size and the exposure time (connected to the image quality) can be decreased to minimize the dose.

An example of risk consideration is the discussion about the extent to which mammography examinations should be performed on a regular basis. The key to successful cancer treatment is to discover the cancer at an early stage and remove the tumour while it is still small. Thus if all females are regularly examined then, statistically, most of the breast cancers will be cured successfully. However, if the examinations are performed too often, then the examinations will themselves begin to generate breast cancers, which is unacceptable. This also affects the age at which mammography examinations should start. It is unlikely that a tumour will be found in a younger person, as the risk of developing cancer increases with age. Young females also have denser breasts, and thus a higher radiation dose is required to be able to distinguish a tumour in a young female [4]. To obtain the best outcome for the health of the whole female population, a national mammography programme must carefully consider both the regularity and the initial age for examinations. The statistical risk for some undiscovered cancers in females has to be kept at an acceptably low level.

\textsuperscript{1} These data originates from three sources: patients irradiated for radiology treatment; studies of survivors of Hiroshima and Nagasaki 1945 and studies of workers who were accidentally exposed to radiation [2].
1.4.2 Increasing doses

The trend in clinical imaging over the past few years is that X-ray systems based on photographic film have been replaced by semiconductor detectors. There are several reasons for replacing X-ray films:

1. Digital imaging is fast, since no time for film development is needed.
2. Digital imaging has positive environmental effects due to less use of raw materials (film) and less use of chemicals for film processing.
3. Image processing can be introduced to enhance image quality.
4. Digital images can rapidly be distributed to other parts of the world for expert opinions.
5. Semiconductor detectors can be made more sensitive than film leading to reduced doses to the patients.

Digital images can be achieved by replacing film cassettes in existing devices by storage phosphor plates. These plates are read out by a laser scanner. This is the cheapest way for a hospital to start digital imaging.

When improvements of X-ray equipment are implemented, the choice is often to keep the old dose of the specific examination constant and, instead, gain increased image quality. If the image quality is increased the risk of missing an early tumour decreases. A rapid increase in the dose delivered to patients seems to be taking place in clinical X-ray imaging. This is mainly due to the increased use of CT (Computer tomography) to obtain correct diagnoses in cases where unclear diagnoses were previously accepted. If this trend is sustained, then the research concerning dose reduction, where image quality remains, becomes ever more important.

1.5 Some basics on semiconductor X-ray detectors

The first electric detectors for X-rays were gas-chambers, where the X-ray photons ionized atoms of the gas and created a discharge that could be detected. However, it is more efficient to use a solid material to absorb X-ray photons. When a photon is absorbed in a semiconductor such as silicon, a charge cloud is released. For silicon the conversion factor, or the energy necessary to create one electron hole pair, is 3.6 eV [5]. Normally the detector is designed as a diode with reversed bias applied. In this electric field the electron and the hole will move in opposite directions. When the charge cloud drifts towards the readout electrode a current pulse is generated. The size of this current pulse is proportional to the amount of energy deposited in the detector. The applied reversed bias will result in a leakage current through the diode, causing noise which limits the minimal detectable pulse.

Silicon detectors are efficient for photons in the visible range, but for X-rays the quantum efficiency is low, since the X-ray absorption in silicon is low. To improve the quantum efficiency a highly absorbing scintillator layer can be applied above the detector [6]. A scintillator is a crystal which absorbs X-rays and emits the
absorbed energy as visible light. Several scintillator materials exists, but one commonly used materials is Tl-doped CsI (CsI(Tl)) [10].

1.5.1 Charge coupled devices

The CCD is the state-of-the-art imaging sensor. It consists of a matrix of small MOS capacitors. When a photon is absorbed and induces a charge cloud, the nearest capacitor will receive a charge. To retrieve the images the charge is shifted from pixel to pixel to the readout electrode.

![Charge coupled device](image)

Figure 2: Illustration of the readout principle for a charge coupled device from [7].

The CCD is popular due to its simplicity. Since only three capacitors are required for each pixel, detector matrices with a small pixel size can be easily constructed. This is important since normal X-ray imaging systems do not contain any optical magnification. Relatively large-area CCD detectors can be made at a low cost.

1.5.2 Scintillator grids

CCD’s are usually designed to work within the region of visible photons, the quantum efficiency for X-rays are usually considerably low. In X-ray imaging systems, the detector is usually covered with a scintillator layer. One problem with a scintillator layer is that the light emitted in the scintillator is spread laterally [8], causing the spatial resolution of the detector does deteriorate. Hence, structured scintillator films have been developed to maintain an optimal spatial resolution and sensitivity. One approach is to grow a columnar CsI(Tl) film by evaporation of CsI on a pre-patterned substrate. Light emitted from CsI(Tl) is then confined
within the columns due to total internal reflection, which improves the spatial resolution [9]. Alternatively, advanced etching techniques can be used to create a silicon matrix of pores, which are subsequently filled with a scintillator [10]. The silicon pore walls can be oxidized [11] or metallised [12] in order to obtain an array of scintillating waveguides, whose operating principle is similar to scintillating optical fibres. It has been demonstrated that it is possible to improve the signal to noise ratio close to the theoretical limit for a specific spatial resolution and X-ray dose [13].

Figure 3: Scanning electron micrographs of a pore array filled with CsI(Tl) from [13].

1.5.3 Strip detectors

Strip detectors are popular in high energy physics. Strip arrays in one or two dimensions are suitable for following the path of a high energy particle. Attempts have been made to use detectors with crossed strips for X-ray imaging. In astronomy applications, strip detector systems are successful. The position of a specific X-ray absorption is determined from the coincident pulses between two strips. One problem that occurs is that the speed of the ASIC limits the photon intensity, since only one pulse at a time can be retrieved. For medical imaging this is a problem since the exposure time must be short enough to ensure that the patient does not move. In commercially available medical systems scanning of a moving 1D-strip array is a successful method. The strips are put “edge-on” towards the beam, and the absorption path for the photons becomes the full strip length that is of the order of 1 cm [14].

1.5.4 Pixel detectors

Pixel detectors form the main concept of this thesis. A detector matrix consisting of pixellated diodes can be bump-bonded to a front-end readout chip [15]. Hence the pixel size defines the available space for the pixel readout
electronics. Therefore design of pixellated readout systems requires deep submicron CMOS processes.

Figure 4: Schematics of the Medipix chip bump-bonded to a pixellated detector from [16].

The photon counting technique is well known in nuclear physics. The pulse shape of the discharge signal from a gas-chamber has, for a considerable time, been a common method for retrieving spectra in multi-channel readout systems. But these measurements are performed using a single detector connected to an external device containing the large-size front end electronics.

State of the art imaging systems such as the CCD (or film) works via a charge integrating principle. The sum of the charge accumulated in a pixel corresponds to the total X-ray energy absorbed by that pixel in the image. At present this is the standard technology for imaging applications. Quantum imaging has become possible due to advances in microelectronics, which allows the designs and fabrication of chips with pixellated pulse processing front-end
electronics with low noise. The measured charge pulse defines the energy of the absorbed photon, and it is possible to perform spectroscopic discrimination on each pixel. The front-end electronics is usually bump-bonded to the pixel detector.

Photon counting is a simple way of quantum imaging, when discrimination of low energies is used to cancel the readout noise. Through sweeping the threshold voltage and subtracting consecutive images a full spectrum can be achieved. One problem in pixellated detectors is that the energy threshold will vary between pixels due to variations in the amplifiers. Pixellated systems must therefore be equipped with an individual calibration of the threshold voltage for each pixel. This can be achieved by a threshold adjustment circuit in each pixel. The Medipix system is equipped with a 3-bits register in each pixel, which allows threshold adjustment in eight steps.

Several projects concerning photon counting pixellated detectors exist. An overview with brief descriptions of several projects can be found in [14].
2 THE MEDIPIX PROJECT

The Medipix project is a collaboration between 16 universities and research institutes including Mid Sweden University [16]. The project is coordinated by the CERN microelectronics group.

2.1 BRIEF DESCRIPTION OF MEDIPIX1

The Medipix1 chip, which became available in 1997, was the first full-scale quantum imaging ASIC for hybrid semiconductor pixel detectors [17]. The design was based on previous experiences of pixel chips for high-energy particle trackers. The pixel size is 170 x 170 µm$^2$ and they are arranged in a 64 x 64 matrix. The readout chip is designed in 1 µm SACMOS technology and each pixel is equipped with a preamplifier, a discriminator with 3 bits threshold adjustment and a 15-bit counter. Both Si- and GaAs detectors have been bump-bonded to the chip. The chipboard can be operated using a standard PC equipped with a data acquisition card and an analogue output card connected to an interface circuit called MUROS1 developed by NIKHEF. The user interface software for Medipix1 is Medisoft 3 described in [18].

2.2 BRIEF DESCRIPTION OF MEDIPIX2

The Medipix2 chip is a further development of Medipix1 [19]. One purpose was to reduce the pixel size to 55 x 55 µm$^2$ to become competitive with state of the art CCD detectors and a further purpose was to introduce a high threshold for the energy window discrimination. 0.25 µm CMOS technology is used to construct the chip of size 1.4 x 1.6 cm$^2$ containing 256 x 256 pixels [20]. Dead areas are minimized on three sides of the chip, so several detectors can be tiled to make a large area detector. The readout circuit in each pixel is improved by means of an individual leakage current compensation and a high energy threshold with 3 bits calibration. The counters are decreased to 13 bits, to obtain approximately the same area sensitivity as Medipix1. Both positive and negative charged inputs are accepted by the readout, so detector materials with large hole trappings such as CdTe can be used. The chipboard can be operated via a standard PC either by a USB serial adapter developed by IEAP or by a data acquisition card connected to an interface circuit called MUROS2 developed by NIKHEF. To operate the system, the software Medisoft 4 [21] was developed, and lately also the software Pixelman [22].

A redesigned version of Medipix2 called Mpix2MXR has also been developed. This chip is improved concerning for example radiation hardness, temperature dependence and leakage current compensations. The counter dynamic range is 11810 counts with overflow control [19].
2.3 BRIEF DESCRIPTION OF MEDIPIX3

Medipix3 will be developed to overcome the problem associated with charge sharing in Medipix2 and to allow simultaneous counting and readout. In paper IX in this thesis, charge summing schemes based on pixel to pixel communication were proposed. In the Medpix3 prototype, a scheme based on summing between four pixels (4PS) is implemented [23]. The prototype is fabricated in 0.13 µm CMOS technology with eight metal layers. The system can work either in charge summing mode or in single pixel mode, with the charge summing turned off. Another feature of Medipix3 is that the readout contains two counters and two thresholds for each pixel. The system can be configured for Simultaneous Read-Write operation, where one counter is read out while the other counts, or for Sequential Read-Write, here two threshold levels are available. The latter case makes imaging with three energy bins possible, as proposed in [24].

The normal working mode of the Medipix3 is Fine Pitch Mode, but one additional feature of the chip is that it can be bump-bonded on detectors with 110 µm pitch, i.e. four times the readout pixel area. This case with a large pixel size is called Spectroscopic Mode, since for each detector pixel 8 thresholds from the four readout pixels will be available for colour imaging. The ability for charge summing is also available in Spectroscopic Mode.

Figure 5: Description of the Medipix2 readout circuit for each pixel from [20].
Figure 6: Description of the Medipix2 readout circuit for each pixel from [23].
3 CHARACTERISATION OF THE MEDIPIX SYSTEM

3.1 CHARGE SHARING IN PIXELATED DETECTORS

An important problem for the design of high resolution pixel detectors is that some of the induced charge will drift to neighbouring pixels and thus contribute in the wrong pixel position. This effect is called charge sharing. If the pixel size of a pixellated detector is small compared to the wafer thickness, charge sharing will be a serious problem and will reduce both spatial resolution and energy resolution of the detector [25].

When an image is taken, small details of the object will always be degenerated. The capability to show small details is referred to as spatial resolution. This is usually measured using the “line spread function” (LSF), which describes the extent to which a sharp edge will be sharp in the image. A “point spread function” can be defined as being analogous to LSF. The mechanisms that can cause the image to degenerate in an X-ray imaging system are:

- Beam geometry and scattering
- Quantisation error
- Absorption width
- X-ray fluorescence
- Charge drift
- Back scattering

Although the resolution is degenerated by charge sharing, it will have an even more drastic influence on the spectroscopic properties of the system. For a photon counting system, the spectroscopic properties of such a system are severely degenerated by the charge sharing phenomenon [1]. In the effective spectrum obtained by the system, high energy photons will be redistributed towards lower energies, as shown further on in this section. This occurs when a high energy photon hits the detector close to the border between two pixels and the charge cloud is distributed over both pixels. The event detected by the system is two photons which share the energy of the high energy photon.

Although several research projects concerning pixel detectors exists, the Medipix2 system is the only system which has reached such small pixel sizes that charge sharing has become a large problem. Due to the high resolution and the low noise of the Medipix system, it can be used as a tool to study the charge sharing phenomenon.

3.1.1 Beam geometry and scattering

Divergence of the X-ray beam and scattering of X-rays in air or in soft tissue will cause the image to degenerate. This problem does not depend on the detector technology, but to some extent X-ray scattering in soft tissue can be compensated for when using “colour” X-ray imaging.
3.1.2 Signal sharing from quantisation error

The quantisation error present in pixellated X-ray detectors exists in all digital imaging systems. Since the image is digitalized a diagonal line will be represented by a “stairway” where the pixel size defines the size of the steps as in Figure 7. The maximum width of the line spread function can therefore not be smaller than the pixel size [26]. This is the driving force behind the development of detectors with reducing pixel sizes.

![Column profile](image)

Figure 7: The line spread function, LSF, can be measured with an illuminated slit with smaller dimension than the pixels. In the illustration, the intensity for a pixel in a column will correspond to the LSF-value for the corresponding distance between the slit centre and pixel centre. The proportion between the plateau and the slope of the LSF depends on the width of the slit.

3.1.3 Signal sharing from X-ray absorption width

When an X-ray photon is absorbed in a semiconductor, a chain of events is initiated where the total energy is deposited. This process can involve locations in more than one pixel. In paper V and [10] different aspects of how the distribution of deposited energy can be calculated by simulations using the Monte Carlo simulator MCNP [27] are studied.

The cross section, which is the probability for absorption of X-ray photons in the detector, differs depending on the photon energy and on the detector material. Silicon is quite transparent and only about 10 % of the photons from a standard dental X-ray source are absorbed in a 300 µm silicon detector. Materials such as GaAs and CdTe are much better X-ray absorbers and can therefore offer increased quantum efficiency, as can be seen in Figure 8. But the superiority in electrical properties for silicon makes it an attractive detector material even though its absorption properties are poor.
3.1.4 Signal sharing from X-ray fluorescence

X-ray fluorescence is a process related to X-ray absorption. It occurs when an atom emits the absorbed energy as a new X-ray photon. The energy of the generated X-ray photon typically corresponds to the binding energy of a K-shell electron. The energy of fluorescent photons increases with atomic number. Fluorescent X-rays are emitted in any direction and can be absorbed again “far away” from the original absorbing position in the detector. If the effect of X-ray fluorescence is large, as it is for heavy atoms, it will cause serious degeneration of the image quality. When a CdTe-detector is irradiated using a 40 keV beam, approximately 43% of the beam energy will be carried away by fluorescence. X-ray fluorescence for CdTe and GaAs is quantified in paper V. Although the effect of fluorescence is significant in CdTe and GaAs, the effect of charge drift is larger.

Table 1: Energies for fluorescence X-rays for some detector materials together with the escape energy if a 40 keV source is used.

<table>
<thead>
<tr>
<th>Material</th>
<th>Kα-energy (keV)</th>
<th>Escape energy for a 40 keV source (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cd</td>
<td>23.17</td>
<td>16.83</td>
</tr>
<tr>
<td>Te</td>
<td>27.47</td>
<td>12.53</td>
</tr>
<tr>
<td>Ga</td>
<td>9.25</td>
<td>30.75</td>
</tr>
<tr>
<td>As</td>
<td>10.54</td>
<td>29.46</td>
</tr>
</tbody>
</table>
3.1.5 Signal sharing from charge drift

Charge drift is the single most important contribution to signal sharing, and therefore signal sharing is often referred to as charge sharing. The energy deposited by X-ray photons in a semiconductor generates a cloud of free electron-hole pairs.

![Illustration of charge sharing](image)

Figure 9: Illustration of charge sharing. In silicon the absorption path can be approximated by a uniform line through the detector. A normal wafer thickness is 300 µm so the charge has to travel 150 µm on average. In CdTe the absorption can be approximated as a point on the surface. The charge has to travel through the whole wafer which normally is 1 mm. In both cases the applied field is about 350 V/mm.

Normally a field is applied to force the induced charges to accelerate towards the contact for the readout electronics. But during the transport time the charge cloud will increase in size due to diffusion as illustrated in Figure 9. Over-depletion will suppress the diffusion and therefore to some extent improve the situation.

3.1.6 Signal sharing due to backscattering of X-ray photons

Backscattering of X-ray photons from underlying layers is a problem in the case of detector materials with low absorption such as silicon. In the Medipix system the detector chip is bump-bonded to the readout chip using an indium or tin/lead bump for each pixel. The size of the bump is comparable to the pixel size and it will reflect a significant amount of X-ray photons thus contributing to the charge sharing [28].

3.2 Observations of charge sharing in slit measurements

The most straightforward means of visualising that charge sharing is present in Medipix2 is to investigate the spatial resolution with slit measurements. To measure the line spread function, a 10 µm wide and 6 mm long slit was used. Since
the slit is tilted, the line spread function can be achieved from the signal distribution in columns [26], as discussed in chapter 3.1.2.

Figure 10: Experimental setup. The 10 µm slit is exposed by 70 keV X-rays.

The distance between the slit and the detector surface was approximately 2 mm, since the slit could damage the bonding wires between the readout chip and the circuit board if they had physical contact. The width of the focal spot point of the source is 100 µm. The divergence of the beam is small, since the distance of 350 mm between the slit and the source is much longer than that between the slit and the detector. The spot size of the X-ray beam on the detector after the slit can be estimated to 11 µm due to the width of the focal spot point.

Figure 11 displays an image of the slit obtained using a Medipix2 silicon detector and a wide energy threshold window. In this image the pixels with a centred slit can be clearly distinguished from the shared areas.

Figure 11: Image of a slightly tilted slit achieved using Medipix 2.
Figure 12: Row profiles of the slit in Figure 11.

The achieved row profile in Figure 12 is due to the geometrical position of the slit [26] and charge sharing properties [1] of the detector. The left and right solid lines correspond to rows where the centre of the slit is placed exactly in front of the centre of pixel columns 139 and 140 respectively. The signal achieved by the neighbouring pixels is due to charge sharing only. The signal due to charge sharing does not exceed 10%. The dashed line shows a row where the centre of the slit lies exactly in between pixel columns 139 and 140. The dotted lines correspond to rows where the spot size is close to the border, but does not overlap the neighbouring pixel.

3.2.1 Colour images of a slit

It is possible to compose RGB images of a slit, by combining three images achieved with three different energy windows. The dental X-ray source has a wide spectrum seen in Figure 1. The low energy image is assigned to red (10 – 15 keV), the medium energy to green (25 – 30 keV) and the high energy to blue (43 – 47 keV). If the full spectrum is present the resulting colour will be white. The reason for this rather low RGB scheme is the shift of the recorded spectrum due to charge sharing discussed in chapter 3.3.4.
It can be noted in Figure 13 when the slit is centred the pixel colour is white, which means that the full spectrum range is recorded. The red colour corresponds to pixels not illuminated that are reached by the charge cloud from an illuminated neighbour pixel.

The image of the slit on a CdTe detector in Figure 14 shows a width of approximately 400 µm. The increase in width is due to charge drift and fluorescence as discussed in chapter 3.3.2.

3.2.2 Slit profiles with low threshold discrimination

The column profiles for this slightly tilted slit are more interesting than the row profile, since it corresponds to the LSF [26] and offers information concerning the signal as a function of the position in the pixel. If these column profiles are studied for different low threshold discrimination levels, information about signal sharing properties can be obtained.

The lower curves displayed in Figure 15 show the series of the signal distributions along the columns for the detector with a tilted slit. The dotted curves correspond to a low level of the lower threshold. For the dashed curves, the lower energy counts are discriminated by applying a 22 keV higher value of the lower threshold.
Figure 15: The lower curves are column profiles of two images of the slit obtained using different threshold values. The dotted profiles correspond to a low threshold and the dashed profiles to a high threshold. The peaks of the profiles correspond to pixels where the slit intersects with the centre of the pixel, while the crossing of profiles corresponds to the border between pixels. It can be seen that plots for the high threshold are always narrower than those for the low threshold. This means that discrimination of lower energy counts reduces the influence of the shared signal and therefore improves the resolution. This can be explained as a suppression of charge sharing, where the counts originating from charge clouds shared between two pixels will be discriminated for the high threshold setting. A conclusion that can be drawn is that increasing the lower energy threshold decreases the signal contribution from the shared high energy photons.

Other information can be extracted by summing the columns (upper thick lines in Figure 15). The dotted thick lines correspond to a higher threshold than the solid thick line. If all generated charge is collected, then the signal equal to the sum of the signals over a row for separate columns, should not exhibit any changes as a function of row number. This is the case if the intensity of the X-rays beam is constant along the slit. From Figure 15 it can be seen that the modulation of the sum of the signals occurs in the case, when lower energy photons are excluded. In
the borders the signal drops showing that the collection of charge is different for different thresholds. This is due to charge sharing. The signal originating from shared photons with a low energy is suppressed if the threshold is increased. Therefore the sum over a row decreases for high thresholds, when the slit intersects the gap between two pixel columns. The sum for the lower threshold shows a slightly higher count rate in between the pixels. For low thresholds photons absorbed near a pixel border will be counted twice, since the charge cloud from the photon is divided between the two pixels.

3.2.3 Slit profiles for a 1 mm CdTe detector

In the case of a 1 mm thick CdTe detector, the charge cloud on the readout side of the wafer becomes larger than a pixel for low energies. Row profiles corresponding to the line spread function can be achieved for CdTe in the same manner as was discussed for silicon. The achieved profiles show a width larger than one pixel. For a low energy setting as in Figure 16 an inverse modulation can be seen.

Row profiles for energy window centered at 16 keV

![Row profiles for energy window centered at 16 keV](image)

Figure 16: Column profile for two pixels and their sum for a slightly tilted slit on a CdTe detector. The threshold energy window is 3 keV wide with middle energy value at 16 keV.

In Figure 17 this modulation disappears and the sum of the profiles becomes uniform. For a higher energy window as in Figure 18 the strongest signal comes from the centre of the pixel, indicating that discrimination takes place for shared counts on the border between the pixels. The widths of the profiles are however still wide compared to profiles for a silicon detector.
Figure 17: Column profile for two pixels and their sum for a slightly tilted slit on a CdTe detector. The threshold energy window is 3 keV wide with middle energy value at 22 keV.

Figure 18: Column profile for two pixels and their sum for a slightly tilted slit on a CdTe detector. The threshold energy window is 3 keV wide with middle energy value at 41 keV.
3.3 SIMULATIONS OF DETECTOR PERFORMANCE

Computer simulations are important for gaining understanding about the basic physical phenomena leading to particular experimental results. The ever-increasing computer power leads to simulation models which have higher and higher degrees of complexity. The computer environment for simulations in this thesis is the Electronic Design Computer Centre at Mid Sweden University. The mainframe is a Sun Fire 15K StarCat with 72x - 900 MHz CPUs, 288 Gbyte of memory and 3.93 Tbyte of RAID5 disk space.

The purpose of the simulations described in this chapter is to reproduce the different measurements in this thesis and thereby provide understanding about the charge sharing mechanism in Medipix2 detectors. Simulations of X-ray absorption, X-ray fluorescence and backscattering have been performed using MCNP [27]. Simulations of charge drift have been performed using Medici [29] and GEMS [1]. The latter simulator is based on a full band Monte Carlo method and has been developed to explore the characteristics of novel materials and novel devices [30]. All simulations in the following sections are done using 10 µm wide beam of 40 keV photons, except the flood image simulations in section 3.3.4.

3.3.1 Simulations of X-ray absorption in silicon

When the Medipix2 chip is equipped with a 300 µm thick silicon detector, it can be used for imaging with good spatial resolution, in spite of the charge sharing problem. Simulations of one pixel in 300 µm silicon using MCNP gives a quantum efficiency of 3.4 % and an absorption profile according to Figure 19. The X-ray absorption is approximately uniform through the detector, but scattering events broaden the half width of the beam to about 12 µm.

![Simulation in MCNP showing the absorption profile in 300 µm Si for a 10 µm wide source with monochrome 40 keV X-rays.](image)

Figure 19: Simulation in MCNP showing the absorption profile in 300 µm Si for a 10 µm wide source with monochrome 40 keV X-rays. The plot to the left is a contour plot of a pixel of the detector along the beam (horizontal axis scaled 4 times); the plots in the middle and to the right are profiles along and perpendicular to the beam.
The increase in absorbed energy at the bottom of the pixel is due to X-ray fluorescence from the bump-bonds [28]. This fluorescence is sometimes referred to as “backscattering”. The bump-bonds can consist either of Indium or of a mixture of Tin and Lead. Backscattered X-rays are considered in paper VII, since they will give a contribution of low energy counts in the achieved spectrum [28]. In Figure 20 the energy spectrum due to fluorescence in Lead/Tin and Indium are shown.

![Backscattering spectrum in silicon](image)

**Figure 20:** Spectrum achieved from MCNP due to fluorescence from a bump bond consisting of Indium or Lead/Tin.

The Medipix detectors characterised in this thesis contain bump bonds of Lead/Tin, so this spectrum is more interesting. The $K\alpha$ energy for Tin at 25 keV can be observed. The distinct peak at 12 keV originates from the L-shell in Lead. For sources with higher energy the $K\alpha$ energy for Lead at 75 keV could be expected to occur. The backscattered radiation can be seen as a spherically radiating source at the bottom of the pixel as in Figure 21.

![Detail of the contour plot](image)

**Figure 21:** Detail of the contour plot to the left in Figure 19, showing the absorption profile in a pixel bottom up to 20 µm above the bump bond.
A contribution of backscattering from the silver glue (7 µm thick on the back side of the ASIC) and from the copper ground plate will also be present in the detector [31]. In the case of a narrow beam on one single pixel this fluorescence will be radiated spherically through the ASIC and affect an area of approximately 500 pixels. Therefore backscattering from silver glue has been omitted in the model. It is implied in [32] that the main backscattering contribution comes from the bump bonds.

Medipix chips with 700 µm thick silicon detectors are available. Simulations using MCNP give a quantum efficiency of 7.3 % and an absorption profile according to Figure 22. The X-ray absorption is approximately uniform through the detector, but scattering events broaden the beam to about 13 µm.

![Figure 22: Simulation in MCNP showing the absorption profile in 700 µm Si for a 10 µm wide source with monochrome 40 keV X-rays. The plot to the left is a contour plot of a slice of the detector along the beam (horizontal axis scaled 4 times); the plots to the right are profiles along and perpendicular to the beam.](image)

Due to improved quantum efficiency and almost no increased Compton scattering, the 700 µm thick silicon detector would be a better choice than the 300 µm silicon detector. However, degeneration of the spatial and spectral resolution of thick silicon detectors occurs because of charge sharing. Another complication
discussed in chapter 3.4.2 and in paper VII involves alignment problems in thick detectors.

3.3.2 Simulation of X-ray absorption in CdTe

Almost all direct absorptions of 40 keV photons in a CdTe detector take place close to the detector surface, as can be seen in Figure 23. However, the influence of fluorescence causes significant degeneration of the quantum efficiency of one single pixel. MCNP-simulations show that due to fluorescence only 70% of the incoming X-rays will deposit their energy inside the 55 µm pixel. Some fluorescence X-rays will be emitted upwards, but 22% of the incoming energy from the beam will be deposited in other pixels of the detector and thus the image will be degenerated.

Figure 23: Simulation in MCNP showing the absorption profile in 1 mm CdTe for a 10 µm wide source with monochrome 40 keV X-rays. The plot to the left is a contour plot of a slice of the detector along the beam (horizontal axis scaled 4 times); the plots to the right are profiles along and perpendicular to the beam.
Imaging using a 1 mm thick CdTe detector requires discrimination of energies below 30 keV in order to eliminate image contributions from fluorescence. If only Compton scattering were present, the spatial resolution of CdTe would be similar to silicon, apart from considerably improved quantum efficiency. In paper V simulations using MCNP of the fluorescence spectrum in different detectors were presented. The resulting spectra can be seen in Figure 24. The distinct peaks are escape peaks, or the remaining energy of the incoming photon after the transfer of energy to a core electron of an atom. When the atom is de-exciated the released fluorescent X-ray photon will have a range of some hundred micrometers, so the probability that it is absorbed in the same pixel as the incoming photon is low.

![Figure 24](image)

**Figure 24:** Simulation in MCNP of the resulting spectrum when different detector materials are illuminated with a 10 µm wide beam with monochrome 40 keV X-rays. The escape peaks in the various materials can be clearly seen.

### 3.3.3 Simulations of charge drift for a narrow beam

In paper V and paper VI simulations of charge transport with MEDICI for a CdTe detector were presented. In Figure 25 a typical absorption event is illustrated.

![Figure 25](image)

**Figure 25:** Schematic sketch of a typical X-ray event in a detector of a high-Z material. A fluorescent photon is emitted and captured outside the pixel. The charge cloud from both absorption events widens by diffusion during drift towards the readout electrode.
The simulation shows that the charge cloud of a 40 keV photon will become larger than the pixel size when it reaches the readout electrode, see Figure 26. Without charge summing between pixels it is therefore impossible to detect the full energy of an X-ray photon by means of a 1 mm CdTe detector.

Figure 26: Extension of the charge cloud at the pixel contact for 40 keV photons captured in a 1 mm thick CdTe detector biased at 300 V. Both the constant field case (compensated material) and the linear field case (diode like structure) give the same result for this specific case.

3.3.4 Simulations of charge sharing with GEMS

With a pixel size as small as 55 x 55 μm, charge sharing is a significant effect, as stated in [1]. The simulations show that for the pixel size 50 x 50 μm², measured energy spectra will be highly distorted and the line spread function of an image of a slit will be broadened. Three levels of Monte Carlo simulations for a pixelated detector system are performed. Photon transport is simulated with MCNP. The full band charge transport is simulated using GEMS. The results are interpreted at the system level using a Monte Carlo methodology for a detector array implemented in Matlab. Since the GEMS simulations are very timeconsuming, a large number of absorption events have been pre-calculated. This GEMS-generated database is used as the input to the system level Monte Carlo scheme in order to directly extract charge sharing information from the silicon detector, for a given absorption profile from MCNP. Simulated and measured spectra of an X-ray tube have been compared in paper IX.
Figure 27: Comparison between simulated and measured spectra for the Medipix2 system. The bias voltage of 25 V corresponds to the depletion voltage of the detector. The solid line shows the actual spectrum of the source.

The achieved spectrum shows that at the depletion voltage the Medipix2 system with a 300 µm silicon detector is unable to resolve the spectrum of a dental X-ray source. As seen in Figure 28, the charge sharing is suppressed if the bias voltage is increased. But even at 100 V bias voltage, which is close to the breakdown voltage of the 300 µm silicon detector, the achieved spectrum is far from the original spectrum of the source.
3.4 MEASUREMENTS WITH A NARROW MONOCHROME BEAM

Measurements on Medipix2 using a monochrome source were performed at ESRF in Grenoble, France [33]. The purpose of these measurements was to isolate specific contributions to the charge sharing process for Si- and CdTe-detectors. These measurements are described in papers V and VI for CdTe-detectors and in paper VII for Si-detectors.

The actual beam line delivered an intense and monochrome beam consisting of 40 keV photons. The beam size was approximately 900x900 µm but it was limited to a 10x10 µm square with two crossed slits. The detector was mounted perpendicularly to the beam.

3.4.1 Measurements on a 1 mm thick CdTe detector

The measured point spread function for the beam on a CdTe-detector is shown in Figure 29. When the discriminator level is close to the energy of the source, the signal on the neighbouring pixels is three orders of magnitude lower in intensity. However if the energy threshold falls below the fluorescence energies of CdTe, a strong signal is present on the neighbouring pixels and a significant signal is observable even at a distance of 165 µm from the beam. This agrees with simulations of X-ray absorption and fluorescence using MCNP, as seen in Figure 29.

Figure 28: Comparison between simulated and measured spectra for the Medipix2 system when the detector bias is 100 V.
Figure 29: Measurement of the signal in neighbouring pixels of a CdTe detector for different threshold energies. The bold line indicates the simulated response due to fluorescence, without taking charge transport into account.

The spectrum achieved from the centred pixel offers more information regarding what is actually happening in the detector. In Figure 30 a spectrum achieved using a 2.1 keV wide threshold window is shown. The plateau for lower energies is expected if charge sharing and fluorescence are present, so it is evident that, although the beam is centred, a fraction of the photons will deposit a part of their energy outside the pixel. The neighbouring pixel shows a spectrum without an actual energy peak, with a significant count rate at low energies. The spectrum intensity of the neighbour slowly decreases to zero when the energy becomes close to 40 keV. The photon energy 40 keV cannot be present in the readout from the neighbouring pixel since no photons are absorbed in this pixel. The shape of the spectrum for this neighbouring pixel is typical when the dominant process is charge sharing.
Figure 30: Spectrum achieved from a CdTe detector when the beam is centred on one pixel. The charge sharing spectrum from the neighbouring pixel is also presented. The spectrum is achieved using a 2.1 keV wide threshold window.

A small tail towards higher energies can be seen in Figure 30. In Figure 31 this low energy tail is shown to a level above 90 keV. Note that above the double energy (80 keV) the spectrum intensity drops to almost zero. This indicates that the high energy tail originates from double counts, where two photons hit the pixel within the time resolution of the input amplifier.

Figure 31: Spectrum on a CdTe detector to study the high energy tail originating from double counts.
Measurements of the spectral response when the beam is scanned over the pixel area provide information of the charge drift in the detector. These measurements in paper VI are very noisy, but in Figure 32 it is evident that the simulated size of the charge cloud in Figure 26 is reasonable.

![Figure 32: Position of the energy peak as a function of beam distance form the centre of the pixel. The line shows the simulation results. The dots represent experimental data without correction for incomplete charge collection.](image)

For low energy settings of the lower threshold on the present 1 mm CdTe-detector, the spatial resolution is insufficient. Bright details in an image will be smeared out over a distance of approximately 400 µm, as in Figure 14. Since the charge cloud is as large as the pixel size, the energy information will be highly dependent on the position of photon absorption in the pixel. Therefore no relevant energy information can be recorded using a CdTe detector on Medipix2 under flood illumination. With charge summing schemes for the closest pixels, the charge drift problem can be corrected, but still fluorescence will distort the energy information, unless charge summing schemes for about 7x7 pixels are implemented.

### 3.4.2 Measurements on 300 µm and 700 µm thick Si detectors

In paper VII measurements of a narrow beam on a 300 µm silicon detector are reported. Figure 33 displays a spectrum achieved from a low threshold scan. This generates a cumulative spectrum which can then be differentiated. The spectrum is achieved from a pixel where the narrow monochrome beam is centred. A plateau for lower energies due to charge sharing is present in the differentiated spectrum, since there is a slope in the cumulative spectrum. The simulations in chapter 3.3.1 reveal that this charge sharing is due to backscattering from the bump.
bond. The charge cloud is not sufficiently large to reach neighbouring pixels when the beam is centred in a pixel. The low energy slope in the cumulative spectrum can be reproduced if Indium or Tin/Lead is present below the detector in the MCNP model.

![40 keV synchrotron spectrum on 300 µm Si](image)

**Figure 33:** Spectrum of the 40 keV source achieved from a low threshold scan with a 300 µm thick Si detector.

To be able to see how the charge sharing influences the spectrum, the beam was moved from the centred position towards the border of the pixel. The best spectrum should be achieved with the beam centred above a pixel (0 µm), while the beam position on the border between two pixels (28 µm) should give a minimal contribution of 40 keV counts. The charge drift for this measurement situation has been simulated using GEMS. The simulation results compared to the measurements are presented in Figure 34. To reproduce the slope of the 40 keV edge in the spectra, a readout noise level of 80 electrons had to be added, which corresponds to the expected value. The soft bending at about 38 keV for some positions indicates that a part of the charge cloud has escaped from the pixel. The presence of lower energies is mainly due to backscattering. In paper VII both aligned and 2° tilted beams are studied. The tilt of the beam introduces a beam widening, but the position sensitivity will become quite complex. If the beam is tilted, charges created close to the detector surface will reach the readout on a shifted position compared to charges created close to the readout. Charge clouds created close to the surface have a long drift path and will result in wider clouds reaching the readout. Note that the peak shift for the border position is reproduced in the
simulation if the beam is tilted, which indicates that the simulation accuracy is reasonable.

Figure 34: Comparison of measured and simulated cumulative spectra for a narrow 40 keV beam on a 300 µm thick silicon detector. The beam is tilted 2°.

Widening of the charge cloud due to angular misalignment of the incoming beam on the detector is illustrated in Figure 35. If the actual tilt is 2°, the beam absorption width will be significantly increased by approximately 10 µm for a 300 µm detector and 24 µm for a 700 µm detector. Alignment problems will hence have a large impact on thick silicon detectors.

Figure 35: Illustration of the increase in absorption width due to a slight tilt of the incoming beam.
When these measurements were performed on a 700 µm thick silicon detector, the achieved spectra for a centred beam showed much less distinct energy edges than those for 300 µm. The charge sharing in 700 µm silicon is expected to be stronger than in 300 µm, since the average charge transport path is 200 µm longer. When the measurements are compared with GEMS simulations in Figure 36 the agreement is reasonable.

Centring of a tilted beam is a complex problem according to simulations. For some angles the 7 µm position will show a more distinct energy edge than the centred position, which also appears to be the case in the measurements. The simulated peak shift for the 28 µm position cannot be found in the measurements, which indicates either that the centring of the beam is not correct or that the beam tilt is lower than 2°.

![Figure 36: Comparison of measured and simulated cumulative spectra for a narrow 40 keV beam on a 700 µm thick silicon detector. The beam is tilted 2°.](image)

### 3.5 Design Strategies to Overcome Charge Sharing

#### 3.5.1 The 3D-detector structure

One important conclusion can be drawn when paper I is compared to [1], namely that crosstalk between pixels due to charge transport is suppressed in a 3D structure compared to a traditional planar structure. This is because the electric
field between the n- and p-contacts of the pixel in the 3D-structure forms a potential well forcing the charges to drift to the “correct” pixel. The field is perpendicular to the incoming photons so the drift path of the charges will not be longer than half a pixel diagonal, as seen in Figure 37. In a planar silicon detector the average drift path will be half that of the detector wafer thickness.

A Medipix1 system equipped with a silicon 3D-detector is demonstrated in [34]. It is argued that the charge sharing can be reduced by a factor of ten compared to a similar planar detector design with four times the bias voltage.

Figure 37: Principal difference between charge sharing in: (left) a 2D pixel detector viewed from the side, and (right) a border (tilted 45º) between two 3D pixels viewed from above.

Another advantage of the 3D design is the possibility to increase the quantum efficiency of silicon detectors by constructing thick detectors, theoretically without increasing the charge sharing. For planar detectors without charge sharing correcting schemes, the gain in quantum efficiency for increased thickness is lost since the increased charge sharing reduces the spatial and spectral resolution.

A problem for a thick silicon 3D-detector is that the source alignment perpendicular to the detector will become important, see section 3.4.2. When the detector is thick compared to the pixel pitch, a small misalignment will cause the photon paths to cross several pixels. Then, details of the object will be smeared out over several pixels, which causes degeneration in both the spatial and spectral resolutions.

For highly absorbing materials the ability to construct thick detectors is not so important, but here the gain in charge sharing is larger since the photons are absorbed close to the surface and the charges have to travel through the entire detector wafer, as mentioned in section 3.1.5. Fabrication of GaAs 3D-detectors is demonstrated in [35].
3.5.2 Charge summing pixel to pixel communication

The problem associated with charge sharing in planar pixellated detectors structures can be overcome if the readout electronics can sum the charge of several pixels for simultaneous events. This solution is mentioned in [20] and one possible low level implementation is presented in [36]. In paper IX, different summing schemes were compared. Summing of charges between pixels will unfortunately always provide additional noise. The noise will increase as the square root of the number of contributing pixels. Four different charge assignment schemes were evaluated concerning their energy resolution and increase in readout noise.

![Figure 38: Four pixel summing configuration (4PS) to the left and five pixel summing configuration (5PS) to the right.](image1)

![Figure 39: Three pixel summing configuration (3PS) to the left and seven pixel summing configuration (7PS) to the right. These schemes requires a Odd Column Shifted Matrix (OCSM)](image2)

The readout electronics must firstly decide the pixel to which to assign the position of simultaneous shared events and this will be the position of highest signal value. The energy level is however based on the sum of charges for this pixel and its neighbours. For the non-symmetric configurations (4PS and 3PS) the neighbouring pixels are not defined, therefore an additional logic step is required to decide which set of pixels should be summed. But this will actually increase the covered area of the non-symmetric schemes, since for example a corner hit will be assigned to the most appropriate set of pixels.
Figure 40: Simulated energy spectra for different charge assignment schemes assuming a mono-energetic 30 keV X-ray source. The assumed electronic noise level is 100 electrons.

The charge summing scheme to be implemented in the Medipix3 is suggested to be (4PS) [23]. This charge sharing correction will solve much of the problem associated with charge sharing due to charge drift. It can however not compensate for all of the charge sharing due to fluorescence that occurs in for example a CdTe detector, see chapter 3.4.1. To solve this problem, a correcting scheme for 7x7 pixels should be required, which will probably introduce too much noise, and might also be difficult to implement with a 0.13 process on the actual pixel size. But CdTe detectors with 110 µm or 165 µm pixels might be an alternative to achieve energy information utilising the better quantum efficiency of the CdTe detector.

3.6 IMAGE CORRECTION FOR THE MEDIPIX SYSTEM

Before an X-ray imaging system can be used, the system must be calibrated to compensate for non-uniformities in the individual detector and in the readout chip. These calibrations must consider the actual application of the system, i.e. the energy and intensity of the X-ray tube. The most well known calibration is “flat field correction”, which must be done for both integrating and photon counting systems. Most readout systems are also designed with a compensation for leakage current. For photon counting system some consideration must be given as to how the threshold settings that discriminates unwanted energies are to be calibrated.

3.6.1 Flat field correction

All detector systems will contain imperfections either in the readout or in the detector which causes different quantum efficiencies for different pixels. This is referred to as fixed pattern noise and is observed when the detector is irradiated by
a uniform X-ray beam. From the signal $S_n$ for each pixel in this flat field image a gain factor $G_n$ can be achieved as follows:

$$G_n = \frac{1}{S_n} \sum_{n=1}^{N} S_n$$

(1)

The number of pixels on the detector is $N$. The flat field correction is normalised using the average number of counts on the detector to preserve the achieved count rate of the image. At full depletion a fluctuation pattern due to radial doping non-uniformities can be observed in most detectors [37]. This effect is considerably reduced by over-depletion. The gain map is thus valid for the specific bias voltage applied to the detector.

Normally a large number of flat field images are averaged in order to reduce the statistic noise in the gain map [14]. Another detail for increasing the accuracy of the gain map is to use the full dynamic range of the counters. For a silicon detector the histogram resolution is narrow and flat field images can be achieved close to the maximum number of counts. However, for CdTe-detectors the non-uniformities of the detector are much larger than for Si and the histogram resolution is therefore very wide. The achievable flat field image is then significantly below that of the maximum number of counts. Gain factors corresponding to counter readings of the maximum number of counts will not produce a properly corrected image. In the first version of the Medipix2 chip, the counters will restart from zero once they have reached the maximum number of counts (8192). A small number of “over-counted” pixels in the gain map can thus cause considerably degeneration to the image quality.

Something that must be taken into consideration is that the value of the gain factor in flat field correction depends on the energy of the incoming photon [38]. If the number of photons of a specific energy is $\Phi_e$, then the number of counts $S_e$ for a detector thickness $t$ will be described by:

$$S_e = \Phi_e (1 - e^{-\alpha_e t})$$

(2)

Since the linear attenuation coefficient $\alpha_e$ is strongly energy dependent, it can be concluded that the flat field correction only works properly for a limited photon energy interval. This is a problem when images taken using a wide spectrum source require correction. In the object, radiation hardening of the source spectrum takes place, meaning that low energy photons are more likely to be absorbed in the object. So the average photon energy is higher in the low intensity parts of the image than in the bright parts.

A flat field correction method, which compensates for radiation hardening in tomography applications, is proposed in [39]. The method uses a set of aluminium test samples of varying thicknesses. The measured transmission $f(x)$ can be assumed to correspond to the sample thickness according to:

$$f(x) = O + A \cdot e^{G \cdot x}$$

(3)
The set of measurements is fitted to this equation for each individual pixel, and a flat field correction consisting of three matrices is achieved. The achieved gain map is valid for objects consisting of materials with transmission spectra equal to the material of the test samples.

3.6.2 Threshold calibration

The applied threshold voltage in a pixellated system will not correspond to exactly the same discrimination energy in each pixel. The Medipix system is designed with a three-bit threshold adjustment and a test pulse intended to calibrate the threshold for each pixel. The test pulse charges a capacitor at each pixel to a specified voltage and then discharges the capacitance into the corresponding readout. It is then possible to scan the threshold for each calibration setting and observe at which setting discrimination of the capacitive charges occurs for each pixel. In this way a uniform threshold setting is achieved for the whole readout matrix. This procedure is referred to as the ANIN test [18]. There are problems associated with equalizing the threshold settings of the matrix with the ANIN test. It is not possible to construct a chip with exactly the same capacitance in the test capacitor of each pixel. Another method used to equalize the threshold settings of the chip is to equalize the edge of the noise floor. This method, supported by the software Medisoft 4.1 and Pixelman, is at present the standard method used in the equalization of the threshold for Medipix2.

The equalisation described above only takes into account the electronics and not the conversion layer and the bump bonds. It is possible to generate an “absolute” calibration mask that considers the whole system by using a source with a known narrow spectrum [40].

If the energy window capability of the Medipix2 system is going to be utilized, the high energy threshold must also be calibrated. Otherwise the fluctuation in the high energy threshold values will introduce large variations in count rates for different pixels for narrow energy windows. The high threshold mask can be generated by taking images of a wide spectrum X-ray source while the energy window is decreased [41]. When the high threshold for an individual pixel falls below the low threshold, this high threshold will be switched off for this pixel and the count rate will increase significantly. It is also possible to use the ANIN test signal instead of a source. This way to achieve a calibration mask for the high threshold is implemented in pixelman.

3.6.3 Proposed method for energy dependent threshold calibration

A problem considered in paper II is that the threshold setting for individual pixels does not vary linearly with energy. Therefore the threshold equalisation is energy dependent. The behaviour of the cumulative function for individual pixels varies within a wide range for different thresholds. Therefore, it is impossible to create a single calibration mask for all energies. The mask is valid for only a narrow energy range due to the variation in the individual cumulative functions.
This range can be estimated from analysis of the cumulative functions for the pixels. The noise floor corresponds to the lowest energy that is possible to resolve with the system. Therefore the quality of the equalization, achieved from the noise floor, degenerates for threshold settings at high energies.

To solve this problem a method of generating threshold masks for different energies was proposed in paper II. In the study, the Medipix1 system and a 70 keV dental X-ray source with wide spectrum were used. The produced masks will be “absolute”, since both detector and channel stationary influences are included, and it is not necessary to use a special calibration source for the creation of correction masks, which simplifies the calibration. For mask creation, histograms of image series were obtained for different thresholds. These histograms correspond to threshold distributions for particular energy thresholds. The model for the histograms was created assuming our standard dental X-ray source. To find suitable parameters, such as the threshold range, it is possible to vary the FWHM (full width half maximum) in the model, until the histograms correspond to experimental data. When creating a mask, it is possible to use the principle of a “flat” correction, since the number of counts from individual pixels is proportional to an integral, which depends on the threshold. The spectral non-uniformity of the detector can be compensated for by the same procedure, if the spectral content of X-rays is similar for each pixel. This is valid, when the number of photons per pixel is sufficiently large (32000). As a result, the image is divided into a number of separate arrays with different count rates and this procedure can be executed for each energy level requiring investigation. Based on these arrays, mask files narrowing the threshold distribution to close to the theoretical limit, were prepared.

This approach can be called a spectral “flat” correction principle, assuming that the threshold distribution dominates the image noise. The histogram shows the distribution of count rates for the pixels of the detector. For the Medipix1 detector and photon energies from 30 - 70 keV, the effective spectrum \( \tau(E) \) is close to the source spectrum \( S(E) \). The number of pixels in the histogram, \( h \), and the number of detected photons, \( g \), are functions of the threshold of the pixels and can be expressed as \( h = f(g) \) or:

\[
h(th) = \frac{R(th - th_0)}{S(th)} = \frac{\int_{th}^{th_{max}} S(E)dE}{\Delta_s} = f(g(th))
\]

Figure 41 shows a graphical representation of this function. The function \( R \) can be approximated by a Gaussian distribution which has a half width \( \Delta_R \). When \( \Delta_R/\Delta_s \ll 1 \), the left part of Eq. (4) represents the threshold distribution. The value is normalised with the factor \( \Delta_s \).
Figure 41: Qualitative description of how histograms relate to the spectra threshold distributions. 1- effective source spectrum from the wide spectrum X-ray source, 2 and 3- pixels threshold distributions for two applied thresholds, 4- cumulative function for pixels, 5 and 6- histograms of images from Medipix1 for threshold distribution 2 and 3, respectively.

Based on this model histograms were created. The model assumes a Gaussian distribution of $R$ and a standard X-ray dental source with a 70 keV spectrum. Varying the FWHM of the threshold distribution, the theoretical histograms in Figure 42a were similar to those of the experimentally obtained histograms in Figure 42b. The images were taken using Medipix1 in the threshold scan regime.

Each histogram in Figure 42 represents an image for a particular threshold voltage. From this data the best fit to the experiment was found when the ratio $\Delta_R/\Delta_S \approx 1/7$. The threshold distribution found is $\Delta_R \approx 0.06 \text{ V}$, assuming that the source distribution is $\Delta_S \approx 0.4 \text{ V}$. Histograms for thresholds closer to the ends of the spectra do not fit the histograms given by the model because of noise. For low thresholds, high noise exists because of detector and amplifier limitations, for higher thresholds on the other hand the signal is low, and the statistical noise increases.
Figure 42: Model – (a) and experimental histograms of flat field images – (b) for different thresholds voltages.

After achieving the range of the count rate distribution from the histogram in Figure 42, it is possible to split the image into the number of levels in the system correction mask. The transformed image contains seven layers describing the resulting mask. After applying this mask the width of the distribution of photo counts was reduced by a factor of 3 compared to the width without a mask. The narrowing of the distribution can be seen on the histograms of these images in Figure 43. A comparison of the achieved histogram of the image after correction can also be made with a theoretical histogram for a perfect seven-level calibration describing the limits of such a correction with this source and system. It is possible to create an optimal compensation mask for each case, because the shape of the
histogram of the corrected image depends on parameters such as the number of levels in the calibration mask, the noise to signal ration, etc.

![Histograms of images](image)

Figure 43: Histograms of image taken without (1) and with (2) mask, compared to a histogram for the calculated image with mask (3).

Using the parameters of the Medipix1 system with a dental X-ray source, the achieved spectral resolution can be estimated. The total range of the threshold energy adjustment for Medipix1 with a dental X-ray source is of the order of 40 keV, which may result in a contrast change of the order of 80%. In this case changing the contrast of an image by for example 1% could be achieved by an energy shift of the order of 0.5 keV. Using these estimations, the limits of resolution can be found because of the minimal contrast difference, which can be detected for two images or objects. Such a minimum can be found, when two images are taken using the same conditions on two separate occasions. The resulting minimal contrast is about 2%, which corresponds to 1 keV. The corresponding standard resolution for 3 sigma is ≈ 3 keV, which is close to the limit of spectral resolution for this system due to Poisson noise, when the number of photons is of the order of 10^4 per pixel. This case corresponds to the highest spatial frequency for the image. The contrast is expected to increase if the spatial frequency decreases. Note that the minimal contrast deteriorates as the threshold rises.
4 POSSIBILITIES FOR PHOTON COUNTING APPLICATIONS

The photon counting principle opens up new perspectives for X-ray imaging. Since photon counting systems are equipped with an energy threshold, the principle is sometimes referred to as “colour X-ray imaging”. The difference in absorption for different “colours” can be used to discern materials in the object. This feature is useful in medical imaging. Today some diseases such as brittle-bone disease are diagnosed by the subtraction of two images taken using two different sources with two narrow energies. This method is called “K-edge subtraction” since it uses the fact that the absorption of photons differs above and below the K-edge energy of a specific atom. For a photon counting system with two energy bins this examination can be done using only one X-ray exposure. In paper VIII it was demonstrated that identification of K-edge energies is possible with Medipix2, if charge sharing is compensated for.

Materials can also be distinguished although no K-edge energy is present in the actual X-ray spectrum. In section 4.1 a method from paper III based on the energy variation in absorption is proposed. This method can be used to distinguish between aluminium and silicon for electronic industry applications. Such methods can also be of interest for fibre quality control in the paper and pulp industry [42].

One important reason behind the development of photon counting systems is to improve the quality of medical imaging and thus to be able to reduce the dose delivered to patients [43]. The possibility of dose reduction with photon counting systems has been demonstrated for dental imaging [44] and mammography imaging [45]. Further image improvement can be achieved by energy weighting algorithms as described in section 4.4. The X-ray energy influence on the image quality is studied in paper IV and briefly described in chapter 4.3.

4.1 MATERIAL RECOGNITION

A thickness-independent method for material recognition using the ratio of absorption coefficients is proposed in paper III. The different dispersion of absorption for the materials can be expressed through the linear absorption coefficient $\alpha$ and depends on the material and on the energy of the X-ray photons. From absorption data tables the energy dependence of $\alpha$ can be achieved. If the thickness of the object is unknown, a convenient material parameter to identify is $K = \alpha_1 / \alpha_2$, which is equal to the ratio of the logarithms of two measured transmissions $\ln(t_1)/\ln(t_2)$. If a database of $K$ for different materials and energies is created, this method can be used for material recognition independent of the thickness of the materials.

To evaluate the method an object consisting of two 0.5 mm thick slabs of silicon and aluminium were used for X-ray imaging with Medipix1. The X-ray absorption for silicon and aluminium is very similar over the range of the 70 keV
dental source, and differs only for lower energies. Hence the possibility to
distinguish between silicon and aluminium is a relevant test of the capability of the
Medipix1 detector.

![Object consisting of two materials and expected images for different photon energies.](image)

Figure 44: Object consisting of two materials and expected images for different photon energies.

For low energy photons, an image should show an evident difference in the relative contrast between the materials, as shown in Figure 44. But with the Medipix1 system it is not possible to discriminate the high energy counts. The system has one threshold that can only discriminate the low energy counts. The relative contrast for a low threshold value is an average over the spectral width from the applied global threshold to the highest energy achieved.

Series of images of the aluminium and silicon object were taken using different energy thresholds. As expected, it was not possible to distinguish between the materials in a high-energy image. For lower energy threshold values the images were studied using histograms as in Figure 45, and it was possible to resolve two distinct peaks and hence identify the region in the image for each material. The resolution of contrasts found by histograms was close to the limit of the system due to the statistical noise of the signal.
Figure 45: Absorption induced histogram shift. Upper figure: selected known regions. Lower figure: the whole detector.

When the regions were known, it was possible to average the transmission over the regions and calculate the relative contrast between the regions. The relative contrast between the materials for each value of the low threshold was calculated using:

$$\text{Relative contrast} = \frac{f_{Si} - f_{Al}}{f_{Si}}$$  \hspace{1cm} (8)$$

The difference of the average relative contrast between the two materials versus threshold after applying a flat field correction is shown in Figure 46. For the averaging an area of about 100 pixels was used.
Another subject concerning the application of this method is the recognition of materials when the object contains mixed or overlapping materials. The proposed method can be extended to this case, but transmission measurements for a larger number of energies are required.

4.2 WINDOW_THRESHOLDS

One improvement to the design of Medipix2 compared to Medipix1 is that an upper threshold has been introduced. Therefore all energies outside a small energy window can be excluded from the images. Without this window such data can only be achieved by differentiation of two images with small differences in the lower threshold settings. To prepare the data for paper III, differentiation of images taken using Medipix1 were primarily intended to be used. Differential images are expected to give a higher relative contrast between the two materials to be distinguished. But the noise of the differential images increased more than the contrast. Therefore differential data were not used in paper III.

For Medipix2 designed with a threshold energy window, a better signal-to-noise ratio was expected, since the statistical noise is not doubled in the differentiation of images. Some speculations on the subject of comparing signal to noise ratios for systems such as Medipix1 without and Medipix2 with threshold window can be made. The differential method requires two images and

\[
\frac{S}{N}_{\text{Medipix}1} = \frac{S_1 - S_2}{2N_{1,2}}, \text{ where } N_{1,2} \text{ is proportional to } \sqrt{S_{1,2}}, \text{ so that}
\]

\[
\text{Accuracy of relative contrast about 0.001}
\]
\((S / N)_{Medipix1}\) is proportional to \((S_1 - S_2) / 2\sqrt{\overline{S}_{1,2}}\). At the same time for window threshold \((S / N)_{Medipix2} = S / \sqrt{\overline{S}}\), where \(S = S_1 - S_2\). Then the ratio between window threshold and only low threshold systems will be 

\[ R_{2/1} = \frac{(S / N)_{Medipix2}}{(S / N)_{Medipix1}} = 2\sqrt{\overline{S}_{Medipix1} / \overline{S}_{Medipix2}}. \]

The advantage of the window threshold method is obvious. Unfortunately the achieved SNR for images using Medipix2 is much lower than for Medipix1 [46]. This is due to two mechanisms:

1. Medipix2 has a much smaller pixel size than Medipix1 and thus the statistics for each pixel must also be smaller. One pixel on Medipix1 covers approximately 9 pixels on Medipix2. If the sum of the signals for these 9 pixels on Medipix2 is taken then the noise will become 
\[ \sqrt{9} \cdot \overline{S} \] and the ratio between the detectors become 
\[ R_{2/1} = 2\sqrt{\overline{S}_{Medipix1} / \overline{S}_{Medipix2}}. \]

2. Charge sharing which destroys the energy information necessary to achieve a difference in contrast between the two compared materials.

Due to these mechanisms the result achieved with Medipix1 in chapter 4.1, that it is possible to distinguish between aluminium and silicon, is not likely to be achieved with Medipix2.

With the Medipix3 system, it is possible to work in “spectroscopic mode”, with four readout pixels bump bonded to one detector pixel. The detector pixel pitch will then be 110 µm, almost as in the Medipix1 system. If the noises of the counters are negligible, the SNR will be improved as discussed above, and a better material resolution than that what is achieved in chapter 4.1 might be possible.

At present spectroscopic X-ray imaging is merely used in astronomy applications, where the images must contain many energy bins to provide useful information revealing for example the material composition of a star. Due to the low photon flux in astronomy it is more suitable to use crossed strip detectors connected to an external multichannel readout, than to use pixellated detectors.

If less accuracy in spectroscopic resolution is required, then the energy information can be visualised by colour coding or contrast enhancing techniques such as energy weighting discussed in chapter 4.4. This can enable the use of energy resolved X-ray imaging in new types of applications. If three energy bins are available, then for example RGB colour coding can be used to visualise the energy information of the image. The possibility of obtaining three energy channels with two thresholds and two counters as in Medipix3 is demonstrated in [24]. In the next two sections it is demonstrated that contrast enhancement is possible with photon counting systems.
4.3 Energy Dependent Imaging

The possibility of achieving better contrast in dental images by selecting specific X-ray energies has been investigated in paper IV. By applying both the low and high energy thresholds of the Medipix2 chip, only a narrow interval of the spectrum of a 70 keV standard dental source is selected. Images of a lower jaw are taken using a setup similar to a dental X-ray imaging system. A Medipix chip with a 300 µm thick silicon detector is mounted inside the jaw.

![Image of a tooth achieved with Medipix2. Two regions with high respectively low X-ray absorption are defined to study the contrast variation.](image)

To obtain a quantitative measurement of the image quality, one region with high photon absorption and one with low photon absorption were defined for the tooth, see Figure 47. The contrasts for these two regions of interest relative to the part of the image above the tooth were extracted. It is relevant to investigate the contrast as it is dose independent. If the contrast is improved, more noise can be tolerated in the image and thus the dose to the patient can be reduced.

A window of 4 keV was used. This window was used to sweep the available energy spectrum and achieve the energy dependence of the contrasts of the defined regions of the tooth, see Figure 48. It can be concluded that the relative contrast can be improved from 0.59 for the full spectrum to 0.70 when a narrow energy interval centred at 30 keV is applied. A maximum relative contrast near this
energy could be expected, since at this energy, region 1 is still absorbing strongly while region 2 is beginning to become transparent [47].

![Graph showing contrast of regions over energy window](image)

**Figure 48:** Achieved contrast for the two defined regions of the tooth and the relative contrast between these regions for a series of 4 keV wide energy windows. The measurements are compared to the value for a full spectrum.

The main conclusion was that it is possible to improve the contrast in dental images by selecting specific X-ray energies. The relative contrast between the parts of the tooth with the highest and lowest absorptions respectively increases by 18 % when the energy spread of the X-rays is narrowed from the full spectrum to a 4 keV window centred around 30 keV. The result implies that it is possible to reduce the dose delivered to dental imaging patients by improving the energy profile of dental sources, or by applying energy weighting algorithms to improve the image quality.

A simple way to generate a colour X-ray image of the tooth in Figure 47 is to use RGB coding for three energy windows within the source spectrum. The colour image of a tooth in Figure 49 is achieved from the images of Figure 4 in paper IV, where the red, green and blue colours are assigned respectively.
Figure 49: Colour X-ray image of a tooth achieved with Medipix2 with a 300 µm thick silicon detector.

4.4 ENERGY WEIGHTING TO IMPROVE THE IMAGE QUALITY

Energy weighting is a method available for implementing the possible improvements of “colour” imaging into clinical X-ray imaging. From the previous chapter on dental imaging it can be concluded that the small tail of high energy photons from the full source spectrum lowers the relative contrast significantly. In a multi-channel photon counting system, the quality of images can be improved by applying an optimal energy weighting algorithm. From theoretical considerations it is proposed in [48] that the optimal weighting factor for mammography applications is \( w = \frac{1}{E^3} \). It should be noted that a charge integrating system corresponds to choosing \( w = E \) while a photon counting system without energy weighting corresponds to the choice of \( w = 1 \). Therefore a charge integrating readout can be said to give unwanted energy weighting favouring high energy photons. This means that using a photon counting principle as in the Medipix system already introduces a significant improvement compared to the state-of-the-art imaging systems with charge integrating readout.

Introducing the energy weighting algorithm for mammography examinations can increase the SNR by up to a factor of 1.9 or reduce the dose by up to a factor of 2.5 compared to integrating systems [48]. A system with only 3 energy channels can still achieve 90 % of this improvement [49]. The Medipix3 system which contains several channels is therefore very promising for the future work on the reduction of doses to patients.
4.5 **K-EDGE IDENTIFICATION**

It is possible to identify the presence of specific atoms in a test sample, if the atoms have K-edge energy in the range of the X-ray source spectrum. The absorption cross section for this atom will be significantly larger for photons with energy just above the K-edge energy than for photons with slightly lower energy. In paper VIII the attenuation of the source spectrum for some different test samples were studied. In Figure 50 the presence of K-edge energies for Iodine (33 keV), Tin (29 keV) and Gadolinium (50 keV) can be distinguished. The measurement setup is described in Figure 51.

![Graph](image)

**Figure 50:** Inverse attenuation spectra for test samples of Iodine, Tin and Gadolinium. The inverse attenuation spectrum is the reference spectrum divided by the transmission spectrum for the test sample.

Photon counting systems such as Medipix can be used for K-edge identification of materials, if:

1. Charge sharing is corrected for.
2. At least two energy channels are available.

In the measurements in Figure 50, charge sharing was suppressed by using a narrow beam centred on one pixel. The beam was collimated with two crossed 10 µm slits, and each energy reading corresponds to one exposure.

The conclusion of paper VIII is that K-edge identification is possible with systems such as Medipix3 where charge sharing correcting schemes as described in paper IX are implemented. The material content can be visualised by colour coding schemes as briefly discussed in chapter 4.2.
One interesting application for K-edge identification is imaging with contrast agents. A contrast agent is a fluid with high X-ray absorption used to fill a void in an object. An example application is the X-ray evaluation of seed, in which bad seed will have a void inside which will be filled with fluid and become black on the image.

Figure 51: Schematic sketch of the measurement setup used in paper VIII.
5 SUMMARY OF PUBLICATIONS

Papers III, IV and VIII in this thesis relate to chapter 4 and concern the practical applications of photon counting systems. This includes image correction, material recognition and improvements to the image quality in dental imaging. Papers II, VI and VII concern characterisation of the MEDIPIX system described in chapter 3. Paper V describes simulations for different detector materials, while paper I contain simulations of electrical properties in a silicon 3D-detector.

5.1 PAPER I

Simulations to demonstrate that charge sharing properties are better in 3D-detector structures than in planar pixellated detectors.

5.2 PAPER II

A simple method to create threshold calibration masks for specific energy intervals is described. A standard threshold equalisation towards the noise edge will not be valid for high energy threshold settings. For wide spectrum measurements, a set of calibration masks can be extracted to cover the whole energy interval.

5.3 PAPER III

A method to distinguish between different materials utilizing differences in absorption for different X-ray energies is proposed. It is demonstrated that Medipix1 can distinguish between aluminium and silicon, although these materials have smooth (no K-edges) and quite similar absorption spectra in the range 10-100 keV. The method is independent of the thickness of the samples.

5.4 PAPER IV

Colour X-ray imaging are exemplified by studies of dental images with Medipix2. It is concluded the image contrast can be improved by 18 % compared to a full spectrum image.

5.5 PAPER V

Simulation of the spectral response for pixellated detectors made of different materials. The spectral response of a pixel is degraded by effects such as X-ray fluorescence, Compton scattering, charge spreading during charge transport, misalignment of the beam and charge trapping. For high-Z materials and high photon energies X-ray fluorescence becomes an important degrading mechanism.
5.6 **Paper VI**

A 1 mm thick CdTe detector with 55 \( \mu \)m pixels was characterised using a narrow mono-energetic 40 keV beam. X-ray fluorescence generates hits in neighbouring pixels with a range of several pixels. Charge sharing causes severe loss of energy information for imaging applications. To achieve full charge collection for the actual geometry, charge summing over several pixels must be implemented.

5.7 **Paper VII**

Silicon detectors, 300 \( \mu \)m and 700 \( \mu \)m thick, with 55 \( \mu \)m pixels were characterised using a narrow mono-energetic 40 keV beam. Measurements of position-dependent spectra can be reproduced by Monte Carlo simulations of the charge transport. Assumptions for readout noise and backscattering of X-rays from the bump bonds can be verified.

5.8 **Paper VIII**

Attenuation spectra of different test samples were achieved using a wide energy X-ray source. It was demonstrated that K-edge energies revealing the material content could be identified. In the study the charge sharing of one pixel was suppressed by crossed slits.

5.9 **Paper IX**

A comparison is given of different charge summing schemes for use in the design of pixellated readout electronics. A scheme which does not significantly increase the electronic noise can be used to compensate for charge sharing in 50 \( \mu \)m size pixellated detector systems.

5.10 **Author’s Contributions**

The author of this thesis has had the main responsibility for papers II, III, IV, VII and VIII. For paper I the contribution was in supplying model input for the simulations. For papers V and IX the contribution was in supplying measurements. For paper VI the contribution was in the preparation of the readout system in the measurement setup and measurement data analysis.
6 THESIS SUMMARY

The work contained in this thesis deals with photon counting detector systems. Possible improvements in the image quality due to the photon counting principle are discussed in Chapter 4. These improvements could reduce the dose given to patients in dental and mammography applications. Another area discussed is material recognition utilizing spectroscopic imaging, which finds applications in process industry, safety, and quality control. Image correction methods are also studied.

Chapter 3 presents a characterisation of the Medipix2 system aimed towards a better understanding of charge sharing. Charge sharing schemes to be implemented in the Medipix3 system are presented, which are expected to solve the problem associated with charge sharing.

6.1 DOSE REDUCTION IN MEDICAL IMAGING

It is proved in paper IV that it is possible to achieve significant improvements in the image quality of dental examinations by introducing “colour” X-ray imaging systems. Energy weighting is one method used to implement the possible improvements of “colour” imaging into clinical imaging. It is proposed in [48] that the optimal weighting factor for mammography applications is \( w = 1/E^3 \). Introducing the energy weighting algorithm for mammography examinations can reduce the required dose by up to a factor 2.5 compared to integrating systems. It can be noted that in a charge integrating system, which is state-of-the-art at present, the energy weighting factor is \( w = E \). In a photon counting system the energy weighting factor is \( w = 1 \), which is closer to the ideal factor. As photon counting systems like Medipix3 containing several channels are being developed, then energy weighting algorithms can be implemented in these systems. At this point the system might be ready to be transferred from applied research into product development in the medical industry.

6.2 MATERIAL RECOGNITION

Photon counting systems open up new perspectives for X-ray imaging. The difference in absorption for different “colours” can be used to discern materials in the object. Today some diseases are diagnosed from analyses of the material content found by K-edge subtraction. For a photon counting system with two energy bins this examination is possible using only one X-ray exposure. In paper III a method for material recognition using a wide spectra source for situations where no K-edge energy is available is proposed. It is demonstrated that the Medipix1 system with only a low energy threshold can distinguish between silicon and aluminium, although the absorption for these materials is very similar.
over the range of a standard dental source. The method is thickness independent and can be extended to distinguish between regions with mixed or overlapping materials. It is also demonstrated that K-edge identification is possible with Medipix2, if charge sharing is suppressed. This can be used in applications for medical imaging utilizing contrast agents.

6.3 IMAGE CORRECTIONS METHODS

It is shown in section 3.6 that for energy dependent X-ray imaging the validity of the standard method of flat field correction for full spectrum imaging is limited. Interpolation methods for flat field corrections for specific energies must be implemented. The threshold calibration for individual pixels is also energy dependent. In paper II a method for generating threshold masks for different energies is proposed. In the study, the Medipix1 system and a 70 keV dental X-ray source with wide spectrum is used.

6.4 CHARACTERISATION MEASUREMENTS OF MEDIPIX2

Due to the size of the charge cloud, 55 µm is close to the physical limitation of pixel size for a planar 300 µm silicon detector. Photon counting systems can be used for full spectrum imaging without too much loss of spatial resolution due to charge sharing but at the cost of reduced quantum efficiency. But the spectral information that is important for energy weighting and material recognition is strongly distorted due to charge sharing as seen in section 3.3.4. But although the spectral information is partly lost for the pixel size 55 µm, energy dependent imaging can still improve the image quality as shown in paper IV.

For a 700 µm thick detector, the alignment of the detector relative to the source and the object becomes very important, since the angle between the centre and the border along the pixels is only 2.2°. The charge sharing is much stronger in 700 µm Si compared to 300 µm Si; which can be understood by geometrical considerations.

In relation to CdTe, it unfortunately appears that a statement in [25] is true. In this paper it is implied that a pixel size below 55 µm for photon counting systems is impossible for a 1 mm thick CdTe detector due to charge sharing. It is concluded in paper V that the size of the charge cloud is as large as the pixel size when it reaches the readout electrode. However there remains a significant possibility of improving the quantum efficiency if high-Z materials could function as good X-ray detectors. A requirement is that the quality of these materials has to be improved giving better uniformity and reduced charge trapping.

6.5 CONCLUSIONS FROM THE MEXIPIX2 PROJECT

The outcome of the Medipix2 project is an X-ray imaging sensor based on single photon processing which have been tested in a large number of applications. Since the device is noiseless it has a very high dynamic range, with the lower limit defined by the background radiation and the upper limit defined by the speed of
the electronics. Spectroscopic imaging can be done either by sweeping one
threshold and subtracting images or by sweeping both thresholds operating in
window mode. However the spectroscopic information is deteriorated by the
charge sharing. The project has resulted in an enormous increase in the knowledge
of the physics of small pixels. This knowledge will be necessary since the
development towards smaller and smaller pixel sizes will probably continue.

For X-ray colour imaging to become a reality, the limitations to the spectral
resolution due to charge sharing must be solved. It is very promising that charge
summing schemes are to be implemented in Medipix3. This will probably solve the
problem associated with charge sharing induced by charge diffusion.

For practical applications it is sometimes profitable to choose a larger pixel
size, since 55 x 55 µm² is not always required. The Medipix3 design will have the
option to bump-bond detectors with 110 µm pixel pitch. If charge summing is
applied to this larger pixel pitch, then CdTe detectors might work properly in spite
of the large range of fluorescence photons.

If energy weighting algorithms were to be implemented, pixellated readout
systems with at least three energy bins are required. Three bins will also make
colour schemes as proposed in [24] possible. Three bins is also a possibility in the
Medipix3 design.

But detector improvement must also be considered. An alternative approach
to suppressing charge sharing is to use the 3D-detector structure instead of planar
pixellated detectors. The transverse electric field in the 3D-structure forces the
electron to drift to the correct pixel. For silicon it is possible to increase the
quantum efficiency by constructing thick detectors, theoretically without
increasing the charge sharing. For high absorbing materials, the advantage of 3D-
design is in the gain involved in the charge transport path. Although the photons
are absorbed close to the surface, the charges only have to travel half a pixel
diagonally, and can then pass through the whole wafer through the doped contact.

The striving for smaller pixel sizes will continue, independent of the
question as to whether it is the 3D-detector or the charge sharing corrections in the
pixellated readout that will prove to be the most successful. The development in
both these areas will in any case continue. In the longer perspective the 3D-
detector might offer the possibility of avoiding the physical limitation that the
diffusion of the charge cloud introduces into the planar detector, that is the ratio
between the pixel size and the wafer thickness. In the future it might be possible to
reach below 25 µm in resolution for a 1 mm thick 3D-detector.
7 REFERENCES


[16] Medipix webpage (http://medipix.web.cern.ch/MEDIPIX/)


[22] Pixelman webpage (http://www.utef.cvut.cz/medipix/)


[33] ESRF webpage (http://www.esrf.fr/)


