DIXI – a Hybrid Pixel Detector for X-ray Imaging

BY

FREDRIK EDLING
Abstract

Medical X-ray imaging is an important tool in diagnostic radiology. The ionising-radiation dose to the patient is justified by the clinical benefit of the examination. Nonetheless, detectors that operate at even lower doses and provide more information to the radiologist are desired. A hybrid pixel detector has the potential to provide a leap in detector technology as it incorporates a more advanced signal-processing capability than currently used detectors.

The DIXI digital detector is a hybrid pixel detector developed for X-ray imaging. It consists of a readout chip and a semiconductor sensor. The division in two parts makes it possible to optimise each part individually. The detector is divided into square pixels with a size of $270 \times 270 \ \mu m^2$. DIXI has the ability to count single photons and every readout pixel has two embedded counters to allow the acquisition of two images close in time. A discriminator enables the selection of photons with energies above a preset threshold level.

The readout chip Angie has been developed and its performance has been evaluated in terms of noise, threshold variation and capability to perform energy weighted counting. Silicon sensors have been fabricated, and a control system for DIXI has been designed and built. An electroless process for deposition of Ni/Au bumps on the chip and sensor has been optimised as a preparation for the assembly of a complete detector, which is being assembled by flip-chip bonding using anisotropic conductive film.

A simulation library for the DIXI detector has been set up and results on the image quality are reported for different exposures and working conditions. A theoretical model for hybrid pixel detectors based on the cascaded linear-system theory has been developed. The model can be used to investigate and optimise the detector for different detector configurations and operating conditions.

Keywords: hybrid pixel detector, photon-counting, X-ray, imaging, simulation
till Silvia
List of Papers

This thesis is based on the following papers, which are referred to in the text by their Roman numerals.


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## Contents

1 Introduction ............................................................................. 1

2 Medical X-ray imaging ............................................................. 3
   2.1 X-ray interactions with matter and biological effects .......... 3
      2.1.1 Interaction mechanisms ............................................ 3
      2.1.2 Dosimetry ................................................................. 4
      2.1.3 Biological effects of X-rays ........................................ 5
   2.2 Detectors for medical X-ray imaging ................................. 6
      2.2.1 Area versus scanning detectors ................................. 7
      2.2.2 Energy weighting ..................................................... 7
      2.2.3 Detector concepts for dynamic X-ray imaging .......... 8

3 Image quality ........................................................................... 9
   3.1 Modulation transfer function ............................................ 9
   3.2 Noise power spectrum ..................................................... 11
   3.3 Detective quantum efficiency ......................................... 11
   3.4 Noise equivalent quanta .................................................. 12
   3.5 Undersampling ............................................................... 12

4 The hybrid pixel detector for imaging applications ............... 13
   4.1 Sensor ........................................................................... 14
      4.1.1 Sensor materials ..................................................... 14
      4.1.2 Semiconductor sensors .......................................... 17
   4.2 Readout chip ................................................................. 18
   4.3 Noise in the sensor and the readout chip ......................... 19
   4.4 Radiation damage ........................................................ 22
   4.5 Interconnection technologies .......................................... 23
   4.6 Detector module construction ......................................... 24

5 The DIXI detector ................................................................. 27
   5.1 Applications .................................................................. 27
   5.2 Angie – the readout chip ............................................... 27
      5.2.1 Performance of Angie ............................................. 28
   5.3 Fabrication of silicon sensors ........................................ 33
   5.4 Anisotropic conductive film as interconnection technology 34
      5.4.1 Electroless nickel plating ........................................ 34
      5.4.2 The flip-chip process with anisotropic conductive film 37
   5.5 Prototype detector of a silicon sensor wire-bonded to Angie 37
1 Introduction

A little more than a hundred years ago, on the 8th of November 1895, Wilhelm C. Röntgen discovered what he named X-rays. The possibility to image the inside of the body was both thrilling and useful and within a couple of months it had turned into an ubiquitous tool in medicine. Ever since then X-rays have played a very important role in the diagnostics and treatment of patients.

The rapid and impressive development of information technologies in the last decade has had profound influences on the society and so also on diagnostic medical radiology. To begin with, processing of X-ray images has opened up new possibilities to extract a maximum amount of information. What is more, a radiologist has fast access to the images and he/she can with ease consult specialists in other cities or countries for advice. The necessary information, such as journals and X-ray images, is exchanged in a few seconds. More and more hospitals have radiology departments that are purely digital.

A new generation of digital X-ray detectors for image acquisition has appeared in the radiology departments. These detectors have the potential to reduce the radiation dose to the patient, to improve the image quality and to shorten the examination times. An example of such a detector, DIXI, is the subject of this thesis.

This dissertation is outlined as follows. Chapter 2 presents a brief overview of medical X-ray imaging, biological effects of X-ray on the human body, detector types and challenges for new detectors. Figures of merits for evaluating and comparing imaging detectors are treated in chapter 3. Chapter 4 discusses the hybrid pixel detector and the possibilities it has to offer. The DIXI detector is described and discussed in chapter 5, which deals with hardware, simulation and theoretical modelling. Concluding remarks are presented in chapter 6 and finally a summary in Swedish is given.
2 Medical X-ray imaging

Several modalities are used in radiology to image the human body. The focus in this thesis is put on planar projective X-ray imaging. Other ionising modalities are for example computed tomography (CT), positron emission tomography (PET), nuclear medicine and three-dimensional angiography. Non-ionising modalities are for example ultrasound and magnetic resonance imaging (MRI). These will not be discussed here, although it can be noted that the modalities are complementary. For example, PET gives a functional image of the body, while MRI is good at imaging soft tissue.

Planar projective X-ray imaging is what we normally think of as X-ray imaging. The X-ray tube is positioned on one side of the patient, usually above and the imaging plate is positioned underneath the patient. The X-ray photon beam passes through the body and the acquired X-ray image is an inverted map of the attenuation in the body. For example, bone attenuates the X-ray photon flux to a greater extent than soft tissue.

Photons are not only absorbed in the body, but they are also scattered. These scattered photons will, if not removed, decrease the signal-to-noise ratio in the image. The exposure has therefore to be increased, to obtain an equivalent contrast, compared to the case of no scattered photons. Methods used to remove the scattered photons include air-gaps and grids. Common to both is that they nevertheless increase the dose to the patient and it is therefore desirable to find other methods to discriminate the scattered photons.

2.1 X-ray interactions with matter and biological effects

2.1.1 Interaction mechanisms

The photons interact with matter through the photoelectric effect, Compton scattering and pair production. In the photoelectric effect the photon is absorbed by an electron bound inside an atom. The electron is subsequently absorbed in the material. The vacancy in the electron shell is filled with an electron from an outer shell and in this transition a characteristic X-ray is emitted. The cross-section exhibits steps at the points where the different atomic shell energies are located. At energies above a shell energy, the electrons of that shell are no longer available for interactions and the cross-section
subsequently drops.

In Compton scattering the photon interacts with atomic electrons and looses part of its energy and changes direction on its way through matter. Besides Compton scattering resides the two cases of Rayleigh and Thomson scattering. These are classical processes and for the energies of interest for X-ray imaging their influences are very small and can mostly be neglected. The threshold energy for pair production is 1.022 MeV, which is far above the energies used in X-ray imaging.

An X-ray beam is attenuated in intensity when passing through the material. The attenuation of a photon beam is described by the total absorption coefficient, $\mu_{\text{tot}}(E)$, which is the sum of the absorption coefficients of the different interaction mechanisms. If $x$ is the thickness of the material and $I_0$ is the intensity of the incoming photon beam, then the number of photon in the attenuated beam is given by

$$I(x) = I_0 e^{-\mu_{\text{tot}}(E)x} \quad (2.1)$$

The absorption length $\lambda$, or the mean free path, is defined as the absorber thickness for which the photon intensity is reduced by a factor of $1/e$.

$$\lambda = 1/\mu_{\text{tot}} \quad (2.2)$$

The absorption length differs between materials. Dense materials with high electron density have a short $\lambda$, while porous materials with low electron density have long $\lambda$. For soft X-rays the photo-electrical effect is the dominant mechanism, but as the energy is increased Compton scattering takes over. This is shown in Figure 2.1 that shows the attenuation coefficient for silicon. An increase in the atomic number of the matter has the effect of pushing the point, at which Compton scattering supersedes in importance over the photoelectric effect to larger photon energies. A photon can scatter several times on its way through matter and it can also be absorbed in the material after that it has been scattered.

### 2.1.2 Dosimetry

The amount of radiation deposited in matter is the absorbed dose, $D$, which expresses the total energy absorbed by an unit volume of unit mass. It is measured in units of Gray (Gy). The damage inflicted on a living organism depends on the type of radiation. Different radiation types\footnote{Different radiation types are for example: protons, photons, $\alpha$-particles etc.} have different relative biological effectiveness (RBE). The differences come from how the different radiation types deposit their energy. This is accounted for by using different radiation weighting factors, $w_R$, where $R$ defines the radiation type.
For example, $\alpha$-particles has $w_R = 20$, while photons have $w_R = 1$. Moreover, the damage depends on which organ that receives the dose as some are more sensitive than others to the radiation damage. The sensitivity is expressed as a "tissue weighting factor", $w_T$. Combining the two weighting factors together with the average absorbed dose in the organ $T$, $D_{T,R}$, the effective dose, $E$ is given as [2]

$$E = \sum_T w_T \sum_R w_R D_{T,R} \quad (2.3)$$

The unit of the effective dose is Sievert, Sv. The effective dose cannot be measured in practice, as it is the dose delivered inside the human body. Instead the entrance surface dose or the dose-area product (DAP) is measured and converted into an effective dose with the aid of conversion coefficients [3].

### 2.1.3 Biological effects of X-rays

An irradiated organ or tissue will experience both deterministic (cells are killed) and/or stochastic (mutated cells that may lead to cancer of hereditary effects) effects. The damage to the DNA can either be direct through interaction with ionising radiation or from chemical damage following the creation of free radicals through radiolysis of water. [4]

The deterministic effects have a threshold below which the probability of harm is zero, while above the threshold it increases fast to unity and the severity is then proportional to the dose. The threshold for the deterministic effects is a few Gy for single exposures or dose rates of one Gy per year. It is believed
that doses above the threshold kill a sufficient large amount of cells to hinder the function of the tissue or the organ.[5]

The stochastic effects are believed to occur at even the lowest doses. The cancer risk is directly proportional to the absorbed dose in the organs and tissues. There exists no dose level below which the radiation is not harmful, i.e. no threshold. In the case of stochastic effects the irradiated cells are not killed, but instead they are mutated. In some cases this can lead to the development of cancer after a time. The probability of developing cancer increases with the dose, but the severity of the cancer does not depend on the dose [5]. The doses delivered in radiological examinations vary depending on the examination performed. In addition, there are large fluctuations between individual cases and between radiologists.

2.2 Detectors for medical X-ray imaging

The detector that is utilised to register the photons has a profound impact on the image quality. Photographic film gives high spatial resolution, but it has a low detection efficiency and a low dynamic range. Film is therefore combined with scintillating screens to achieve a higher efficiency [6]. A commonly used detector is the phosphor storage plate, which can be reused and in contrast to photographic film does not need any development chemicals. [6, 7]. For dynamic imaging the most frequently used detector type is an image intensifier [6, 7]. In last years digital flat panel detectors have been introduced into the market [7, 8, 9].

Digital detectors can be divided into direct and indirect energy converting detectors. Detectors with indirect conversion use several steps to convert the energy of an absorbed photon into electric charge. An example of an indirect detector is a scintillating screen combined with amorphous silicon, where two steps are involved in the energy conversion, from X-ray to visible light and finally to electric charge. In every step there is a risk of reduced sensitivity, added noise and smearing of the signal. In a direct digital system the conversion is performed at the earliest stage, from X-ray to electric charge.

Digital detectors have advantages compared to conventional analogue detectors as they have a larger dynamic range and a linear response. This allows the contrast and latitude of the image to be adjusted in post-processing. An important benefit is that the number of exposure retakes, due to over- or underexposure of the image, is minimised, which reduces the average dose to the patient. The resolution cannot be influenced in neither of the systems, however, for the digital system the sharpness of the images can be adjusted in post-processing. The radiation dose to the patient can be reduced with the new generation of detectors, because of a better detective efficiency.
The possibilities of image post processing has increased with the advent of direct digital detectors. Examples of post-processing are computer aided diagnosis (CAD), image subtraction, image enhancement and 3-D techniques, as for example 3-D angiography. Considering the different clinical applications it can be noted that mammography benefits from an increased contrast, fluoroscopy from noise suppression and all applications benefit from the increased probability of directly obtaining an image with the correct exposure.

2.2.1 Area versus scanning detectors

In radiology, it is possible to use either an area or a scanning-slit detector to image the object of interest. Both approaches have advantages and disadvantages. For dynamic imaging, such as angiography, there is a need of an area detector that can image the whole area instantaneously. A disadvantage with the area detector is that it requires a grid to reduce the influence of scattered photons, which blur the image. With scanning slit detectors only a collimated line of photons reaches the detector. The slit is scanned over the field of view in order to image the full object. Thus, the detector is not well suited for fast dynamic imaging. Moreover, a scanning system is mechanically more complex than an area detector that does not contain moving parts. However, the scanning-slit detector has an excellent rejection of scattered photons.

2.2.2 Energy weighting

Low-energy photons are attenuated to a larger extent than high-energy photons when passing through matter. The consequence is that low energy photons carry more contrast information [10, 11]. The response of a detector depends on the energy of the absorbed photon. If the photon spectrum is divided into \( N \) energy bins labelled \( E_i \), with \( n_i \) photons in each bin and with an associated weight factor \( w_i(E) \), then the detector response, \( R \), is given by

\[
R = \sum_{i=1}^{N} n_i w_i(E) \tag{2.4}
\]

Studies have been performed on how the information carried by the photons can be optimally used. The conclusion is that the response of the detector should mimic the attenuation in the imaged object. The optimal weighting factors in medical imaging has approximately an \( E^{-3} \) dependence and are at low energies independent of material type and thickness [11, 12, 13]. The energy weighting of a particular system can be expressed as an weighting coefficient, \( \alpha \), and the counter response is proportional to \( E^\alpha \).

In nearly all commercial detectors the principle of charge integration is used. In charge integrating detectors the noise is added to the total deposited
energy. Examples of detectors with charge integration are the charge-coupled device (CCD) and the flat panel detector (FPD). The photons are given a weight proportional to the amount of charge they deposit, thus proportional to their energy, and $\alpha = 1$.

Systems that use the photon counting principle are now appearing on the market and several research prototypes are being built. In a photon counting detector all photons that deposit a charge larger than a predefined threshold give an equal counter response. No information about the energy of the photon is preserved, which gives $\alpha = 0$. Another advantage with photon counting detectors is that noise corresponding to an energy less than the threshold is not registered.

The more efficient use of information in photon counting detectors has in simulations shown to reduce the radiation dose needed to obtain a given image quality compared to charge integrating detectors. An even larger dose reduction is achieved with a detector that has a response proportional to $E^{-3}$.\cite{11, 13}

2.2.3 Detector concepts for dynamic X-ray imaging

Dynamic imaging poses extra demands on the detector. The frame rate has to be sufficiently high to image a dynamic process in the body and no trace of a previous image should be present in a later image, for example lag and ghosting. Two common dynamic applications are angiography and fluoroscopy and the associated dose levels can be very high, up to several Gy for fluoroscopy. The need to reduce the dose levels is thus very important.

The most widely used device today for dynamic X-ray imaging is the image intensifier\cite{7} coupled to a CCD. This system has several drawbacks, such as high geometric distortion, non-uniformity in output brightness, limited dynamic range of the camera, loss of image fidelity from the large number of image transformation steps and instabilities in amplitude, noise, scan geometry and focus. The newly introduced flat-panel detector is capable of capturing images in real-time and providing readout in digital form. The detector exists both as an indirect and as a direct conversion version. The detector consists of a converting layer deposited on top of an active matrix array of thin film transistors (TFT). Some advantages are a high quantum efficiency and that large area detectors can be manufactured. The disadvantages are cross-talk between pixels, the noise level, the use of charge integration instead of photon counting and memory effects due to trapping in the TFT.\cite{14, 15}
The concept of image quality is in medical applications always linked to the clinical efficiency of the image. This implies that a medical imaging system should be developed with a particular medical task in mind.

The detection of an object in a digital system is related to its contrast, the pixel size and the background noise [16]. An often-quoted example is the imaging of needles versus the imaging of beans [17]. It can be seen as a trade-off between the detection of high contrast details with sharp edges and bigger details with low contrast. The two demands are not easily incorporated in the same detector. The problem is less severe for digital systems than for film, since digital systems are linear with a large dynamic range and the contrast may be enhanced in post-processing of the image.

Another aspect is that a smaller pixel size lead to an increased radiation dose to the patient. A high dose is required to obtain the same signal-to-noise ratio as for a large pixel. Smallest possible pixel size is thus not always the best solution.

The evaluation of image quality is most often limited to the physical properties. Observer performance studies, the impact on the patient outcome and the optimal usage in terms of cost-benefit are more seldom reported [18]. The focus in this thesis is on the performance of the detector itself. This detection step lays the foundation for the examination of an X-ray image, recognition and identification [18].

The assessment of image quality of a detector is done not only to provide means of comparing different detectors, but also as an aid to optimise its performance.

### 3.1 Modulation transfer function

A perfect imaging system should image a delta function in the object plane, the input, as a delta function in the image plane, the output. Due to imperfections this is not the case in practise and the delta function is instead smeared out. The intensity distribution in the image plane is given by the point spread function (PSF). The two-dimensional Fourier transform of the point spread function is the optical transfer function (OTF). The OTF consists of a real and an imaginary part: the modulation transfer function (MTF) and the phase
transfer function (PTF). For a shift-invariant detector, such as film, the PTF is zero and only the MTF is of interest. In the case of a digital detector the PTF is most often non-zero. The result is that the spatial resolution depends on the position of the point source. For example, close to a pixel boundary charge sharing may occur, which lowers the spatial resolution. A solution is to consider the average of the OTF over all phases, which is called the expectation modulation transfer function (EMTF) [19]. It is given by the following expression where \( b \) is the pixel pitch and \( a \) is the relative distance to the pixel centre.

\[
EMTF(u) = \langle MTF(u_1; a) \rangle = \frac{1}{b} \int_0^b \frac{|OTF(u_1; a)|}{|OTF(0; a)|} da, \quad 0 \leq u_1 \leq u_N \tag{3.1}
\]

where the Nyquist frequency, \( u_N \), is the maximum observable spatial frequency for a sampling distance \( b \).

\[
u_N = \frac{1}{2b} \tag{3.2}
\]

In the digitisation process the image is modelled as a band-limited function. That is, its Fourier transform (FT) is zero outside a bounded region in the frequency plane corresponding to the spatial bandwidth. If the sampling frequency is lower than twice the bandwidth a phenomenon called aliasing occurs. The frequencies above \( u_N \), the so-called fold-over frequencies, will then be present in the sampled data and influence the spectrum between zero and \( u_N \). The system is called undersampled when the sampling is not fine enough to record all the spatial frequencies without aliasing [19]. The discrete sampling has two implications. The first is replications of the Fourier transform in frequency space, due to the infinite sum used in the transform. The second is the overlapping of the replicated FT segments resulting from aliasing. The replication in itself does not have any detrimental effect on the image, but aliasing might have. Aliasing is avoided by band-limiting the system so that no image content exists above the Nyquist frequency.

The presampling MTF, \( MTF_{pre} \), can be used when the Fourier transform exhibits replication but no aliasing. Otherwise the digital MTF, \( MTF_d \), is used. \( MTF_d \) can be defined either for the transfer of a sinusoid or a delta function. The latter is standard, but the former definition is commonly used [2], especially for analogue systems where the two definitions are equal. Here \( MTF_d \) is defined for the transfer of delta functions. The amount of aliasing in the system is demonstrated by the difference between the \( MTF_{pre} \) and the \( MTF_d \). The problem for undersampled digital systems is that the \( MTF_d \) does not allow for a quantitative comparison of different imaging systems. The choice of utilising \( MTF_{pre} \) or \( MTF_d \) to compare two different systems depends on what is being imaged.
3.2 Noise power spectrum

Prior to the detection of the X-rays the noise is governed by Poisson statistics and is white, i.e. spatially uncorrelated. The detection process may introduce a colouring of the noise; it becomes spatially correlated. A common measure of the noise is the root-mean-square (rms) variance, but it does not provide the spatial correlation of the noise. Noise should instead be specified in terms of its mean value and its covariance matrix. The covariance matrix gives a description on how the average noise power varies with spatial frequency. Analysed as a function of its frequency content it is called the Wiener spectrum or Noise Power Spectrum (NPS) [2].

If the value of the covariance function only depends on the relative distance between the noise components, and not on their absolute locations, then the noise is considered stationary. The noise power spectrum for stationary noise is measured as the average of the square modulus of the Fourier transform at each frequency, properly normalised. It is composed of both the signal, the deterministic part, and the noise, the stochastic part.[19]

The problem of undersampling is worse for the NPS, than for the MTF; nearly all digital systems are usually undersampled as far as the noise is concerned. The problem of aliasing is present also for the noise, but any phase dependence is averaged out by the measuring procedure. The aliasing occurs in the same way as for the MTF and can cause an increase in the NPS values. The noise is considered to be random, and will just add up in the overlapping of the Fourier transform replications [19].

The presampling NPS, NPS prec, is the noise before sampling. In contrast to the MTF, it is not possible to measure the NPS prec, as the input to the system contains the whole frequency range including frequencies above $u_N$ and only the digital NPS, NPS d, can be measured.

3.3 Detective quantum efficiency

The concept of detective quantum efficiency (DQE) is extensively used to give a figure of merit for the degradation of the incident information by the detector. The DQE is not the same as the quantum detection efficiency, which only describes the fraction of the input quanta that contribute to the output of the detector. The DQE describes the transfer of the signal-to-noise ratio (SNR) through the detector and it is spatial-frequency dependent. It can be defined as [20]

$$DQE = \frac{\text{SNR}_{out}^2}{\text{SNR}_{in}^2}$$  \hspace{1cm} (3.3)
Ideally the signal-to-noise ratio (SNR) should not be degraded by the detector, i.e. $DQE = 1$. The DQE can be expressed in terms of the EMTF, the $\text{NPS}_d$, the mean number of hits per cm$^2$ in the image ($\bar{d}$) and the incident quantum flux per cm$^2$ ($q$).

$$DQE(u,v) = \frac{\bar{d}^2 \cdot \text{EMTF}(u,v)^2}{q \cdot \text{NPS}_d(u,v)}$$  \hspace{1cm} (3.4)

### 3.4 Noise equivalent quanta

Ideally, the signal-to-noise ratio of the output image should depend on all of the incident quanta. However, any imperfections in the detector reduces $\text{SNR}_{out}$, so that it corresponds to the SNR of an ideal detector with an input of $q'$ quanta. The number $q'$ is named noise equivalent quanta (NEQ) and is sometimes the preferred unit instead of the DQE.

$$\text{NEQ}(u,v) = q \cdot DQE(u,v)$$  \hspace{1cm} (3.5)

### 3.5 Undersampling

Undersampling gives rise to problems on how to define signal and noise. This is the case since they depend on the input to the system. For example, the response to a delta input is governed by the MTF$_{digital}$, while the response to a more complicated signal, such as a sinusoid, is governed by the MTF$_{pre}$. The EMTF is sometimes used instead of the MTF$_{digital}$, as the latter has a phase dependence. The signal and noise is then defined for the same spatial-frequency spectrum. It is helpful to compute the DQE using both the MTF$_{pre}$ and the EMTF. A lower and an upper bound of the DQE are then obtained, which corresponds to the input of a combination of sinusoids or to a broadband of frequencies. An unavoidable problem is that the frequency space with which the DQE is measured is not necessarily the same as the frequency content in the image of a patient.
4 The hybrid pixel detector for imaging applications

The CCD and the flat panel detector have an electronic circuit that simply integrates the deposited charge and all possible noise currents during a given exposure time. A more advanced signal-processing capability is achievable in a hybrid pixel detector. The detector consists of two physically separated parts: the readout chip and the sensor. The photons which interact with the sensor deposit charge that is collected and transferred to the readout chip for further signal processing. The interconnection between the two parts is done using flip-chip bonding. A cross-section of a hybrid photon detector is shown in Figure 4.1.

![Figure 4.1: The principal parts of a hybrid pixel detector.](image)

The division of the detector in two parts enables a separate optimisation of each part. Furthermore, the whole pixel area in the sensor is sensitive to photon interactions, a fact that increases the quantum efficiency compared to flat panel detectors, in which the transistors occupy valuable space. The availability of more space for the electronic circuit has made it possible to include amplification of the collected charge, individual treatment of charges, discrimination of background noise and the choice of only recording photon
hits with a charge deposition exceeding a preset threshold.

The sensor and the readout chip are divided into pixel cells, which in medical applications usually are quadratic. The pixels in the sensor and the readout chip do not need to have the same form, even though this is normally the case. Nevertheless, the sensor and readout chip pixel pitches must match to ensure the connections. The interconnection of the sensor and the readout chip pixels has, together with the assembly of a large-area detector, turned out to be a challenging fabrication problem. This is mainly true when cost is considered in particular.

A method to avoid interconnections between sensor and readout chip is to deposit the sensor directly onto the readout chip, for example amorphous selenium or lead iodide. Prototype detectors with selenium and PbI₂ as sensor material has been built [21]. More work is, however, needed to develop this method into a commercially viable product.

4.1 Sensor

The conversion of the absorbed X-ray photon into electrical charge takes place in the sensor. The most commonly used sensor in a hybrid pixel detector is a crystalline semiconductor operated as a reverse biased diode. Basic criteria on the sensor are

- The stopping power should be sufficiently high, given a specified X-ray energy spectrum.
- The band gap should be high to enable room temperature operation and to keep the noise as small as possible.
- The energy that is needed to create an electron-hole pair should be low to the number of charges.
- The $\mu \tau$ product (mobility $\times$ lifetime) of the collected charge should be large enough to ensure an efficient and fast charge collection.
- In imaging applications the spatial blurring of the signal should be less than the pixel aperture in order not to blur the image.
- The material should not be degraded by radiation during its operational life-time. Normally some degradation can be acceptable as long as it can be compensated for.
- Low cost, good availability and high yield.

4.1.1 Sensor materials

A great flexibility is in principle allowed in the selection of sensor material. Crystalline silicon is favoured for its high intrinsic spatial resolution, low cost, possibility to manufacture large areas and possibility to create complicated
Figure 4.2: The linear attenuation coefficients for a 500 µm thick sensor made of either Si, GaAs, CZT, HgI₂, PbI₂ or TlBr.

The main disadvantage is the low atomic number, which results in a fast dropping quantum efficiency at energies above 10 keV.

The quantum efficiency can be increased by using a thicker sensor. There are, however, drawbacks with increasing the thickness. First, a thicker sensor needs a higher bias voltage to achieve full depletion. Second, it will have a larger leakage current. Third, the charge collection will be slower. Fourth, an increased ratio of sensor thickness to pixel pitch makes the charge sharing between pixels to be increased and the spatial resolution to be degraded.

The search for semiconductors with higher stopping power in the energy range 10–100 keV has lead to studies of compound materials such as cadmium zinc telluride (CZT), cadmium telluride (CdTe) and gallium arsenide (GaAs). All of these materials have been demonstrated with hybrid pixel detectors. New materials for pixel detectors that are being developed are mercuric iodide (HgI₂), lead iodide (PbI₂) and thallium bromide (TlBr). Basic properties of the materials are compared in Table 4.1. The total linear attenuation coefficient for medical diagnostic X-ray energies is presented in Figure 4.2 for a 500 µm thick sensor of different materials.

The collected charge is proportional to the absorbed energy. Large leakage current and incomplete charge collection, due to trapping and recombination, may spoil this relation. Defects in the sensor material are closely associated with these losses. Trapping is the capture of charge that at a later time is released and leads to a reduction and fluctuation in signal size. Compound semiconductors exhibit more problems with material defects than pure semiconductors from group IV in the periodic system.
Table 4.1: Basic properties of some sensor materials, data from Ref. [22].

<table>
<thead>
<tr>
<th>Material</th>
<th>Z</th>
<th>Band gap (eV)</th>
<th>ε, e-h pair (eV)</th>
<th>Hole mobility (cm²/Vs)</th>
<th>Electron mobility (cm²/Vs)</th>
<th>Hole lifetime (s)</th>
<th>Electron lifetime (s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Si</td>
<td>14</td>
<td>1.12</td>
<td>3.6</td>
<td>450</td>
<td>1500</td>
<td>$10^{-3}$</td>
<td>$&gt;10^{-3}$</td>
</tr>
<tr>
<td>GaAs</td>
<td>31,33</td>
<td>1.43</td>
<td>4.2</td>
<td>400</td>
<td>8500</td>
<td>$10^{-7}$</td>
<td>$10^{-8}$</td>
</tr>
<tr>
<td>CdTe</td>
<td>48,52</td>
<td>1.44</td>
<td>4.43</td>
<td>100</td>
<td>1100</td>
<td>$2 \times 10^{-6}$</td>
<td>$3 \times 10^{-6}$</td>
</tr>
<tr>
<td>CdZnTe</td>
<td>48,30,52</td>
<td>1.57</td>
<td>4.64</td>
<td>120</td>
<td>1000</td>
<td>$1 \times 10^{-6}$</td>
<td>$3 \times 10^{-6}$</td>
</tr>
<tr>
<td>HgI₂</td>
<td>80,53</td>
<td>2.15</td>
<td>4.2</td>
<td>4</td>
<td>100</td>
<td>$1 \times 10^{-5}$</td>
<td>$3 \times 10^{-6}$</td>
</tr>
<tr>
<td>PbI₂</td>
<td>82,53</td>
<td>2.32</td>
<td>4.9</td>
<td>2</td>
<td>8</td>
<td>$3 \times 10^{-7}$</td>
<td>$10^{-6}$</td>
</tr>
<tr>
<td>TlBr</td>
<td>81,35</td>
<td>2.68</td>
<td>6.5</td>
<td>4</td>
<td>30</td>
<td>$4 \times 10^{-5}$</td>
<td>$2 \times 10^{-6}$</td>
</tr>
</tbody>
</table>

GaAs has been studied intensively, but unsolved problems remain concerning the manufacturing of large-area sensors and incomplete charge collection. Several prototype GaAs detectors have been constructed for X-ray imaging, [23, 24]. Epitaxial GaAs improves the crystal quality having less defects, but the achievable thickness is limited. Recent work has however shown that it is possible to grow four inch wafers of GaAs with thicknesses of several hundred µm. The material still suffers from doping impurities that puts a limit on the achievable width of the depleted zone [25].

CdTe and CZT are commercially available from several manufacturers. The sensor size is limited to about 2 cm² for good mono-crystalline CZT. Imaging pixel detectors with CZT and CdTe sensors have been reported by several groups [26, 27, 28, 29]. These sensors need further development to increase the wafer size, to reduce the charge loss in the inter-pixel gaps and to minimise the charge sharing between neighbouring pixels [30]. CZT has compared to CdTe a smaller number of dislocations, a higher resistivity and smaller polarisation effects [22]. CZT has therefore in recent years been more widely used as sensor than CdTe. The hole mobility is much lower than the electron mobility in most compound semiconductors, for example CZT. Furthermore, the hole lifetime of CZT is low, due to charge trapping, and holes may recombine before they are collected; the total deposited hole charge will not be collected at the electrode. Consequently, it is necessary to use electron collection for CZT.

Very large stopping powers are provided by TlBr, HgI₂ and PbI₂, but these materials are still not mature. They suffer from trapping, polarisation effects and non-uniformities. A HgI₂ imaging detector with $3 \times 3$ pixels have recently been reported in Ref. [31].
4.1.2 Semiconductor sensors

The active area in a pixel sensor is divided into pixels by electrode structures processed on the surface. The structure is either a highly doped implant with a metal contact or a metal contact directly processed on the semiconductor bulk. The backside has one electrode that extends over the whole surface. Low doped, high resistivity, materials are used for the sensors in order to keep the leakage current low. The semiconductor pixel sensor is based on the diode junction. If the sensor is reverse-biased a large depletion region is created. This region is completely void of charge carriers and should ideally fill the whole detector volume. Figure 4.3 shows an example of a diode. In this case it is a silicon pixel sensor with a p-n junction.

Around the pixel matrix, a guard ring structure for control of the potential distribution is processed to sink currents generated outside of the active area, prevent avalanche breakdowns and to improve the long-term stability. The guard rings are implemented as one or several ring structures around the pixel matrix.

When a photon enters the semiconductor and interacts through the processes described in section 2.1 electron-hole (e-h) pairs are created. The average energy, $\varepsilon$, needed to create an e-h pair depends on the semiconductor material, as listed in Table 4.1. The number, $N$, of created electron-hole pairs is proportional to the absorbed energy in the sensor, $E$, and inversely proportional to $\varepsilon$.

$$N = \frac{E}{\varepsilon}$$  \hspace{1cm} (4.1)
The number of created e-h pairs has a variance, $\sigma_N^2$, that depends on the Fano factor, which describes fundamental processes in the sensor [32]. Silicon has, for example, a Fano factor of 0.115 [33].

$$\sigma_N^2 = F \frac{E}{\varepsilon}$$

(4.2)

The created charge cloud drifts towards the electrodes because of the applied bias voltage. A charge pulse is induced at the contact electrodes from the very moment of charge creation due to the coupling between the charge and the electrode which creates induced mirror charges. The magnitude of the induced charge can be calculated using the Shockley-Ramo theorem [34, 35]. Typical charge collection times in silicon are in the order of 10-20 ns for both electron and holes.

The diffusion of the charge cloud can pose a limit on the spatial resolution, especially at pixel borders. A faster charge collection time, through the use of a larger applied bias, reduces the spreading, since the diffusion takes place during a shorter time period.

### 4.2 Readout chip

The charge created by the interaction of X-ray photons in the sensor is very small and has to be amplified in a low-noise circuit before any further signal processing. A single-photon-counting readout chip should be able to detect the photon without an external trigger. The trigger function is introduced in the readout chip by a discriminator, which has a threshold to select pulses of a minimum size. The hit information is stored in a counter until read out.

The signal induced on the electrodes of the sensor is transferred to the readout chip, where it is integrated in a charge sensitive amplifier (CSA), from now on referenced as the preamplifier. It consists of an inverting amplifier that ideally gives a voltage output that is directly proportional to the input. The feedback resistor is placed in parallel with the integrating capacitance, $C_f$, to remove the accumulated charge, since otherwise the amplifier would soon saturate. The voltage at the output, $V_{out}$, is given by

$$V_{out} = -\frac{Q_i}{C_f}$$

(4.3)

The input impedance at low frequencies is for large amplification dominated by a capacitance.

$$C_{eff} = (A + 1)C_f + C_i$$

(4.4)

where $A$ is the gain of the amplifier and $C_i$ is the capacitive load at the input.
that is dominated by the gate capacitance of the input transistor. It is important to keep the effective impedance, $C_{\text{eff}}$, larger than the capacitance of the sensor to ensure that all collected charge is transferred to the readout chip.

The preamplifier output signal is in many cases amplified and shaped in a subsequent stage, called the shaper. In its simplest form it is constructed as a RC-CR filter that shapes the signal into a semi-gaussian pulse form. The shaper optimises the signal-to-noise ratio and band-limits the signal to remove low-frequency noise. The preamplifier and the shaper are often referred to as the front-end of the readout chip.

In single X-ray photon imaging the information recorded in most chip designs is the number of hits per pixel. A threshold determines the charge size needed to trigger the counter. Recently, attention has been paid to the inclusion of two discriminators with individual thresholds to bin photons into energy windows [36, 37]. The motivation is to extract more information from the detected photon, in order to optimise the image quality and the dose efficiency. To enable energy weighting of the incoming information requires the use of several discriminators and counters, which is similar to the inclusion of a simple analogue-to-digital converter inside every pixel cell [38].

The readout chips are fabricated in state-of-the-art industrial CMOS processes. The transistors and line widths are continuously shrinking in size, while new processes embark on the market. This allows more logic to be introduced into a pixel or the use of a smaller pixel size. Furthermore, the power consumption is reduced when lower supply voltages can be used. A drawback of the development is the cost, which has increased with the reduction in line-width. Not only the processing itself, but the design tools have increased in cost. The new processes are primarily developed for digital electronics and may not be well suited for pixel detectors that include a lot of analogue circuitry.

4.3 Noise in the sensor and the readout chip

Information lost in the detector cannot be restored later, therefore the noise of the detector has to be minimised. It is useful to express the noise sources as an equivalent noise charge (ENC) at the preamplifier input. It is the charge that is to be injected at the input transistor of the readout chip to produce an output voltage which is equal to the root-mean-square (rms) value of the noise. The total noise value is the square root of the sum of the noise contributions squared.

Noise is generated by different sources in the front-end and the sensor. The noise can either be a current or a voltage and they are often referred to as

\[\text{CMOS} = \text{Complementary Metal Oxide Semiconductor}\]
parallel and serial noise. The parallel noise has two sources.

Thermal excitations lead to random fluctuations in the drift velocity. This thermal, or Johnson, noise is generated in the bias resistor of the sensor and in the feedback resistor of the charge sensitive amplifier.

\[ ENC_{br} = \frac{e}{q} \sqrt{\frac{T_p k T}{2R_p}} \]  

(4.5)

where \( q \) is the electron charge, \( k \) is Boltzmann’s constant, \( T \) is the temperature in Kelvin, \( T_p \) is the peaking time of the shaper and \( R_p \) is the parallel resistance of the sensor bias resistor and the preamplifier feedback resistor.

Another parallel noise source, the shot-noise, is created by currents moving across the potential barrier in the sensor diode. Its name derives from the fact that the charge carriers break the barrier in impulses and not smoothly. Another source for shot noise is the charge trapping in the sensor. The charge is trapped and then released after a time delay.

The frequency spectrum of the parallel noise sources is originally white, but as the current is integrated over the capacitance of the sensor, it gets a frequency dependence that peaks at low frequencies. This peak is, for a properly designed front-end, positioned below the frequency range of the shaper and the noise is seen to have a 1/f dependence and a magnitude given by

\[ ENC_{\text{leakage}} = \frac{e}{q} \sqrt{\frac{q I_L T_p}{4}} \]  

(4.6)

where \( I_L \) is the leakage current of the sensor.

The serial noise has typically four sources. To begin with, serial resistance noise in the sensor and in the interconnects gives rise to a thermal noise with a white spectral density.

\[ ENC_{sr} = \frac{C_{tot} e}{q} \sqrt{\frac{k T R_s}{2T_p}} \]  

(4.7)

where \( C_{tot} \) is the total input capacitance, which is the sum of the sensor capacitance, interconnection capacitance and the gate capacitance of the input transistor, and \( R_s \) is the serial resistance.

Secondly, crystal defects that leads to trapping of charge carriers in the input transistor create Flicker-noise. For MOSFET transistors it has a 1/f frequency dependency\(^2\), while for JFET\(^3\) and bipolar transistors it is constant at low frequencies and has a 1/f\(^2\) dependence at higher frequencies. For JFET and bipolar transistors this noise source is negligible. For MOSFETs it can be

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\(^2\)MOSFET = Metal-oxide-semiconductor field-effect transistor.
\(^3\)JFET = Junction field-effect transistor.
expressed as

\[ ENC_{1/f} = \frac{C_{tot} e}{q} \sqrt{\frac{K_F}{2WL_{eff}}} \]  (4.8)

where \( K_F \) is a process-dependent constant, \( W \) is the width of the input transistor and \( L_{eff} \) is the length of the input transistor.

Thirdly, channel thermal noise in the input transistor is due to the resistance of the channel which gives rise to a white frequency spectrum.

\[ ENC_{ct} = \frac{C_t e}{q} \sqrt{\frac{nkT}{K_{inv} g_m T_p}} \]  (4.9)

where \( g_m \) is the gate-to-channel transconductance, \( n \) is the slope factor and \( K_{inv} \) is a constant that depends on if the channel is operated in strong inversion \( K_{inv} = 3 \) or in weak inversion, \( K_{inv} = 4 \). In reality, the measured value is found to be somewhere in between these two limiting values.

Lastly, the bulk serial resistance noise in the input transistor, which also has a white noise characteristic must be considered.

\[ ENC_{bulk} = \frac{C_t e}{q} \sqrt{\frac{R_{bulk} \eta^2 kT}{2T_p}} \]  (4.10)

where \( R_{bulk} \) is the bulk resistance and \( \eta \) is the ratio between the bulk-to-channel and gate-to-channel transconductances. The total equivalent noise charge is given by

\[ ENC_{tot} = \sqrt{ENC_{be}^2 + ENC_{pn}^2 + ENC_{sr}^2 + ENC_{1/f}^2 + ENC_{ct}^2 + ENC_{ bulk}^2} \]  (4.11)

The serial noise decreases with a lower input capacitance. The peaking time\(^4\) of the shaper determines the centre-frequency of its filter. A decrease in the peaking time increases the serial white noise, does not affect the serial 1/f noise and decreases the parallel noise. The use of a higher order filter will reduce the parallel noise by a factor of three and leave the serial noise virtually unaffected [39].

In the design, the peaking time is tuned to minimise the noise, which corresponds to equal contributions from the serial and parallel noise. The maximum allowable peaking time is in practise constrained by other design factors, as for example the maximum pulse-hit frequency that the circuit should be able to handle. For short peaking times the dominant noise is the serial noise and the improvement gained by selecting a higher-order filter is not very significant [39].

\(^4\)The peaking time is the time taken for the signal at the shaper output to reach its peak value.
Threshold dispersion
Process variations and non–uniform distributions of power and biases over the chip result in different effective threshold settings for the pixel discriminators. The threshold dispersion means that a pulse counted as a hit in one pixel, may escape being counted if it had been incident on another pixel. The response of the detector is thus not spatially shift-invariant. Local trimming of each pixel is a commonly used method to decrease the threshold dispersion. The trimming is done with a digital-to-analogue converter (DAC) or with a correction charge stored in a capacitor. The latter solution results in a very small threshold spread [40]. The threshold dispersion should at most comparable in size to, and preferably lower than the noise of the front-end amplifier, to give a good energy determination.

Pick-up and interference
The large number of pixel cells in a small area means that the digital and analogue parts in the pixel are located very close to each other. This may generate pick-up of digital signals in the front-end or cross-talk between pixels. Adding more metal layers decreases the impedance of power and bias lines, which reduces the spikes on the lines. Furthermore, with the addition of a metal layer, the sensor can be shielded from the readout chip. The switching of logic signals should be minimised during the data acquisition and the use of differential logic signal lines decreases the sensitivity to pick-up.

4.4 Radiation damage
The low photon energies used in X-ray imaging gives minimal damage to a DC-biased sensor. The readout chip will be affected due to damage of the MOS$^3$-structure in the transistors. The total dose effects due to radiation damage manifest themselves as threshold drifts, variation in the transconductance and as increased noise. In the case of a medical X-ray imaging detector, a design criterion is to maximise the quantum efficiency of the sensor. Preferably more than 70% of the photons are stopped in the sensor, a fact that reduces the radiation exposure of the readout chip.

Radiation tolerant readout chips can be fabricated using special radiation hard process technology, for example DMILL, but the cost is a concern and also the availability. The development of deep sub-micron technologies have made it possible to make radiation hard chips in standard processes, when certain design and layout rules are followed. The sub-micron process enables a smaller oxide thickness in the NMOS structure, which gives a reduced threshold sensitivity to the absorbed dose. Furthermore, the use of enclosed

MOS = Metal-oxide-semiconductor

22
NMOS\textsuperscript{6} structures and the use of guard rings make the chip much more radiation hard.[41]

4.5 Interconnection technologies

The division of the detector into two separate elements has advantages as was explained in the previous sections. A disadvantage with the hybrid solution is the very large number of fine-pitch interconnections that are necessary for connecting each pixel in the sensor to a pixel in the readout chip. Flip-chip bonding is the only available technique to connect the two parts at small pixel pitches. It is critical to keep a high yield in the flip-chip manufacture as unconnected pixels do not detect photons. The flip-chip bonding has to show long-term reliability and mechanical stability, withstand thermal cycling and last, but not least, be cost competitive.

In the flip-chip method, bumps have first to be deposited on the sensor electrodes and on the input pads of the readout chip. In a second step, the sensor and the readout chip are aligned before they are pressed together to form a single-unit detector. The flip-chip requires alignment with an accuracy better than one third of the bump diameter [42]. Some processes exhibit self-alignment by heating of the bumps, so-called reflow, which relaxes the demands on the alignment procedure.

The bumps are the electrically conductive paths from chip to sensor. They also provide the mechanical connection of the two pieces. An underfiller is sometimes employed to further enhance mechanical stability. Furthermore, the bump height controls the spacing between sensor and chip. Too small a spacing results in a large capacitive coupling between the two parts. On the other hand, a too large distance may increase the contact resistance and capacitance. Visual inspection, X-ray micro-radiography and finally irradiation with radioactive sources are methods to verify the quality of the interconnections [43].

The bump deposition is preceded by cleaning the chip. An under-bump-metalisation (UBM) step is then usually needed, as aluminium is not a suitable material for direct bump bonding. The most commonly employed bump formation technologies for large batches are solder bumps and indium bumps. Both approaches use photolithography to define the bumps and are performed only on full wafers.

The solder bumps made of Pb/Sn were introduced by IBM three decades ago. Advantages with this approach are very good electrical characteristics, bump uniformity, good self-alignment properties and high yield. A disadvantage is that the process requires high-temperature processing steps that may

\textsuperscript{6}NMOS = N-channel enhancement or depletion transistor.
strain the interconnections because of thermal expansion. Furthermore, the UBM step of this method is complex.[44]

Indium bumps were developed for infrared sensors. The process is carried out at moderate temperatures and allows a faulty chip to be replaced by reworking. The quality of the method depends heavily on a very good bump uniformity and planarity.[44]

An often used technology for prototyping is gold-stud bumping as it permits bonding of single dies. It consists of placing a gold ball on the pad with the use of a ball bonder that steps through the chip. In a reflow process, the balls are heated, to obtain a spherical bump. There is no need to use any UBM. The method can only be used on single chips and no self-alignment is provided. An advantage is that it is a lead-free process, which is good both for environmental reasons and that impurities in lead can give alpha emissions that may interact with the sensor to producing false counts. The gold-stud bonding is however not suited for processing large quantities.

Electroless plating does not need any photolithography and can be performed on both single dies and wafers. The bump material is Ni/Au. In this case, the two parts are not directly pressed towards each other, but an isotropic or anisotropic adhesive is used in between. The adhesive, which can be in the form of a glue or a film, contains small conductive spheres that, when squeezed between bumps, make the connection. Advantages are that it is a low temperature process, lead-free and low-cost. No photolithography masks are needed and a fine pitch is achievable. Disadvantages are that there is no self-alignment, higher electrical resistances and a lower yield than for Pb/Sn and In bumps [42]. Finally, yet another method is screen printing with a conductive glue having a very high yield, but the minimum pitch is today 200 \( \mu \)m [42].

4.6 Detector module construction

The physical sizes of the sensor and readout chip in a hybrid pixel detector are not sufficient to cover a large area. Silicon sensors can be fabricated up to 150 mm in diameter [45], but for the other materials the achievable sizes for good sensors are much smaller, for example CZT sensors exist up to about 2 cm\(^2\). The maximum size for a readout chip is around 25 \( \times \) 25 mm\(^2\), limited by the fabrication process and the production yield [45].

The restriction in size can be circumvented, by placing several readout chips side by side to connect to one large sensor, and a detector module is formed. The need for some space at the borders of the readout chips means that there will be an inefficient inter-chip region. A solution is to make larger rectangular pixels in the sensor to cover this region.
The readout-chip is connected to the external control system using wire-bonding. Power, biases, and logic signals are routed to the detector through a substrate that serves as both mechanical support and interface circuit. This implies that one edge of the readout chip has to extend beyond the sensor. It is preferable not to have any connection at the remaining three sides to make it possible to place readout chips as close together as possible. Several detector elements are tiled side by side to cover a large area. The largest reported hybrid pixel silicon detector for imaging to date covers $8 \times 18 \text{ cm}^2$. It consists of five modules that each comprises one sensor covering a matrix of $2 \times 8$ readout chips [46].

Another design approach is to use a 3D interconnection where the bond pads are located on the backside of the readout chip. Vias are made in the chip substrate to have wires connect the bond pads with the circuitry on the opposite side of the readout chip. The result is a detector with a minimum amount of dead regions. This technique has recently become commercially available.
5 The DIXI detector

The challenges in the medical X-ray imaging field together with the development of the hybrid pixel detector technology stimulated the start of the development of the DIXI detector. The main features of DIXI are a photon counting capability, an adjustable threshold and the implementation of two counters in each pixel cell. The number of hits per pixel is counted during a preset time interval and then transferred to a computer for processing and display. The two counters make it possible to acquire images very close in time under different conditions. Important features can be extracted through the processing of the two images.

5.1 Applications

The DIXI detector can be used for static or dynamic X-ray imaging and in particular angiography and fluoroscopy. The detector was developed especially for digital subtraction angiography (DSA), where the displayed image results from the subtraction of two image frames, to enhance the study of blood vessels. The present examination method gives a high radiation dose to the patient and a detector that can perform the examination at a lower dose is attractive.

Another application area for DIXI is bone mineral area density (BMD) measurements. BMD is used to diagnose and monitor osteoporosis for establishing the bone health. It serves as an intermediate marker for the prediction of risk of bone fractures. The Dual X-ray and Laser (DXL) technology combines X-ray images taken at two energies with a laser measurement of the bone thickness to determine the BMD [47, 48]. Today a charge integrating linear array detector is used. The DIXI detector has the potential to reduce both the measurement time and the radiation dose needed for the examination.

5.2 Angie – the readout chip

A central part of the DIXI detector is the readout chip Angie. It consists of 992 quadratic pixels arranged in a 31 × 32 matrix, with a pixel size of 270 × 270 µm². A pixel cell of the readout chip Angie is composed of a front-end
Figure 5.1: The circuit block diagram of Angie $31 \times 32$.

(preamplifier and shaper), a high-pass filter, a discriminator and two counters as is shown in Figure 5.1. The inclusion of two counters into the pixel has been made to enable the collection of two images separated in time by only 1 $\mu$s. A detailed description of the circuit is provided in Papers I–III and in Ref. [49]. The latest version of the chip was fabricated in a 0.8 $\mu$m CMOS process with two metal layers by AMS in Austria. The design of Angie was done by Ideas ASA together with Uppsala University. A photograph of the chip mounted on a test card is shown in Figure 5.2.

5.2.1 Performance of Angie

The performance of Angie is reported in Paper I – III, where the improvements in the results illustrates the deepened knowledge of the chip. Two test configurations were used to evaluate the performance of Angie. Wafers and diced chips have been characterised in a probe station. A chip is labelled as good if it has no short-circuits, draws a correct amount of bias currents and shows a uniform response for all pixel cells. The yield of good chips is 70%.

The tests have been performed with charge injection. A voltage step of known size is applied to a capacitor common to all chips located in the chip.

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1Austria Microsystems AG, A-8141 Schloss Premstätten, Austria.
2Ideas ASA, N-1323 Fornebu, Norway.
periphery and the charge is injected into one selected pixel. The biggest uncertainty in this method is the error due to the capacitor value, which is 20%. A more precise calibration of the injected charge could not be performed, due to the lack of a strong enough mono-energetic X-ray source. The charge from the capacitor is distributed to a selected pixel cell through a test injection network. This network is the dominant noise source when employed, as is shown in Figure 5.3.

Threshold scans were also performed for the case of injecting charges of different sizes. The threshold at which the obtained response curves intersected a specified counter response level was recorded and plotted as a function of injected charge, see Figure 5.4. Three different counter response levels, 10, 30 and 50 ADU, were used. A first order polynomial was fitted to the data points. The noise for the two cases of either using the test injection network or not was then calculated by utilising the fit parameters. The true noise of the circuit is 138 e$^-$, while the noise with connected test network is 2454 e$^-$. All reported results that include charge injection are influenced by this noise.

The detector should have a uniform response over its surface. The major reason for a non-uniform response in the readout chip is found in the threshold variations between pixels. The threshold dispersion reported in Paper II is 365 e$^-$ for an injected charge of 2 fC (12500 e$^-$) and 306 e$^-$ for 1 fC (6250 e$^-$). The results in Paper II is a factor of 50% better than what was reported in
**Figure 5.3**: Two threshold scans of the counter response with no test pulse injected, only noise, shows the difference in noise level between having or not having a connected test injection network. Also plotted is a threshold scan with a killed front-end. The counter response is reported in analogue-to-digital units (ADU). From Paper III.

**Figure 5.4**: The threshold at which a threshold scan intersects a counter response level as a function of injected charge. A first-order polynomial has been fitted to obtain the gain and offset. Three different counter response levels, 10, 30 and 50 ADU, were used to obtain three fits. From Paper III.
Figure 5.5: The mean threshold per column as a function of column position for an injected charge of 2 fC (12500 e\textsuperscript{−}). From Paper III.

Paper I; the improvement is due to optimised bias settings.

The threshold dispersion shows a strong column dependence as displayed in Figure 5.5. The further away from the centre column, the more the mean threshold deviates from that of the centre column. No dependence on the row position is noted. The column dependence of the dispersion is due to voltage drops in the ground lines. Correcting for this drop, the threshold dispersion shows no column or row dependence and a dispersion of 169 e\textsuperscript{−} is calculated for an injected charge of 1 fC.

The relation between the injected charge and the counter response is described by the transfer function, which depends on the biasing of the chip. It is measured by scanning the size of the injected charge and recording the counter response. Figure 5.6 shows a typical measured transfer function for Angie, which is divided into three regions:

1. Flat region with zero response.
2. Transition region described by a tanh-function.
3. Linear slope region.

From the measured transfer function, the coefficient of energy weighting, $\alpha$, can be calculated for region 3, where $\alpha = 0.1$, which is close to the performance of an ideal photon counter that has $\alpha = 0$. A charge integrating device has $\alpha = 1$.

The hit information is stored as small voltage steps in an capacitor that serves as an analogue counter. The single-ended signal from the chip is converted into a differential signal on the test card before being transferred to the control system, where it is converted to a digital word in a 16-bit analogue-to-
digital converter (ADC). A dynamic range of 12 bits has been achieved. The image acquisition and the readout gives rise to a measured root-mean-square (RMS) value of the readout noise of 6 analogue-to-digital units (ADU). When oversampling is used the noise figure is reduced to 4 ADU. The readout is then carried out at a lower speed.

A capacitor discharges with time and this affects the readout chip for long exposure times, i.e. more than some seconds. The drift of the counter has been measured to be 2 % per minute. It is, however, possible to compensate for the drift in post-processing, with the penalty of a reduced dynamic range.

Angie is designed to work with a sensor that collects holes, but some compound semiconductors require electron collection. Studies have shown that by tuning the pulse shape from the front-end it is feasible to count photons also in electron collection mode. The gain in the front-end is, however, decreased by a factor of four.

A test pixel placed outside of the pixel matrix provides valuable qualitative information on the chip performance. The pulse shape in the front-end chain and the signal and noise propagation through the circuit can be studied. The noise of the front-end could not be measured quantitatively, since the noise of the injected charge is the dominant noise source. The reason for this large noise contribution is the use of an external larger injection capacitor\(^3\), as consequence very small voltage steps have to be used.

\(^3\)The test pixel cannot use the internal injection capacitor.
Figure 5.7: A 500 µm thick silicon wafer with different sensor structures.

5.3 Fabrication of silicon sensors

Silicon sensors have been manufactured for the first prototype series of DIXI. The properties of silicon are well known and high quality sensors can easily be manufactured at low cost. Sensors of three different thicknesses (300, 500 and 1000 µm) were designed by Uppsala Universitet and processed on high resistivity substrates at VTT. A photography of a wafer is shown in Figure 5.7. The design includes a single unit sensor that covers one readout chip and a ladder that covers eight readout chips. The readout chip has been designed with minimal dead space at the edges, thus allowing several chips to be assembled on the ladder. In addition, four other sensor structure were fabricated. One smaller pixel sensor, 11 × 11 pixels, with fan-out to connections on the four sides, is meant for spectroscopic studies of the DIXI detector. Two structures are dedicated for use in bone densitometry measurements and finally there is a test diode. The sensors are designed to be DC-coupled to the readout chip.

VTT, Espoo, Finland
5.4 Anisotropic conductive film as interconnection technology

Anisotropic conductive film, together with an electroless Ni-Au bump deposition, was chosen as the interconnection technology for DIXI. The reasons were low cost and the availability of single-chip processing. The work is done at IVF\(^5\) in collaboration with Uppsala Universitet.

5.4.1 Electroless nickel plating

The contacts on sensor and readout chip are surrounded by a passivated area that is elevated with respect to the contacts. The contacts pads are made higher by plating nickel bumps on the pads. The height of the bumps is determined by the properties of the anisotropic film that is placed between the sensor and the readout chip. The bump height needs to be twice the particle diameter in the film. For the DIXI geometry, where the sensor has a much larger area than the chip bump, the required bump height is 15 $\mu$m. Bumps are grown on both the readout chip and the sensor.

The formation of the bumps is achieved through an electroless nickel plating process that requires no lithography masks. The process steps are:

1. Pad preconditioning
2. Zincate treatment
3. Electroless Ni bump deposition
4. Gold coating

The speed of growing the bumps is strongly dependent on the contact size. Prior to the plating process, the large wire-bonding pads are therefore protected with an adhesive coating to prevent short-circuits between neighbouring pads. The edges and backside of the sensor are also coated, since any deposition on the edges and backside has a very low adhesion and will gradually fall off.

**Pad preconditioning**

The success of the process requires a good surface preparation to remove contaminations on the bond pad, which is first cleaned and then etched with HNO\(_3\) to remove the aluminium oxide. The surface should be smooth and uniform to ensure a good surface coverage of the zinc layer in the next step.

**Zincate treatment**

The etching in the first step removes the aluminium oxide. However, a thin oxide layer will be formed immediately afterwards, and it prevents the aluminium from reacting with nickel. Therefore a zinc layer is processed through

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\(^5\)IVF, SE-431 53 Mölndal, Sweden

34
an alkaline zincation process, also known as double zincation. The chip is put twice in the solution and in between it is immersed in a HNO$_3$ solution. The first zincation step removes all remaining aluminium oxide and deposits a coarse zinc layer. The second step creates a finer more uniform zinc layer. The size and uniformity of the zinc layer have a strong influence on the bump formation. If this step is carried out in an improper way, the result is an inhomogeneous growth of the Ni bumps, or no growth at all. A non-uniform layer leads to lower shear strength and a higher contact resistance. The Zn layer needs only to be around 0.5 $\mu$m thick.

**Bump deposition**

The bump formation is performed in a plating bath that is based on a commercially available electroless Ni bath. At a temperature of 80 degrees Celsius the bump grows with a speed of 3 $\mu$m per hour for the readout chip and 5 $\mu$m per hour for the sensor. This process step is sensitive to any potential difference between pads, a difference that can lead to an inhomogeneous bump height distribution.

**Gold coating**

When the correct height has been obtained the bumps are plated with a thin layer of gold to prevent oxidation.

Photographs of a contact pad on the readout chip before the nickel plating and resulting nickel bump are is shown in Figures 5.8 and Figure 5.9. An image showing 16 bumps on the Angie chip is shown in Figure 5.10. The bump height is 14 $\mu$m and the variation over the chip surface is less than 1 $\mu$m.
Figure 5.9: A contact pad on the readout chip with a 14 $\mu$m high nickel bump.

Figure 5.10: Several bumps on the readout chip with a height of 14 $\mu$m.
The readout chip is sensitive to electrostatic damage and in the handling and processing the chip need to be protected according to electrostatic devices (ESD) guidelines to prevent fatal damages.

5.4.2 The flip-chip process with anisotropic conductive film

After the bump process, the electronic chip and the sensor need to be tested to verify full functionality, before they are flip-chip bonded. It is not sufficient to perform tests on single diced chip placed in the probe station to obtain the necessary information on its performance. This is particularly true when a ladder consisting of eight chips is to be flip-chip bonded. A method by which chips already mounted and wire-bonded to a test card can be flip-chip bonded has therefore been developed. Another advantage is that it is considerably easier to do the wire-bonding before the sensor is put on top of the readout chip. The flip-chip process in this project has only been tested on diced chips and a fully working DIXI device is yet to be made.

5.5 Prototype detector of a silicon sensor wire-bonded to Angie

To test the functionality of the chip, before the completion of a flip-chip bonded detector unit, nine pixels were wire-bonded to a 500 μm thick silicon pixel sensor. Figure 5.11 shows an image obtained by subtracting one image taken with an Am$^{241}$ radioactive source and one without any source. The exposure time was two minutes in both cases. The average number of hits in per pixel is 50. The image demonstrates the successful operation of the readout chip with a sensor. The small intensity of the source makes it impossible to carry out more extensive studies of the correct working point for the biasing of the front-end and the calibration of the injected charge.

5.6 The ANDAQ control system

The original control system, an adaption of a control system constructed for commercial readout chips, was not optimal for Angie. A complete control system has therefore been designed and fabricated for the DIXI detector, named ANDAQ. It is designed for both image acquisition and for evaluation of the DIXI detector. The control system supplies all the necessary voltages and biases needed by the Angie readout chip. The system generates the logic signals to the readout chip and is able to read out the images acquired by the DIXI detector. The system may trigger an external X-ray tube or be triggered by the
Figure 5.11: Image taken with an Am$^{241}$ source. Nine pixels have been wire-bonded to a 500 µm thick silicon sensor. Black and grey pixels corresponds to counts in the detector.

X-ray tube. The bias voltage to the sensor is not supplied by the control system. The analogue data from the detector arrays are first converted in a 16-bit ADC, before they are transmitted to the computer for storage and display.

In order to accommodate various system configurations for DIXI, the control system is designed in a modular fashion. The functions needed in a particular design are obtained by adding or removing modules in the system crate. The ANDAQ system can handle different test boards, due to its modular design. In Figure 5.12 it is connected to a test board that houses one Angie chip. The displayed system is configured to be able to control and readout two modules of $1 \times 8$ chips each. More modules can easily be controlled by adding cards in the crate. The same system can also test chips or full wafers placed in the probe station.

The computer control program is written in LabView\textsuperscript{6} and communicates with the control system in the crate over USB 1.1. The maximum transfer rate of data is in the present configuration 3 Mbit/s. The communication between the modules is handled by a two-wire serial interface (TWI). Work is currently under way to replace the USB 1.1 communication protocol with USB 2.0 to increase the data transfer rate to support the acquisition of full-sized images at 30 frames-per-second (fps).

Figure 5.12: The ANDAQ control system connected to a test card with one Angie chip.

The control program has functions to acquire images for display, perform flat-field corrections of images, calibrate ANDAQ, perform threshold and charge scans, and enable a closer examination of a single pixel. An example of the user-interface is shown in Figure 5.13.

5.7 Image quality assessment through theoretical calculations and full-size detector simulations

Two approaches have been considered to assess the image quality of the DIXI detector before actual measurements on a physical system take place. The basic motivations are to optimise the design and operation of the detector and to investigate how different sensor materials influence its performance.

The theoretical model provides a quick way to test how design changes influence the behaviour of the detector. For example, how a change in noise level or of sensor material influences the image quality of the detector. Its main strength is the evaluation of new design strategies. The main limitation is the fact that it is only a model that relies on assumptions and on the underlying theory of linear systems.

The full-size detector-simulation approach gives a better prediction of the DIXI performance than the theoretical model. This concerns in particular the non shift-invariance of the system\(^7\), which the theoretical model cannot address. Furthermore, it is possible to use a phantom to assess task-based image quality figures-of-merit.

\(^7\)For example threshold dispersion.
The simulation approach is dealt with in Paper IV, while the theoretical modelling is developed in Paper V and VI. The two methods are compared in Paper VI.

5.7.1 Full-size simulations of a detector module

Simulation tools have been developed to simulate a full-size DIXI detector module. The module has 256×256 pixels and consists of 4×4 Angie chips and a silicon sensor. The simulation of the interactions of the photons with the sensor and the photon transport in the sensor are modelled with the Monte Carlo (MC) based code GEANT\textsuperscript{8} on the NorduGrid cluster \cite{50}. The charge spread is calculated analytically \cite{51}. Different exposures were simulated and saved on file. The data from the MC runs were processed through a code simulating the readout chip. The simulation of the signal processing in the readout chip is done in a program written in the LabView programming environment. The input to this part are measured Angie transfer functions and noise figures for both the ENC at the input of the Angie chip and the rms noise of the readout.

The simulated configuration is a 500 \( \mu \text{m} \) thick silicon sensor, an X-ray spectrum typical of clinical examinations, an input noise of 200 e\(^{-}\) ENC and a readout noise of 9 ADU rms. The transfer function for the readout chip was taken from real measurements. Both a flood-field image and a pencil beam are used in the simulations. The former is used to determine the NPS, while the latter is used in the determination of the EMTF. Exposures in the range $0.01 - $
0.25 μGy were used.

The results show that the detector operates quantum limited down to an exposure of 0.08 μGy. Below this dose the readout noise becomes the dominant noise source and the DQE drops. Above 0.08 μGy the DQE is as expected equal to the quantum efficiency.

It was also found that a decrease in the threshold gives an increase in DQE as long as the threshold is kept well above the input noise. Nonetheless, the optimal threshold setting is not at the low limit. The reason is that charge sharing occurs in the border region between two pixels, which degrades the spatial resolution. Depending on the threshold, the events, for which charge is collected in several pixels, can either be missed, single-counted or double-counted. By optimising the threshold, small charge fragments can be filtered away, while the large ones are still counted. The spatial resolution will thus improve with increasing threshold, approaching that of a system without charge sharing. To conclude, it was observed that an optimal threshold can be found that cause negligible efficiency degradation, but still does not degrade the spatial resolution too much. The optimum threshold corresponds to half of the mean energy of the spectrum.

In Paper III, the same transfer function was used for all pixel cells. In reality the chips have a threshold dispersion, and work has recently been done to include that dispersion in the simulations. The transfer functions for all the pixels in one chip were therefore measured and used in the simulations. No flat-field correction was performed. Figure 5.14 shows the DQE as a function of the exposure dose using a common or individual transfer function in the

Figure 5.14: The DQE at zero spatial frequency as a function of the exposure. The data points are with threshold dispersion, “on” or without threshold dispersion, “off”.

Threshold dispersion
- Off
- On

Exposure, Air Kerma (μGy)
DQE(0)
0.01
0.02
0.03
0.04
0.05
0
0.05
0.1
0.15
0.2
0.25
pixels. It can be concluded that the threshold dispersion of the Angie chip for this threshold setting has no influence on the DQE.

5.7.2 Modelling of a hybrid pixel detector

A theoretical model of a hybrid pixel detector has been constructed using the cascaded linear analysis framework [52]. The detector is treated as a cascade of several independent stages that involve different physical processes. The ordering of the stages matters and generally they do not commute. The theory can predict the image quality parameters, such as NPS and DQE, through an analytical expression. The motivation for this model is to have a fast tool for optimising the detector system. Different configurations can be tested before a prototype is constructed.

Linear-system theory applied to image quality assessment has earlier only been used for charge integrating detectors. The novelties of the theory described in this thesis are the modelling of:

1. The response of the readout chip and its piece-wise linearisation.
2. The analysis of the role of the discriminator in the minimisation of charge sharing.

The first version of the model includes the interaction of the X-ray photon with a silicon sensor, diffusion of produced charge, spatial sampling and the transfer through the readout chip. The modelling of the readout chip was not fully developed in this version. This modelling is not straightforward as it is described by a non-linear transfer function. Nevertheless, in the improved model, a piecewise linear model is used. The model also includes an improved treatment of the readout noise. The second major advancement compared to the first version is the handling of charge sharing between pixels.

The model can be used generally for hybrid pixel detectors and has been tested on the DIXI detector. The NPS and DQE are calculated for different noise conditions. The results are compared to results obtained from the full-field detector simulations and they agree within 15%. Furthermore, it is shown that the model can in a satisfactory way describe changes in the image quality parameters due to different noise conditions.
6 Summary of papers

Paper I
The first paper gives an introduction to the circuit design of Angie 31×32 and on the properties of the processed chip. The yield of working chips is found to be 70%. The threshold dispersion is measured to be 550 e−, for an input signal of 12500 e−. The input to output relationship of the chip, given by the transfer function, exhibits a transition region beyond which the chip operates as a photon counter. The tests were performed and analysed by me and I wrote the article.

Paper II
This paper gives a brief introduction to the DIXI detector and to the performance of the Angie readout chip. It is shown that the transfer characteristic of the chip can be tuned. The threshold dispersion is measured to be 365 e− for hole collection and an input signal of 12500 e−. This is an improvement by 30% compared to earlier results. It was further demonstrated that electrons can be collected, however with a factor of four reduced gain. It is also shown that the dispersion has a column dependence, but no row dependence. The dynamic range of the counter is 12 bits, but the voltage output of the counter exhibits a small drift with time. It is possible to compensate for the drift in a post-processing of the image. I did the measurements and analysed the results. The article was written by me.

Paper III
The paper presents data on the transfer characteristics and on the counter and noise performance of Angie. Insight into the transfer through the chip of the signal-to-noise ratio is gained from measurements on a test pixel that provides information at several internal nodes in the pixel circuitry. The column dependence of the threshold dispersion is caused by voltage drops in the ground lines on the chip. The equivalent noise charge (ENC) at the input of the chip is measured to be 138 e−. A standard mode of operation is found, but the chip can be biased differently to maximise either the dynamic range of the
counter or the signal-to-noise ratio. An energy weighting coefficient, $\alpha$ of 0.1 has been achieved. An ideal photon counter has $\alpha = 0$ and a charge integrator has $\alpha = 1$. The performance of Angie is thus close to that of a perfect photon counter. I carried out the tests and analysed the results. The article was written by me.

Paper IV

The simulation tool for DIXI is described and the results from simulations of a typical X-ray spectrum are presented. The paper shows results on how the threshold setting, noise sources, level of exposure and charge sharing influence the detective quantum efficiency (DQE) of the detector. The discrete spatial distribution of the deposited charge (sensor part) was generated with GEANT and then, the response of the readout chip Angie (readout part) was simulated in a program written in the LabView programming environment. For this second part of the simulation, the transfer characteristic of the readout chip was taken from experimental measurements. My contributions were the development of the simulation program for the response of the readout chip, the calculations of image quality parameters and the analysis of image quality parameters. The part of the article that concerns the simulations of the chip was written by me and I have also calculated the data for the plots.

Paper V

The paper deals with the modelling of the readout part of the detector and the first description of the detector as a linear system. The transport of X-ray photons in the sensor was studied with the Monte Carlo package GEANT and the charge transport was treated analytically. The readout chip characterisation was based on simulations of its transfer function in the circuit simulator PSpice\(^1\). The treatment of the input noise was implemented in the MC program and was used as the overall noise figure. The calculations of the detector response, noise power spectrum (NPS) and detective quantum efficiency (DQE) were implemented in MATLAB\(^2\). The DQE of DIXI with a silicon sensor using two different readout transfer functions was computed. Furthermore, extreme cases such as charge integration and perfect photon counting were evaluated. I performed the PSpice simulations of the readout chip and contributed to the development of the model.

\(^1\)PSpice 9.2, Circuit simulator, Cadence Design Systems, Inc., USA
\(^2\)MATLAB 6.5, trademark of Mathwork (www.mathwork.com)
Paper VI

A way of applying the cascaded linear-system transfer theory to an X-ray hybrid pixel detectors is proposed. The threshold function, characteristic of counting hybrid pixel detectors, is piece-wise linearised and a de-blurring stage describes how different threshold settings influence the charge sharing between pixels. The paper provides a prediction tool for hybrid pixel detectors, where a single analytical expression may be evaluated to obtain the detective quantum efficiency (DQE). I contributed to the development of the theoretical description of the readout chip and the charge sharing model and participated in the writing of the article.
7 Conclusions and outlook

Several goals have been achieved in this work. The readout chip Angie functions as designed, a silicon sensor has been fabricated, a low cost flip-chip bonding alternative shows promising results and a control system is in operation. A tool for full-size detector simulations has been set up and a model for quick evaluations of hybrid pixel detectors has been elaborated. The parts work together in the pursuit of a detector that may lower the radiation doses needed in radiology. The complete DIXI detector is still to be assembled, but with the components developed in this project an X-ray detector that lowers the doses in radiology is near its completion.

Angie is a fully functional readout chip that processes single X-ray photon hits, uses an energy threshold and stores the information in one of the two counters. The two counters make it possible to acquire two images closely separated in time. The chip has nearly the same energy weighting as an ideal photon counter, thus being superior to a charge integrating detector. An ideal photon counter has an energy weighting coefficient $\alpha = 0$, while Angie has $\alpha = 0.1$, compared to a charge integrator that has $\alpha = 1$. The weighting coefficient for Angie cannot be lowered any further without increasing the noise dramatically.

The work presented here has shown that the Angie chip can be improved with minor modifications in the design. The threshold dispersion is larger than the measured noise, but the reason for the systematic threshold variation over the chip has been found. An reduced impedance of the ground lines will remove the dependence of the threshold dispersion on the column. This can be done by adding one more metal layer in the design. To reduce the dispersion even further, a threshold tuning circuit can be introduced. This would require a slightly larger modification of the design by inserting in each cell a trim-DAC. The most important benefit of a smaller threshold variation is the possibility to decrease the lowest threshold setting, to enable the use of the detector in soft X-ray applications.

Another desirable change is to make Angie to work equally well with electron as with hole collection. Collecting electrons is important when compound semiconductor sensors are to be used instead of silicon.

The use of anisotropic conductive film used with electroless plated Ni/Au bumps has been developed as an interconnection method. The technology dif-
ficulties experienced in the electroless plating have been solved and the final step of flip-chip bonding the readout chip with the sensor in order to obtain a complete detector is near its realisation. The use of anisotropic conductive film is a low cost alternative that in addition can be used for both diced chips and full wafers. An extra bonus is that the technology is environment friendly. However, the bump deposition still needs be optimised regarding the adhesive coating with the purpose to reduce cost and simplify the process. This can be accomplished for the readout chip by using a larger space between the wire-bonding pads to protect bumps growing on these pads to come in contact with each other. The backside of the sensor has to be protected with adhesive coating, and for single-chip processing, also the edges of the sensor have to be coated.

The image-quality parameters for a detector with a 500 \(\mu\)m thick silicon sensor have been determined, both in simulations and in linear cascaded theory calculations. The results of the two approaches differ by not more than 15%. The simulations are more precise than the calculations, but also more time-consuming and it is therefore preferable to use calculations for fast evaluations of a detector configuration. DIXI was found to be quantum-noise limited down to an exposure of 0.08 \(\mu\)Gy.

The simulation of the image quality for the DIXI detector shows that the limiting factor for the detective quantum efficiency (DQE) at low exposures is the readout noise. In applications with low exposures, for example fluoroscopy, the needed dynamic range is lower than for applications with higher exposures. To minimise the patient dose, while a good image quality is maintained, an operating mode of Angie can be chosen that maximises the step size and reduces the influence of readout noise. The detector can then be operated quantum limited at even lower exposure settings. In an application that requires a larger dynamic range, an operating mode with smaller step size can be selected.

Up to now a silicon sensor has been employed in the simulations and modelling, but other sensor materials will be investigated. The primary motivation is to increase the quantum efficiency of the detector to reduce the patient dose. The verification of the image-quality results can be compared to measurements on the complete detector. Another task is to include a virtual body phantom in the Monte-Carlo simulations to determine task-based image quality figures-of-merit.
DIXI – en hybridpixeldetektor för röntgenavbildning


Medicinsk röntgenavbildning


Med denna bakgrund så kan man konstatera att det alltid är av stor vikt att minimera stråldosen till patienten. Detta görs genom att utnyttja all den in-
formation som finns i flödet av de röntgenfotoner som träffar detektorn. När foterna passerar kroppen så växelverkar lågenergetiska fotoner en högre utsträckning än högenergetiska fotoner. Detta innebär att lågenergetiska fotoner bär på mer information om kroppens kontrastförhållanden, jämfört med högenergetiska fotoner. Den maximala informationen från detekterade fotoner erhålls genom att vikta fotonernas energi, $E$, med viktningskoefficienten $\alpha$. Detektorns respons, $R$, ges då av följande uttryck:

$$R \propto E^\alpha$$  \hspace{1cm} (8.1)

En ideal detektor som tar till vara maximal information har $\alpha = -3$. Detta ska jämföras med nästan alla kommersiella detektorer som använder en teknik där den totalt deponerade laddningen i detektorn summers. Denna inkluderar allt brus, exempelvis överhörning mellan pixlar och läckströmmar. En laddningsintegrerande detektor har $\alpha = 1$, vilket är en skillnad med fyra storleksordningar relativt den optimala vikningen. De senaste åren har mycket forskning skett om fotonräknande detektorer. För dessa ger varje foton med energi överstigande en bestämd tröskelnivå upphov till ett steg i ett binärt räkneregister, $\alpha = 0$. I simuleringar har det visats att för att uppnå en viss kvalitet på röntgenbilden krävs mindre stråldos för en fotonräknande detektor än för en laddningsintegrerande detektor.

Hybridpixeldetektorer

Som nämnts tidigare är möjligheten till lägre stråldos en av drivkrafterna i utvecklingen av hybridpixeldetektorn för medicinsk röntgenavbildning. Andra fördelar är att den har en räknare med stort dynamiskt omfång och linjär respons, vilket minskar risken för över- och underexponering.

En hybridpixeldetektor är som namnet antyder en hybrid sammansatt av två delar: en sensor och ett utläsningschip. Båda är indelade i pixlar, vilka oftast är kvadratiska, och varje sensorpixel är kontakterad till en pixel i utläsningschipet. På så sätt kan varje röntgenfoton som träffar en pixel behandlas individuellt, oberoende av vad som sker i närliggande pixlar. En fördel med hybridiseringen är att de två delarna kan optimeras var för sig, medan en nackdel är att det krävs ett stort antal kontakter.

I sensorn deponerar fotonerna en elektrisk laddning som transporteras till utläsningschipet, där den jämförs med en tröskelnivå. Om laddningens storlek överskrider tröskelvärdet sparas informationen, vilket för en fotonräknare betyder att en träff registreras. Efter en bestämd exponeringstid överförs bilden till en dator och kan visas på en bildskärm, varefter den kan arkiveras för senare användning. Bildbehandling är också möjlig att utföra för att exempelvis ändra kontrasten i syfte att förstärka strukturer av värde för den kliniska diagnosen.

Det vanligaste sensormaterialet är idag kisel tack vare ett lågt pris och möjlighet att tillverka stora sensorer med komplicerade strukturer. Den största invändningen mot kisel är dess låga atomnummer som ger dålig effektivitet för detektion av röntgenfotoner med energi överstigande 10 keV. Andra material med högre effektivitet, såsom gallium-arsenid, cadmium-zink-tellurid och kvicksilverjodid är under utveckling, men nackdelar är bland annat ofullständig laddningsinsamling, högt pris och att endast små sensorer kan tillverkas.


Vanliga metoder är indiumbumpar och Sn/Pb-bumpar. Dessa kräver emellertid litografiska masker och kan endast utföras på hela ”wafers” av chip/sensorer. Om man vill sätta ihop sågade chip/sensorer kan antingen ”guld-stift” bondning eller anisotrop ledande film med ”electroless” nickelbumpar användas. Den senare kan appliceras både på hela wafers och på utsågade chips och är dessutom både billig och miljövänlig.

**DIXI–detektorn**


Ett utläsningschip, Angie, har designats och utvecklats speciellt för DIXI.
Angies ekvivalenta ingångsbrus har bestämts till 138 e−. Tröskelspridningen mellan pixlar är 306 e− för en detekterad laddningssignal om 6250 e− och den uppvissar ett beroende av avståndet till mittkolumnen av chipet. Orsaken till tröskeldispersionen är jordledningarna, vars impedans ger uppstigningsfall. Vidare har energiviktningskoefficienten $\alpha$ uppmätts till 0,1, vilket är nästan lika bra som för en ideal fotonräknare och väsentligt bättre än för en laddningsintegrerande detektor.

Kiselsensorer om tre olika tjocklekar, 300, 500 och 1000 $\mu$m, har tillverkats för de första detektorprototyperna. För sammanfogningen av de två delarna har en metod baserad på anisotrop ledande film och nickelbumpar utvecklats. De inledande problemen ifråga om nickelpåläteningen har lösts, men trådbondningskontakterna på chipet måste skyddas från nickelefonering, för att undvika kortslutning. En komplett detektorenhet är under ihopmontering. För att testa funktionaliteten hos Angie kontaktades nio pixlar på ett utläsningschip och en sensorn med hjälp av trådbondning. Taster med ett $\mathrm{Am}^{241}$-preparat visar att denna detektorprototyp fungerar som förväntat.

Fortsättningsvis har ett kontrollsystem, ANDAQ utvecklats och byggs för kontroll och utvärdering av detektorn. Det kan användas både för testerna av Angie samt för att styrja den fullskaliga DIXI-detektorn.

Simuleringsverktyg för att simulera en fullskalig DIXI-detektor om $4 \times 4$ cm$^2$ har tagits fram. En detektor med 500 $\mu$m tjock kiselsensor har simulerats för olika brusnivåer och exponeringar. Resultaten från simuleringarna visar att detektorn är kvantbegränsad ner till en exponering om 0,08 $\mu$Gy. I en kvantbegränsad detektor är variationen i antalet infallande röntgenfotoner större än bruset i detektorn, dvs detektorns brusnivå försämrrar inte nämnvärt kvaliteten på bilden. För lägre exponeringar än 0,08 $\mu$Gy är detektorn emellertid begränsad av sitt utläsningsbrus, men bruset på ingången av utläsningschipet samt tröskeldispersionen är fortfarande av mindre betydelse.


Slutligen, har en teoretisk modell för hybridpixeldetektorer utvecklats. Den bygger på linjär systemteori där en beskrivning har tagits fram av funktionen hos en elektronikkrets med diskriminator och av hur inställningen av diskriminatortröskeln kan minimera inverkan av laddningsdelning mellan pixlar. Modellen är generell, men den har utvärderats med hjälp av DIXI. Parametrar för bestämning av bildkvalitet har beräknats och jämförts med simulerade vär-
den; resultaten överensstämmer inom 15 %. Modellen kan användas för att snabbt utvärdera hur förändringar i detektorns konfigurationen påverkar dess prestanda.

Acronyms

<table>
<thead>
<tr>
<th>Acronym</th>
<th>Description</th>
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<tbody>
<tr>
<td>ADC</td>
<td>Analogue-to-digital converter</td>
</tr>
<tr>
<td>ADU</td>
<td>Analogue-to-digital units</td>
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<tr>
<td>ASIC</td>
<td>Application Specific Integrated Circuit</td>
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<tr>
<td>BMD</td>
<td>Bone-mineral area density</td>
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<td>CCD</td>
<td>Charge-coupled device</td>
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<tr>
<td>CMOS</td>
<td>Complementary Metal Oxide Semiconductor</td>
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<td>CSA</td>
<td>Charge sensitive amplifier</td>
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<td>CZT</td>
<td>Cadmium Zinc Telluride</td>
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<td>CT</td>
<td>Compute tomography</td>
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<td>DAP</td>
<td>Dose-area-product</td>
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<td>DAC</td>
<td>Digital-to-analogue converter</td>
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<td>DSA</td>
<td>Digital subtraction angiography</td>
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<tr>
<td>DQE</td>
<td>Detective quantum efficiency</td>
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<tr>
<td>DXL</td>
<td>Dual X-ray and Laser technology</td>
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<td>e-h</td>
<td>Electron-hole pair</td>
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<td>EMTF</td>
<td>Expectation modulation transfer function</td>
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<td>ENC</td>
<td>Equivalent noise charge</td>
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<td>FPD</td>
<td>Flat panel detector</td>
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<tr>
<td>FT</td>
<td>Fourier transform</td>
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<td>Gy</td>
<td>Gray, unit of absorbed dose</td>
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<td>JFET</td>
<td>Junction-field-effect transistor</td>
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<td>MC</td>
<td>Monte Carlo</td>
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<tr>
<td>MOS</td>
<td>Metal-oxide-semiconductor</td>
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<tr>
<td>MOSFET</td>
<td>Metal-oxide-silicon field-effect transistor</td>
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<td>MRI</td>
<td>Magnetic resonance imaging</td>
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<td>MTF</td>
<td>Modulation transfer function</td>
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<td>NEQ</td>
<td>Noise equivalent quanta</td>
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<td>Acronym</td>
<td>Description</td>
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<tr>
<td>NMOS</td>
<td>N-channel enhancement and depletion transistors</td>
</tr>
<tr>
<td>NPS</td>
<td>Noise power spectrum</td>
</tr>
<tr>
<td>OTF</td>
<td>Optical transfer function</td>
</tr>
<tr>
<td>PET</td>
<td>Positron emission tomography</td>
</tr>
<tr>
<td>PSF</td>
<td>Point spread function</td>
</tr>
<tr>
<td>RBE</td>
<td>Relative biological effectiveness</td>
</tr>
<tr>
<td>RMS</td>
<td>Root-mean-square</td>
</tr>
<tr>
<td>SNR</td>
<td>Signal-to-noise ratio</td>
</tr>
<tr>
<td>Sv</td>
<td>Sievert, unit of effective dose</td>
</tr>
<tr>
<td>TFT</td>
<td>Thin film transistor</td>
</tr>
<tr>
<td>TWI</td>
<td>Two-wire interface</td>
</tr>
<tr>
<td>UBM</td>
<td>Under-bump-metallisation</td>
</tr>
<tr>
<td>USB</td>
<td>Universal-serial bus</td>
</tr>
</tbody>
</table>
Nomenclature

\( \alpha \) Energy weighting coefficient

\( C_{\text{tot}} \) Total input capacitance, which is the sum of the detector
  capacitance plus the gate capacitance of the input transistor.

\( \overline{d} \) Mean number of hits per cm\(^2\) in the image.

\( \varepsilon \) The energy needed create an electron-hole pair.

\( \eta \) The ratio between the bulk-to-channel and gate-to-channel
  transconductances.

\( E \) Energy

\( I_L \) Leakage current of the sensor

\( k \) Boltzmann’s constant

\( K_F \) CMOS process-dependent constant

\( K_{\text{inv}} \) A constant that depends on if the channel is operated in strong
  inversion \( K_{\text{inv}} = 3 \) or in weak inversion, \( K_{\text{inv}} = 4 \).

\( \mu \) Carrier mobility

\( \mu_N \) Nyquist frequency

\( \mu_{\text{tot}} \) Total absorption coefficient

\( n \) Slope factor

\( q \) Electron charge

\( q \) Quantum flux per cm\(^2\)

\( q' \) Noise equivalent quanta

\( R \) Detector response

\( L_{\text{eff}} \) Length of the input transistor

\( g_m \) Gate-to-channel transconductance

\( R_{\text{bulk}} \) Bulk resistance
\( R_p \)  Parallel resistance of the sensor bias resistor and the preamplifier feedback resistor.
\( R_s \)  Serial resistance
\( \tau \)  Carrier lifetime
\( T \)  Temperature in Kelvin
\( T_p \)  Peaking time of shaper
\( u, v \)  Spatial frequencies
\( W \)  Width of the input transistor
\( w_i \)  Energy weighting factor
\( Z \)  Atomic number of a material
Acknowledgements

The intense work with my thesis is soon about to end, and time has come to summarise and reflect over the past years of research. There is with mixed feelings I now face the closure, which entails that emotions of joy, relief, loss and gratitude are equally present and manifest. There is a huge joy and relief to witness the creation of a thesis substantiated by many hard hours in the laboratory. Unfortunately, an ending of something is often accompanied by a feeling of loss and so also in this case. However – and most importantly – I experience an immense gratitude towards the people around me who have supported, aided and guided me through these last four years.

First of all I have to thank my supervisor, Richard Brenner, who always takes his time to discuss and explain any ideas and problems. I am grateful to Camilla Rönqvist, who allowed me to write a Master thesis about DIXI and who later on has been my external supervisor.

The DIXI group has been a forum for a lot of discussions and planning of the development of DIXI. Many thanks to my comrade in arms Lilian del Risco Norrlid for all help and discussions. Thanks to Kjell Fransson, Sven Kullander and my co-supervisor Leif Gustafsson. A special thanks to Nils Bingefors who has helped me out a countless times in the lab and to Lars-Erik Lindquist for excellent and quick wire-bonding work.

Thanks to all PhD students who makes ISV a pleasant working environment. A special thank you to Mattias, Nils, Christian and Bjarte for your help concerning \LaTeX{} and Root during the writing of this thesis.

The graduate school AIM has not only payed my salary, but it has provided a forum for the exchange of ideas and knowledge amongst fellow graduate students.

Many thanks goes to Inger, Annica and AnnaLotta for the administrative support.

Finally, my wonderful wife Silvia, whose tender love, bright spirit and never-ending urge to travel with me to unknown destinations fills my heart with warm love. Thank you for always being here for me. Last, but not least, Emma and Adrian, who always show me the true happiness of life!
References


A doctoral dissertation from the Faculty of Science and Technology, Uppsala University, is usually a summary of a number of papers. A few copies of the complete dissertation are kept at major Swedish research libraries, while the summary alone is distributed internationally through the series Comprehensive Summaries of Uppsala Dissertations from the Faculty of Science and Technology. (Prior to October, 1993, the series was published under the title “Comprehensive Summaries of Uppsala Dissertations from the Faculty of Science”.)