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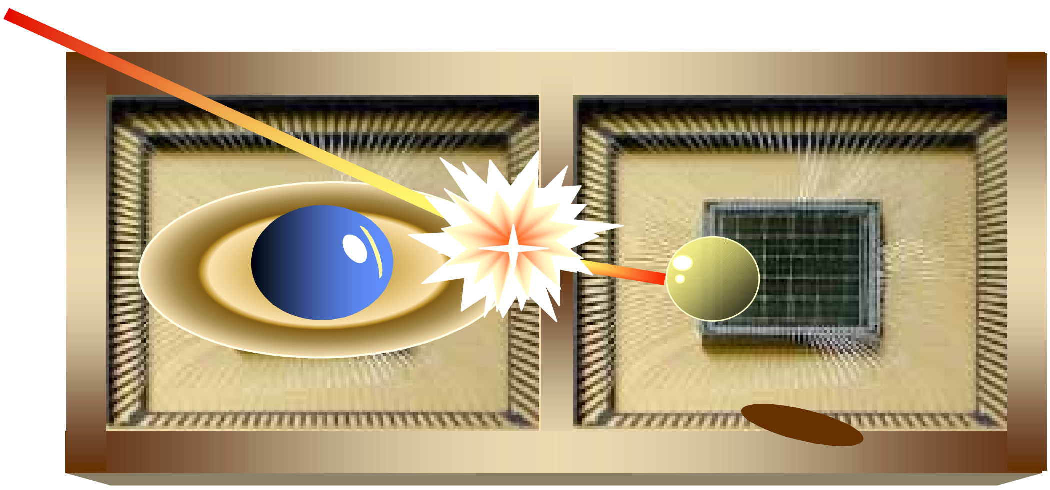
PIXEL DETECTORS AND ELECTRONICS FOR HIGH ENERGY RADIATION IMAGING

by

Munir A. Abdalla



Ph. D. Thesis

The Royal Institute of Technology
Department of Electronics, Solid State Electronics
Electrum 229, SE-164 40 Kista, Sweden

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Abstract

This work has been carried out to cope with the increasingly significant role of image sensors in today's life. The various applications of radiation imaging in biomedical applications, material science and high energy particle physics, activated tremendous research in new pixel detector materials and methods for image capture and acquisition.

In this thesis, various pixel detector types for high energy radiation are discussed with their advantages and weaknesses when suitability for specific applications are addressed. The different readout modes of detectors operations, and ways of detector/readout integration are demonstrated together with circuit techniques and design criteria.

In an effort to investigate new methods for detector/readout realization, several CMOS readout electronic chips for dental X-ray imaging have been prototyped to form hybrid sensors by: 1) bonding to semiconductor pixel sensors, and 2) coating with a scintillator. The later designs dealt with the different photo-sensing device in a standard CMOS process. The main emphasis was on the direct X-ray absorption and sensitivity of these devices. It was shown that the p-diffusion/n-well photodiode will exhibit the best imaging performance if its sensitivity is improved by an in-pixel preamplifier. Both methods proved the CMOS technology as an excellent replacement to the currently dominating charge-coupled devices (CCDs) technology.

Since the pixel design for single photon counting is driven mainly by small area and power consumption, we have devised different circuit techniques for that purpose. A new biasing method to facilitates the implementation of the feedback resistor, utilizing a MOS transistor, for the pixel-oriented preamplifier-shaper, was introduced. We also presented a new all-analog photon counting pixel concept wherein the digital part is replaced by a capacitor, which will significantly reduce the pixel area, and eliminate all the mixed-mode design overhead. Moreover, a design technique for the digital counter in a photon-counting pixel, to reduce the noise introduced by the switching activity of the digital part, has been introduced. Significant improvements in the performance due to these proposed circuit techniques were demonstrated by simulations.

A novel real-time ion beam profiling system for ion implanters using a pixellated graphite has been introduced. The measured results showed a promising performance that will have its high impact on the manufacturing yield of the ever increasing density of small devices per chip, by offering a better control on the implanters parameters.

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Munir A. Abdalla

Sundsvall, July 2001

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1- Papers included in the thesis

(1) ***M. A. Abdalla****, C. Fröjdh, H. Martijn, C. S. Petersson, “*Design of a CMOS Read-out Circuit for Dental X-ray Imaging”, Proc. of "The 6th IEEE International Conference on Electronics, Circuits and Systems (ICECS'99)", PAFOS, Cyprus - Sept. 5-8, 1999.

(2) ***M. A. Abdalla****, C. Fröjdh, C. S. Petersson, “* A CMOS APS for Dental X-ray Imaging Using Scintillating Sensors”, Nuclear Instruments and Methods-A, vol. 460, issue 1, pp197-203, March 2001.

(3) ***M. A. Abdalla****, C. Fröjdh, C. S. Petersson, “*An Integrating CMOS APS for X-ray Imaging with an In-Pixel Preamplifier”, Nuclear Instruments and Methods-A, vol.466, No.1, June-2001, pp.232-236.

(4) ***M. A. Abdalla****, E. Dubaric, C. Fröjdh, C. S. Petersson, “*A Scintillator-coated Phototransistor Pixel Sensor with Dark Current Cancellation”. Proceedings of "The 8th. IEEE International Conference on Electronics, Circuits and Systems (ICECS'2001)- Malta, Sept.2001.

(5) ***M. A. Abdalla****, C. Fröjdh, C. S. Petersson, “*A New Biasing Method for CMOS Preamplifier-Shapers”. Proc. of "The 7th. IEEE International Conference on Electronics, Circuits and Systems (ICECS'2K)", Lebanon, Dec.2000

(6) ***M. A. Abdalla****, C. Fröjdh, C. S. Petersson, “*An All-analog Time-walk Free SCA for Event Counting Pixel detectors”, Proc. of the 5th WSES/IEEE CSCC2001-Crete, July 2001. - Also in Electrical and Computer Engineering Series, ISBN 960-8052-39-4, pp.363-367 .

(7) Mattias O’Nils, ***M. A. Abdalla****, Bengt Oelmann, “*Low Digital Interference Counter for Photon Counting Pixel Detectors”, Submitted to Nuclear Instruments and Methods-A .

(8) J. Marchal, M. S. Passmore, ***M. A. Abdalla****, J. van den Berg, A. Nejim, C. Fröjdh, V. O'Shea, K. M. Smith, M. Rahman, “*Active Pixel Detector for Ion Beam Profiling”, 3rd. International Workshop on Radiation Imaging Detectors, Sardinia, 23-27 July- 2001, (for NIM-A)

2- Papers not included in the thesis

(9) ***M. A. Abdalla**, C. Fröjdh, C. S. Petersson, "*CMOS Pixel Electronics for New Sensors for Dental X-ray Imaging", Proc. 17th. IEEE Norchip Conference, Oslo- Norway, Nov. 1999.

(10) E. Dubaric, C. Fröjdh,M. Hjelm, H. E. Nilsson, ***M. A. Abdalla****, C. S. Petersson, "*Monte Carlo Simulations of the Imaging Properties of Scintillator Coated X-ray Pixel Detectors", The IEEE Nuclear Science Symposium and Medical Imaging Conference, Lyon- France, Oct. 2000.

(12) B. Oelmann, ***M. A. Abdalla****, M. O'Nils, "*An All-digital Window Discriminator for Photon Counting Pixel Detectors", IEE Electronic Letters, Vol.37, No.6, March 2001.

Abbreviations and Acronyms

APS	- active pixel sensor
CCD	- charge-coupled devices
CMOS	- complementary metal oxide semiconductor
CSA	- charge-sensitive amplifier
CTE	- charge transfer efficiency
CTF	- Contrast Transfer Function
CVD diamond	- chemical vapor deposited diamond
dB	- decibel
DEPFET	- depleted field-effect transistor
DR	- dynamic range
FPN	- fixed-pattern noise
FWHM	- full-width-at-half-maximum
GEM	- gas electron multiplier
LFSR	- linear feedback shift register
LPT, LPTGR	- latral phototansistor, latral phototansistor with guard rings
NMOS, PMOS	- n-channel, p-channel metal oxide semiconductor
MCP	- Microchannel plate
MOSFET	- metal oxide semiconductor field-effect transistor
MTF	- modulation transfer function
PET	- positron emission tomography
PMT	- photomultiplier tube
PSF	- point spread function
ROIC	- readout integrated circuit
RGCCD	- resistive gate charge-coupled device
SCA	- single channel analyzer
SNR	- signal-to-noise ratio
SPECT	- single photon emission computed tomography
STJ	- Superconducting Tunnel Junctions
VPT	- vertical phototransistor

1

INTRODUCTION

1.1. Thesis Background and Motivation

The increasingly significant role of radiation imaging in science and technology, with its numerous applications in medical imaging and high energy particle physics, have lead to extensive research for developing new pixel detector materials as well as adequate technologies for image capture and acquisition. The need to replace film radiography by filmless approaches is widely acknowledged in the medical community. Replacing an all purpose medium used for image acquisition, storage and display, with a technology optimizing each task will result in productivity improvements in radiology. Real time imaging, elimination of consumables (film, chemicals) and tasks (film handling), facilitation of image display, archiving and transfer are some of the advantages.

For the past decades, charged coupled devices (CCDs) have been unequal leader in the field of electronic image sensors for all kinds of applications. This has been driven by the market demand for ever larger pixel numbers and better image quality. However, interest in image sensors based on CMOS technology has increased dramatically in the past ten years. CMOS based imagers offer significant advantages over CCDs such as system-on-chip capability, low power consumption and possibly lower cost.

Intensive research in new generations of pixel detectors, as well as adequate readout methods to achieve their best imaging performances, is rapidly developing. The combination of different radiation detection methods and image capturing techniques demand adaptation of the various technologies of detectors processing and readout electronics. The choice between integrating type and single photon counting readout modes is an important decision. Creating design techniques for the pixel circuitries and devising methods to achieve good image properties, such like high resolution and low noise, are typical challenges, to name a few.

In this work we have explored different methods of using new sensors for digital dental X-ray imaging utilizing a standard CMOS technology. We also developed a novel system for real-time ion beam profiling for ion implanters. In addition, new methods have been developed for improving the single photon counting pixel design. Both analog and digital circuits have been introduced. The articles (PAPERS 1 - 8) appended to this thesis which describe the author's contributions in the various pixel detector/readout electronics field is demonstrated in Fig.1.

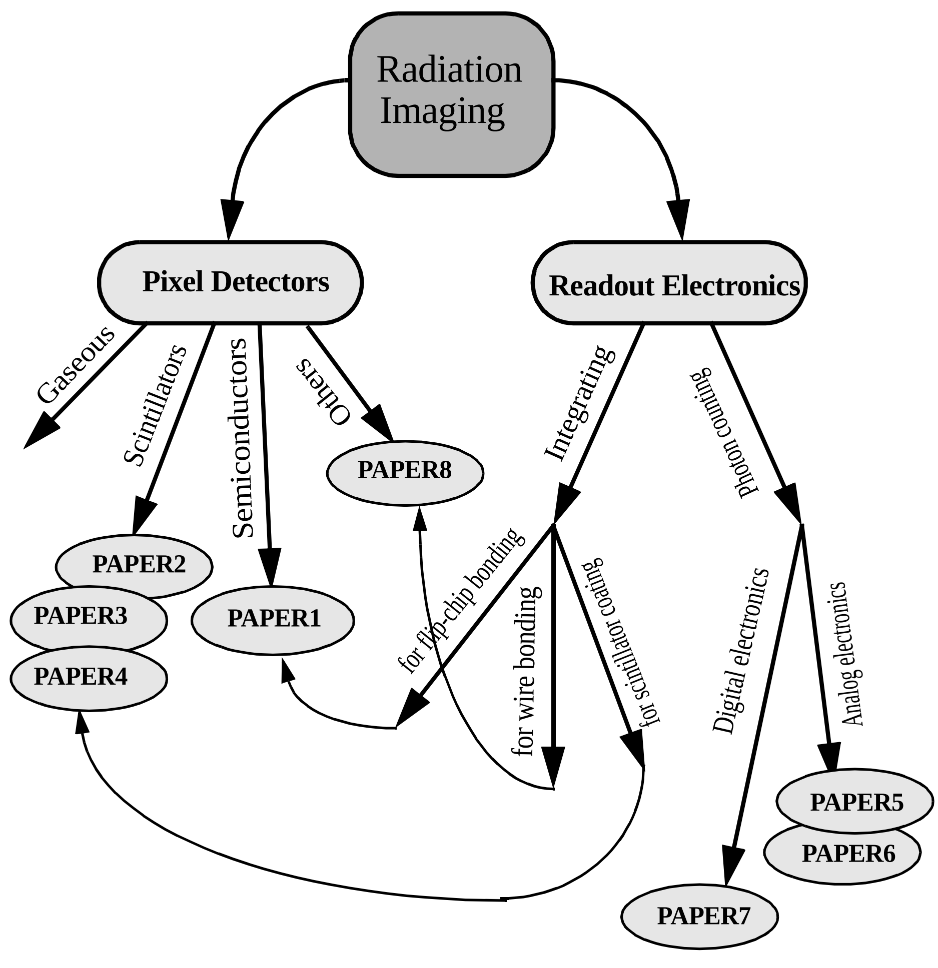

Fig.1: A general thesis overview showing the paper contributions.

1.2. Radiation Imaging

Radiation is energy in transit in the form of high-speed particles and electromagnetic waves. We encounter electromagnetic waves every day. They make up our visible light, radio and television waves, ultra violet, and microwaves with a spectrum of energies. These examples of electromagnetic waves do not cause ionization of atoms because they do not carry enough energy to separate molecules or remove electrons from atoms.

Ionizing radiation

Ionizing radiation is radiation with enough energy so that during an interaction with an atom, it can remove tightly bound electrons from their orbits, causing the atom to become charged or ionized. Examples are gamma rays and beta particles. If the energy of the incident photon/or particle is not high enough to eject an electron from the atom, but is used to raise the electron to a higher energy level, the process is termed excitation.

Non-ionizing radiation

Non-ionizing radiation is radiation without enough energy to remove tightly bound electrons from their orbits around atoms. Examples are microwaves and visible light.

The field of radiation imaging includes a variety of applications ranging from one-dimensional (1D) to two-dimensional (2D) and three-dimensional (3D) imaging. Unlike the usual imaging, the penetration power of high energy radiation enables deep level imaging of objects rather than only surface imaging. One-dimensional images covers all spectroscopy and profiling applications. On the other hand, 2D images are captured by the use of arrays of pixel sensors or scanning methods. 3D images could be obtained by various methods including image processing of 2D captures.

1.3. Medical imaging

The goal of medical imaging is to provide a spatial mapping of some parameter, feature, or process within a biological entity. Generally speaking, two broad categories of medical imaging systems exist: those that provide anatomical information and those that produce a functional mapping of the object under observation. There are two basic ways of performing medical imaging: transmission and emission imaging. In the first type, there are basically three elements: a source of X-rays or gamma-rays, the body of the patient and a detector (which can be a film, a semiconductor detector, a scintillator or a wire chamber). In the case of emission imaging, the source emits inside the body

(i) Gamma-ray imaging

In gamma-ray imaging, a drug labeled with a gamma-ray emitting isotope is injected in negligible amounts into the patient. The drug is chosen according to the metabolism of the organ under study and is detected owing to its high specific activity. The information desired is the spatial distribution of the source within the body. The main goal is the physiological studies of tissues by using specific radioactive tracers. Basically, the number of photons detected by a given detector is proportional to a weighted integral of the activity contained in the region it sees. There are two variants to this kind of imaging: PET (positron emission tomography) and SPECT (single photon emission computed tomography). PET is based on the detection of back-to-back 512 keV photons from the annihilation of positrons (emitted by the drug injected into the patient) with electrons in the neighboring tissue. In SPECT, the detection is sensitive to the direct emission of a photon by a radioisotope in the drug injected into the patient. Usage of semiconductor pixel detectors for gamma-ray imaging in nuclear medicine is discussed by H. B. Barber et. al. in [1]

(ii) X-ray imaging

X rays are electromagnetic radiation emitted by an atom when it rearranges its orbital electrons after the creation of a hole in one of its deeper shells. The origin of this hole or vacancy can be the capture of an electron, an internal conversion in an atom, the effect of ion or electron bombardment, the result of the photoelectric effect or X-ray fluorescence. X-rays produced by medical X-ray tubes are in fact bremsstrahlung produced by the slowing down of electrons emitted by a cathode ray tube. They have a continuous energy distribution, while atomic X-rays are characterized by well defined energies.

X-ray imaging is based on X-ray attenuation by the human body. The patient is illuminated with an X-ray beam from an X-ray tube, and an image of the absorption of parts of the body with different densities is taken. It assumes mono-exponential decay of the monochromatic beam, and bremsstrahlung and k-lines are filtered to improve beam uniformity and narrow the spectrum. The energy range used is from around 10 keV (mammography) up to 70 keV (dental and chest radiography). For radiological examinations, higher X-ray energies than those mentioned would be preferred, in order to reduce the skin dose, but then the soft tissue contrast would be reduced as well. Thus, the working energy is chosen in relation to the type and structure of the object/organ to be imaged

(iii) X-ray detection

The two basic techniques used for X ray detection are direct and indirect detection. In direct X-ray detection, the detector converts the absorbed X-ray directly into a charge signal. For energies below 100 keV, this is possible if the detection medium has good absorption efficiency, and it is enhanced if the atomic number (Z) of the material used is high. For this range of energy, the dominant interaction is the photoelectric effect. Absorption efficiency for film and silicon is poor for energies above 20 keV, therefore a converting medium (typically scintillator) is used between the contrasting detail and the detector (indirect detection). This extra step deteriorates the spatial resolution of the system. The effects of scattering, however, can be minimized by the use of a collimator or a scanning system. The basic operation modes for X-ray detection are: integration mode, counting mode and Compton scattering. In integration mode, the total charge released by the incident radiation is accumulated during the exposure time. In counting mode, each photon is counted independent of energy. In Compton scattering, the position of the emitted photon is defined by back-projection reconstruction. In order to do that, it uses two detecting planes. The first one (closer to the source) is designed so that Compton scattering is the dominant interaction process, while the second one is designed to completely absorb the photons. From the two position measurements and the angle of scatter, it is possible to back-project to localize the photon direction within a cone determined by the measurements.

For a more detailed study of radiation interaction with matter, radiation detection and measurements we refer the reader to Knoll [2].

1.4. Types of Pixel Detectors

Pixel detectors are very appealing for high-energy physics and biomedical imaging. Although the motivations for their choice are different, most of the problems are common [3].

1.4.1. Semiconductor Pixel Detectors

Semiconductor pixel detectors were initially developed for high energy physics applications because of their low noise, high granularity, stand-alone pattern recognition capabilities, good spatial resolution and true two-dimensional position information. Because of their faster collection times, they are able to process higher counting rates in the pulse mode of operation. They can provide high X-ray detection efficiency, relative to gas-filled detectors or conventional X-ray film, for X-ray energies of interest in medical imaging. These pixel detectors are commonly encountered in two broad varieties; CCDs and Active Pixel Sensors (APS). They can be either monolithic devices, or hybrid detectors. When it is impossible to fabricate the detector and read-out electronics on the same wafer, the detector and Read-Out Integrated Circuit (ROIC) are fabricated on separate wafers and then connected electrically by flip-chip bonding. The hybridization has the advantage of allowing the separate optimization of the detector and ROIC, which also provides greater flexibility in the choice of active detection media. However, Hybrdization of pixel detector systems has to satisfy tight requirements: High yield, long term reliability, mechanical stability, thermal compilance and robustness have to go together with low passive mass added to the system, radiation hardness, flexibility in the technology and eventually low cost. The current technologies for the interconnection of electronics chips and sensors are reviewed and compared in [88]. A typical hybrid pixel sensor utilizing flip-chip bonding is demonstrated in Fig.2.

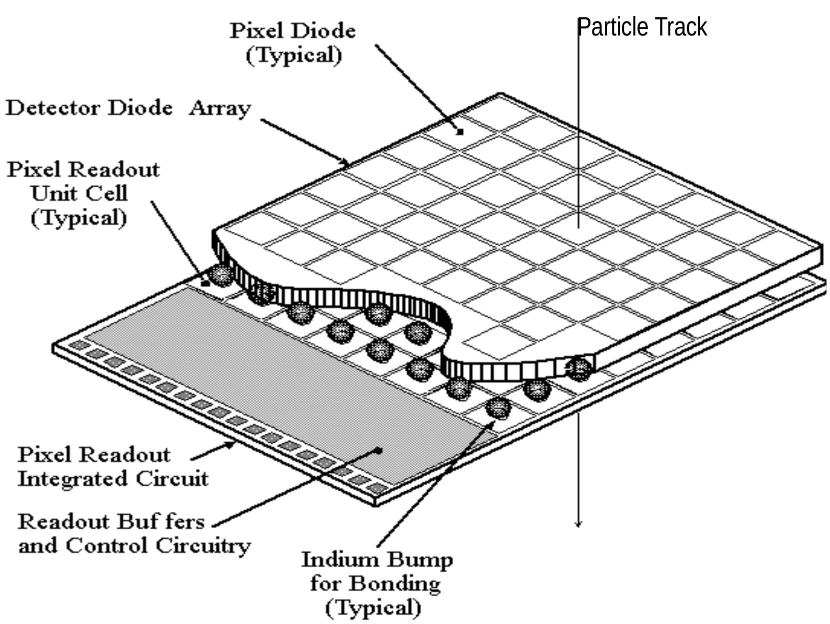

Fig.2: Hybrid pixel sensor

(i) Si detectors: Digital imaging has largely been based on silicon charge-coupled devices, CCD technology, for the past decades. This technology, however, has several weaknesses [4]. A major weakness in using CCDs in radiation imaging is the problem with their charge transfer efficiency (CTE), which makes them very sensitive to defective pixels. This is important in imaging devices that are exposed to energetic particles, such as X-ray photons, which are able to damage the pixels. Apart from the CCD technology, silicon X-ray pixel detectors chips can be bump-bonded to pixel arrays of readout electronics. Reasonable detector thicknesses of 300μm - 500μm give very good efficiency up to 12KeV [5][6][7]. On the other hand, amorphous silicon pixel arrays are becoming an important tool for radiotherapy and diagnostic imaging. They are radiation hard, relatively inexpensive to manufacture, and can be as large as 0.4 m on a side [8]. Moreover, depleted field-effect-transistors (DEPFETs) has been used in radiation imaging, and the first operation of a pixel imaging matrix based on DEPFET pixels was introduced by P. Fischer et. al. [9]

(ii) GaAs detectors: The development of GaAs X-ray imaging detectors is focussed on the application areas of synchrotron X-ray imaging and diagnostic medical X-ray imaging.These applications use X-rays in the energy range of approximately 15 - 60 keV which is particularly well suited to the detection efficiency of GaAs. The photoelectric absorption efficiency of GaAs in the photon energy region of interest is significantly better than that of silicon (Fig.3).

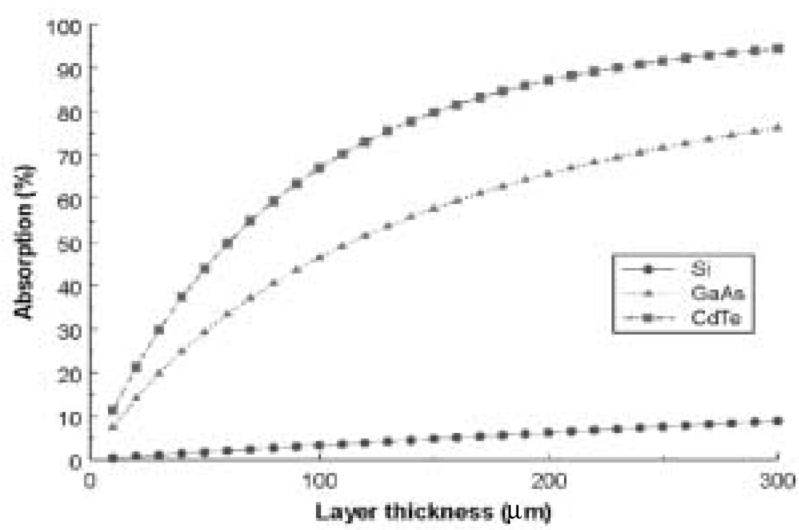

Fig.3: The detection efficiency of GaAs and Si versus X-ray energy for different thicknesses [36]

The use of undoped semi-insulating GaAs is required in order to obtain detectors with thick active depths -typically wafers of between 150μm and 500μm thickness are used. This "industry standard" material is readily available at low cost and is well suited to standard photolithographic processing techniques. GaAs pixel detectors are used in different ways for detecting X-rays. It is implemented as high speed CCDs since its electron mobility is high [10]; X-ray imaging systems based on GaAs-CCDs implemented as the so-called resistive gate CCD (RGCCD) has been developed [11][12]. A hybrid GaAs detector chip can be flip-chip bump-bonded to a silicon CMOS readout circuit. More about developments in GaAs pixel detectors for X-ray imaging is reported in [13].

(iii) CdZnTe detectors: The semiconductor CdZnTe was originally developed as a "room temperature" spectrometer and it has been proposed for use as a gamma camera in nuclear medicine. It has several properties that make it potentially useful for digital mammography. It has a high density (5.8 g/cm3) and a high atomic number which provides excellent absorption efficiency even for very thin detectors (98% at 20 keV for 0.4 mm thickness). This material has a high resistivity (10-100 gigaohm.cm) which provides reasonably low dark currents. It also has a high signal gain (approx. 4000 electron-hole pairs for a 20 keV photon) which provides an excellent signal to noise ratio. One design limitation is imposed by the relatively small hole mobility in CZT. This requires that the CZT be as thin as possible, and biased correctly to provide the shortest distance for the holes to travel. Imaging applications of CdZnTe pixel detectors are reported in [14][15].

(iv) SiC detectors: The limitations of silicon and germanium have promoted studies on the properties of silicon carbide (SiC) as a semiconductor material for radiation detection. Because of its higher band gap energy and greater radiation resistance, SiC should theoretically lead to a detector capable of operating at elevated temperatures and in high radiation fields. The properties of SiC radiation detectors have been tested for both Schottky and p-n junction devices [16]. Good detection properties was measured for alpha particles without external bias voltage.

In general, the properties required for a semiconductor detector material are:

a) Small energy gap to give a large yield of electron-hole pair from the nuclear particle.

b) Low quiescent carrier concentration to give low leakage current.

c) high mobilities of holes and electrons and long carrier lifetime to give efficient collection and rapid rise time of the signal.

d) freedom from delayed trapping to give a rapid rise time and freedom from space charge effects.

e) High atomic number to give a good photoelectric cross-section.

1.4.2. Scintillating Detectors

Inorganic scintillators are conventional detectors for gamma-ray radiation (above 10 keV) measurements. Detailed description of these instruments can be found e.g. in Knoll [2] and Nicholson [17]. The basic mechanism lies in measuring the scintillation light produced by an ionizing high energy photon (or alpha or beta particle) when it interacts with the scintillating material. The most commonly used inorganic scintillators are the activated alkali-alidi crystals NaI(Tl) and CsI(Na or Tl). Conventionally, the element used to activate the crystal is indicated between parenthesis. A gamma-ray photon arriving on the detector deposits all or part of its energy in the material in the form of kinetic energy of one or more electrons, depending on the type and number of interactions. These electrons are able to excite to the conduction band other electrons which can be captured by a trace impurity (the activator) and cause transitions leading to the emission of visible light. The role of the activator is to generate meta-states between the pure crystal valence and conduction bands, so that an electron excited to the conduction band can drop in one of this meta-states and de-excite from it to the valence band. This has the advantages of being a more efficient mechanism with respect to the normal de-excitation from crystal conduction band and to lead to the emission of visible light photons, because of the lower energy of meta-states with respect to the conduction band. Traditionally, the scintillation light pulse is then collected through a light pipe (typically a quartz pipe) to a Photomultiplier Tube (PMT), which finally converts it to an electric signal to be amplified and measured. Scintillating materials can also be hybridized with a semiconductor photosensor for radiation imaging where it is used as a coating layer on a pixel matrix to convert the incident radiation into light, which is then detected in the semiconductor pixel. The coating layer itself can be pixellated to achieve better image resolution. An X-ray imaging pixel detector based on scintillator filled pores in a silicon matrix has been reported in [18].

The relevant factors that contribute to the detector performances are:

a) crystal transparency to its own emitted radiation.

b) The number of photons emitted (*light yield*, usually measured in photons/keV) for each detected event and its proportionality to the incident photon energy (*linearity*). The maximum efficiency in converting initial electron energy into photons is 13% for NaI(Tl). The NaI and CsI light yields are not exactly linear with energy, decreasing with increasing incident photon energy and are affected by temperature changes.

c) decay time of the induced scintillation, determined by the decay time of the meta-state involved in the scintillation, limiting the detector temporal resolution;

d) scintillation light collection efficiency, which depends on the geometry of the detector, on crystal coating and on the position in the detector in which the interaction with the incident photon takes place;

e) crystal/photosensor coupling.

f) photosensor quantum efficiency and gain.

1.4.3. Gaseous Detectors

Gaseous detectors are widely used in modern tomographic X-ray scanners for medical applications [19]. Multiwire proportional counters have been extensively and successfully used as position sensitive X-ray detectors in a wide range of applications. The development of the microstrip gas chamber in 1988 [20] led to higher geometrical accuracy together with narrower anode pitches. The invention of the gas hole counters provided very high gas gains and mechanical stability [21]. The basic idea of hole counters is to focus the field lines from a drift region into holes where a strong electric field (typically 10 -100kV/cm) induces the condition for gas amplification.

The gas electron multiplier (GEM) consists of a thin, metal-clad polymer foil, chemically pierced by a high density of holes. On application of a difference of potential between the two electrodes, electrons released by radiation in the gas on one side of the structure drift into the holes, multiply and transfer to a collection region. The multiplier can be used as detector on its own, or as a preamplifier in a multiple structure; in this case, it permits to reach large overall gains in harsh radiation environment. Systematic studies of single, double and triple-GEM detectors exposed to high radiation fluxes and to heavily ionizing radiation have been made. Large size detectors have been developed for the detection and localization of charged particle trackers for COMPASS[1], a high rate experiment at CERN[2].

Gaseous single photon counters are excellent candidates for X-ray imaging applications like small angle scattering, protein crystallography and medical radiography [22]. The recently developed micro-pattern- detectors (CAT, MICROMEGAS, GEM, microCAT, etc.) are a very promising enrichment for applications with gaseous x-ray detectors. The so-called microCAT detector [23] satisfies most of the requirements for X-ray imaging applications: stable operation at high gas gains, good imaging performance when combined with an adequate readout structure and high reliability and robustness. A study of high rate performance microCAT detector is published in [24]

1.4.4. Superconducting Pixel Detector

Superconducting Tunnel Junctions (STJs) are promising tools for simultaneous imaging and spectroscopy in a very broad range of the electromagnetic spectrum, covering the infra-red to X-ray wavelengths [25][26]. A STJ consists of two superconducting layers sandwiching a tunneling barrier. One of the important applications of an STJ is as an energy-sensitive radiation detector [27]. A higher energy resolution than that with conventional semiconductor detectors can be achieved, because the gap energy of the superconductor is as small as meV, and the statistical fluctuation of the number of quasi particles is also small [28]. Another advantage in STJ-based radiation detector application is the rapid response and radiation hardness. The hybrid superconducting pixel detector principle has been experimented by V. G. Palmieri. The device consisted of a Si detector bonded by ultra-thin Al wire to a high quality Nb/AlOc/Nb Josephson Tunnel Junction acting as current sensitive discriminator. A minimum ionizing particle detection capability was obtained [29].

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1. COMPASS is a high-energy physics experiment under construction at the Super Proton Synchrotron (SPS) at CERN in Geneva, Switzerland. The purpose of this experiment is the study of hadron structure and hadron spectroscopy with high intensity muon and hadron beams.
2. European Laboratory for Particle Physics (CERN), Geneva

1.4.5. Others

Various other types of high energy radiation pixel detectors are in use. However, for high-energy physics experiments the demands on the detector material are numerous. Radiation hardness, charge collection sensitivity and low leakage current are some examples of such demands

(i) CVD diamond:

An important parameter of a material that determines many characteristics for particle detection is the band gap between the valance band and conduction band. The band gap of chemical vapor deposited (CVD) diamond is 5.47eV, which is about five times that of silicon (1.12eV). As a consequence there are very few free charge carriers present in CVD diamond at room temperature, the sensitivity is very high and the leakage currents are very small. Therefore, diamond detectors need not be depleted and thus no diode structure is necessary as in silicon detectors. Diamond is a nearly ideal material for detecting ionizing radiation. Its outstanding radiation hardness, fast charge collection and low leakage current allow it to be used in high radiation environments. However, its stopping power for X-rays is rather low. These characteristics make diamond sensors particularly appealing for use in the next generation of pixel detectors. The first diamond pixel detector was tested in August 1996 in a particle beam at CERN [30][31]. Results from diamond detector systems and the status of diamond particle detectors are reported in [32][33][34], and the first bump-bonded pixel detectors on CVD diamond is published in [35].

(ii) The ceramic screen printed HgI_2 detectors:

These can operate successfully as nuclear particle counters, which are radiation resistant and can also be used for imaging, where no energy resolution is needed. The charge transport properties, i.e., $\mu\tau$ for electrons for poly-crystalline screen printed HgI_2 is at present 10^{-7} cm^2/V, as compared to 10^{-6} cm^2/V for diamond, or 10^{-5} for a-Si. But the latter have much smaller signals due to their much larger value of electron-hole formation energies, which are 14 and 50 eV for diamond and a-Si respectively [36], as compared to 4.2 eV for HgI_2. Because of their polycrystallinity, detectors can be potentially fabricated in any size and shape, using standard ceramic technology equipment, which is an attractive feature where low cost and large area applications are needed. A screen-printed strip detector with 275μm pitch and with about 135μm gap between two conducting strips has been connected to VLSI single particle detection electronics and tested in a high-energy beam at CERN. It was shown that for large area imaging device with direct detection, 100% fill factor could be obtained. A large area (1 in^2) pixel detector has also been measured showing good uniformity in detection response over the whole area. Ceramic mercuric iodide can thus be considered as a good candidate for large area imaging applications [37].

(iii) Microchannel plates (MCP):

MCPs are widely used in scientific research, space equipment, navigation equipment and so on [38]. Govyadinov, et. al. have proposed a new material (porous anodic aluminum oxide) for MCP production where it is theoretically possible to place up to 10^{10} channels per cm^2 in the plates with a ratio of channel length versus diameter in the range of 20 - 300, which is not achievable by conventional MCP technology. Anodic aluminum oxide is diamagnetically weak, making it suitable for strong magnetic fields, which is very important for applications in high-energy physics. Tests of radiation hardness of aluminum oxide showed that it could be used even inside the nuclear reactor. There are no fundamental factors limiting the size of aluminum oxide MCP and it is quite possible to produce MCP with a size up to 50x50 cm^2 [39].

(iv) Pixellated graphite detectors:

The high density graphite is suitable as a detection material in ion implanters because of its low sputtering yield and high thermal and electrical conductivity. A novel application is described in PAPER-8 appended to this thesis. In this article, a pixel detector array consists of pixellated graphite which is wire-bonded to a remote ASIC. The pixel diameter is 8mm, the backplane is 50mm by 50mm in size and the side-walls are 10mm thick. However, commercial implanters need smaller pixels (7mm) and larger detector area (20cm x 20cm).

The new trends in the design and assembly of an integrated system made of pixel sensors as sensitive elements combined with readout electronics and connecting cables or alternative connecting structures, in biomedical and high energy physics, are summarized and critically evaluated in [40].

2

CMOS APS READOUT

CMOS based imagers are beginning to compete against CCDs in many areas of the consumers market because of their system-on-chip capability. Sensitivity, however, is a main weakness of CMOS imagers and enhancements and deviations from the standard CMOS process are necessary to keep up sensitivity with downscaled process generations [41][42]. There is a fundamental distinction between the two most common modes of detector operation when used in radiation imaging. These are the integrating mode, and the photon counting mode discussed below.

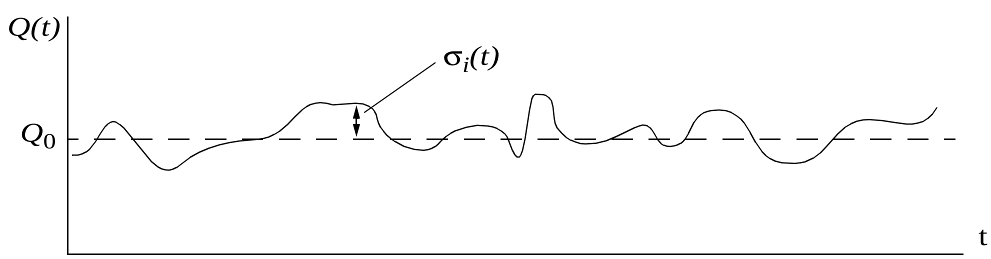
2.1. Charge Integration Mode

In readout circuits employing the integrating mode, the charge generated from the radiation interaction with the detector during exposure time is integrated on an in-pixel capacitor for a later readout. The average charge (Q_0) is given by the product of the average event rate and the charge produced per event, the event being a photon hit on the detector.

$$Q_0 = rQ = r\frac{E}{W}q \qquad (1)$$

where : r = event rate, $Q = Eq/W$ = charge produced by each event, E = average energy deposited by event, W = average energy required to produce a unit charge pair, q =1.6 x 10^{-19} C.

For steady-state irradiation of the detector, this average charge can also be rewritten as the sum of constant charge Q_0 and a time-dependent fluctuating component $\sigma_i(t)$, as shown below.



where $\sigma_i(t)$ is a random time-dependent variable that occurs as a consequence of the random nature of the radiation events interacting with the detector. The statistical measure of this random component is the variance or mean square value, defined by as the time average of the square of the difference between the fluctuating charge $Q(t)$ and the average charge Q_0, given by:

$$\overline{\sigma_I^{\,2}(t)} = \frac{1}{T}\int_{(t-T)}^{t}[Q(t') - Q_0]^2 dt' = \frac{1}{T}\int_{(t-T)}^{t}\sigma_i^{\,2}(t')dt' \qquad (2)$$

and the standard deviation follows as

$$\overline{\sigma_I(t)} = \sqrt{\overline{\sigma_I^{\,2}(t)}} \qquad (3)$$

For a Poisson statistics the standard deviation for in the number of recorded events n over a given observation period is expected to be

$$\sigma_n = \sqrt{n} \qquad (4)$$

Therefore, the standard deviation in the number of events occurring at a rate r in an effective measurement time T is given by

$$\sigma_n = \sqrt{rT} \qquad (5)$$

Since the output signal (S) in a measurement system is mainly related to the number of events, n, it follows that the signal to noise ratio (SNR) will be given by

$$SNR = \frac{S}{\sigma_n} = \frac{kn}{k\sqrt{n}} = \sqrt{n} \qquad (6)$$

where k is a constant representing the signal generated by each event.

Therefore, it is clear that for a good noise performance of an imaging system, high numbers of recorded events give higher SNR. This implies that both the exposure time, as well as radiation detection and collection efficiencies, are the key factors for high quality imaging system performance. This mode of operation does not preserve any information about the incoming event (photon) as far as energy and timing are concerned. It is most suitable when the count rates are high. In imaging applications, the consequent result of the simplicity of the required pixel electronics, small pixel sizes and hence high resolution imaging is achievable. Various pixel topologies can be implemented, and the simplest one consists of a single transistor and a storage capacitor as in

Fig.4a. However, this topology is a destructive readout since the information (charge stored in the capacitor) is lost after readout.

Usually a non-destructive readout topology is employed wherein a buffer and reset transistors are included in the pixel circuit. Thus the charge on the storage capacitor is always available for multiple readouts. A typical electronic topology of an integrating nondestructive pixel, together with associated electronics is shown in Fig.4b, and various integrating pixel topologies are also revised in [43]. The circuits in Fig.4 show a pixel circuit that is bump-bonded to an external pixel detector by an indium ball. Wire-bonding could also be used when remote pixel sensors are used. However, if a scintillating layer is to be used, the diode D can be employed as a photosensor.

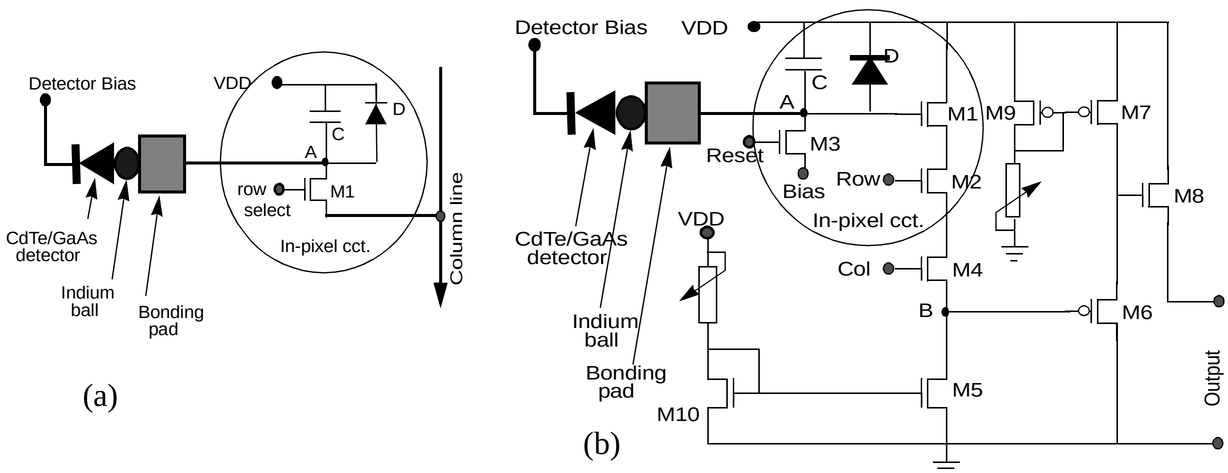


Fig.4: Schematic diagram of an integrating pixel and associated electronics. a) a distructive readout pixel, and b) non-destructive readout.

2.2. Photon Counting Mode

Photon counting mode of radiation detectors operation is more common in use than the integrating mode because of several inherent advantages. First, the sensitivity that is achievable is often many times greater than the integration mode because each individual quantum of radiation can be detected as a distinct pulse. Thus, the lower levels of detectability are set by the background radiation levels. In contrast, in charge integration mode the minimum detectable charge may represent an average interaction rate in the detector that is many times greater. The second and more important advantage is that each pulse amplitude carries some information that is often useful or even necessary part of a particular application. Most applications are better served by presenting information on the amplitude and timing of individual events that only photon counting mode can provide. This feature in the photon counting mode means energy resolution capability through pulse height analysis, which is the key factor in nuclear spectroscopy. A very important advantage of photon counting readout is the readiness of data directly in digital form which is needed for subsequent computational image processing. However, the main disadvantage is the complexity of the readout electronics require. The pixel electronics contain all the analog and digital pulse processing components that can sum up to hundreds of transistors per pixel. In Fig.5 a block diagram of a typical pixel array and a sample photon counting pixel circuit are sketched.

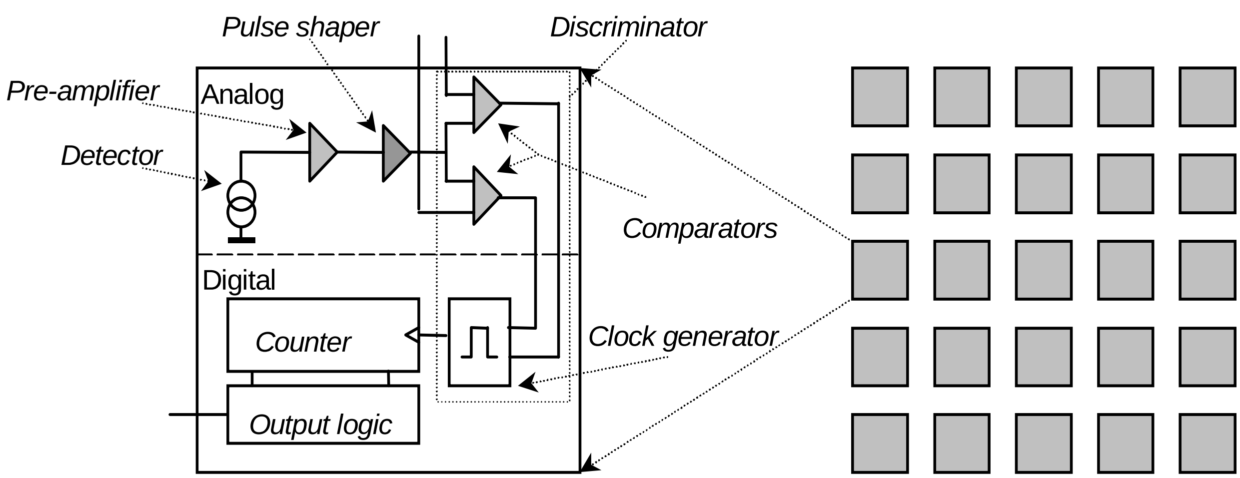


Fig.5: Pixel array and a pixel block diagram for photon counting image sensor.

3

INTEGRATING TYPE PIXEL SENSORS

Pixel design is the fundamental factor in the design of image sensor. In digital X-ray radiation imagers implemented in CMOS technology two methods are used for the sensor implementation. A pronounced example for digital imaging of soft X-ray, employing integrating readout mode, has recently been worked on in the XIMAGE[1] project, where various detector materials and sensor/ readout electronics hybrid designs were investigated [44][45]. The first method uses heavy atoms semiconductor pixel detectors with high stopping power to X-rays hybridized to CMOS read-out chips. [46][47]. The other method uses scintillator coated silicon detectors [48]. However, the first integrating, scintillator coated CMOS chip used for digital dental X-ray imaging has been introduced in [49].

In this chapter, we discuss design criteria for integrating pixel sensors. The emphasis, however, will be on work that has been done within the framework of XIMAGE and other projects.

3.1. Scintillator-coated X-ray Active Pixel Sensor Design

Here the pixel structure consists of the scintillating layer which converts the incident X-rays into visible light. The photo-diode or photo-transistor in the silicon pixel then generates an electrical charge corresponding to the scintillator output light. Though the type of photo-detector that has widely been used in this method is implemented in CCD technology, the discussion in this section is focused on the currently emerging active pixel sensors (APS) in CMOS technology.

1. XIMAGE is a BriteEuRam project for research on pixel detectors for X-ray imaging

However, photo-detectors in other technologies can be employed. As an example, the performance of a novel high gain photo-detector on Silicon-On-Insulator (SOI) technology that could serve the purpose can be considered [50].

There are many factors that will determine the image quality of this type of image sensors. The choice of the scintillating material that will output the highest light quantity is the first part of the pixel sensor. Various scintillators with different light output capabilities are available [51]. On the other hand, pixel structures in CMOS technology can be based on various intrinsic photo-sensing elements that differ in their performance characteristics. A certain photo-sensing element is to be chosen depending on its optical sensitivity and noise performance. The imaging performance of the chosen photo-sensing type depends on the dynamic range (DR), signal-to-noise ratio (SNR) and spatial resolution which determine the pixel size. Since these imaging criteria are contradicting each other when deciding the pixel size, a fundamental trade-off must be made to select the pixel size. A large pixel size is desirable because it results in higher DR and SNR, while a smaller pixel size results in higher spatial resolution.

The various design criteria to reach an optimal integrating X-ray image sensor implemented in CMOS technology with a scintillation detector coating are described below.

3.1.1. Scintillator Choice

The choice of the scintillating layer is the first step in specifying the subsequent design strategy. It is determined by the quantity and wavelength of the light emitted by the scintillator. When using scintillators there is a trade-off between sensitivity and spatial resolution. In order to get a higher light output the layer thickness should be increased, but an increased layer thickness reduces the spatial resolution because of light diffusion in the scintillator. The absorption of X-rays for a number of known scintillators is shown in Fig.6. For these materials a thickness of 150-300μm is required to absorb 80% of the radiation from a standard dental X-ray unit (up to 70 keV). This thickness is significantly larger than the normal pixel size for dental X-ray imaging systems (40-50μm), which results in a significant degradation in image resolution. The spatial resolution can be improved by defining pixels in the scintillator [18][52]. The most commercially used scintillators are GdO_2S_2 (GADOX) and CsI. However, LuPo4:Eu is an interesting alternative to the currently used scintillators. The absorption is comparable to CsI and GADOX and the light output, for a 700nm, was found to be twice the light output from CsI [53].

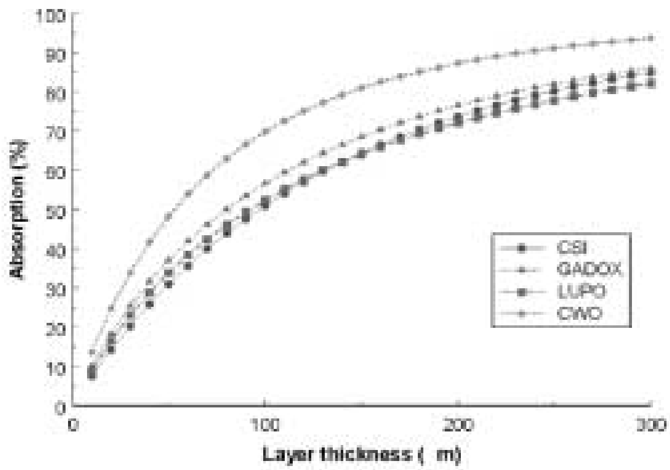


Fig.6: X-ray absorption in a number of different scintillators as a function of layer thickness when illuminated from a dental X-ray source operated at 60kVp [36]

3.1.2. Photosensor Choice

Both photo-diodes and photo-transistors can be used in detecting the light output from the scintillating layer. Photo-diodes offer excellent packing density and low noise performance, but have lower sensitivity than photo-transistors, typically by a factor of β (the transistor gain). Typical photo-diode junctions in an n-well CMOS process are shown in Fig.7.

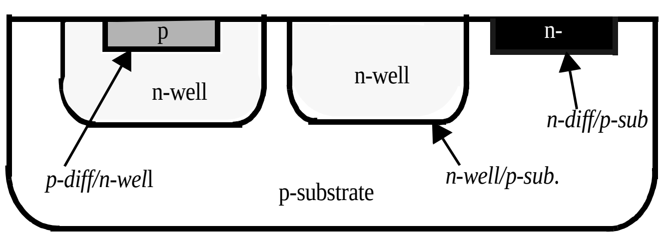


Fig.7: Typical photodiode junctions in an n-well CMOS process.

Several methods can be used to realize photo-diodes with independent spectral responses in a standard CMOS process utilizing only the masks, materials, and fabrication steps. The spectral responses can be controlled by [54]: 1) using the SiO_2 and polycrystalline Si as thin-film optical filters., 2) using photo-diodes with different junction depths, and 3) controlling the density of the interfacial trapping centres by choosing which oxide forms the Si/SiO_2 interface.

Depending on the scintillator used, emitted wavelength, the above mentioned methods can be used to design a photo-diode-based pixel. The selection of the method used can be estimated from the equation that estimates the number of absorbed light photons below:

$$F_{abs}(\lambda) = F_o(\lambda)(1 - R(\lambda))(1 - e^{-\alpha(\lambda)x}) \qquad (7)$$

where

$F_{abs}(\lambda)$ = number of absorbed photons.

$F_o(\lambda)$ = number of photons incident on the Si surface;

$R(\lambda)$ = fraction of photons reflected from the surface;

$\alpha(\lambda)$ = absorption coefficient;

λ = wavelength of the incident light;

x = depth into the photosensor.

According to the above equation, the p-diffusion/n-well photo-diode in an n-well CMOS process exhibits a fairly low quantum efficiency because of its low sensitive depth compared to the n-well/p-substrate photodiode. However, since the fundamental structure of the n-well/p-substrate photo-diode has the anode connected directly to the substrate, it is directly subject to noise pick-up and cross-talk due to large drift field in the bulk. These drawbacks are not present in the p-diff/n-well photo-diode. On the contrary, the p-diff/n-well photo-diode shows a very low sensitivity to direct X-rays which means a great advantage over all the photo-detectors in a standard CMOS process as far as SNR is concerned. Moreover, the shallow sensitive depth makes the quantum efficiency peaks in the shorter wavelength region (the blue light). This property is advantageous if an appropriate scintillator is used since it will act like an optical filter that enhances the dynamic range. However, its low optical sensitivity makes it necessary to implement an in-pixel signal preamplification if it is used in an efficient X-ray imaging [55]. Our experiments with the different photo-sensors in CMOS pixels showed a good performance of the p-diff./n-well photo-diode as far as direct X-ray absorption and sensitivity are concerned. The experimental results are described in PAPER2 and PAPER3 appended to this thesis.

On the other hand, several photo-transistor structures can be produced in a CMOS process [56][57]. Except for the p-n junction photo-diode, the vertical photo-transistor is the simplest photo-detector device to implement in a standard CMOS process (Fig.8a). A lateral pnp photo-transistor structure (LPT) as shown in Fig.8b is fabricated by surrounding the p-emitter diffusion with a p-collector ring, all situated in an n-well base. A lateral pnp transistor with guard-ring (LPTGR) is also shown in Fig.8b. The collector-base reverse bias causes a depletion region to be formed. If the energy of illumination on the junction is great enough, electron-hole pairs are created and photon induced current occurs. There are two significant features of the lateral photo-transistor as designed in a CMOS process:

a) the positively biased minimum width polysilicon gate, and b) the n+ diffusion in the n-well base.

The positively biased polysilicon gate sets the base width of the device, thereby reducing the sensitivity of β to fluctuations in bias conditions as seen in the lateral pnp transistor fabricated in a true bipolar process. The n+ diffusion in the base increases the majority carrier concentration, thus more photo-induced current is generated for a given illumination intensity. Therefor, the pnp LPT in CMOS process has a higher, more stable β than the lateral pnp available in a true bipolar process.

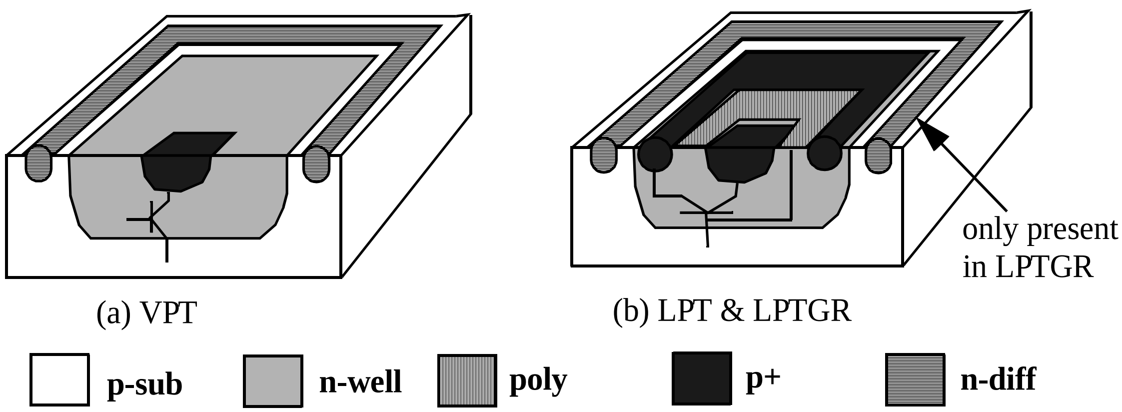


Fig.8: Configuration of various pnp transistors

Despite their high optical sensitivity compared to photo-diodes, photo-transistors are not commonly used in X-ray imaging application because of their high dark current which is the result of the base-collector leakage current multiplied by the gain β. This dark current severely lowers the dynamic range. In addition, β in a photo-transistor array can vary with more than 20% over the chip. The variation in β is caused by poor control of the base width, and it is also very sensitive to temperature variations. In an image sensor application this leads to high fixed pattern noise (FPN) in the captured image. Nevertheless, the transistor gain can be a great advantage over photo-diodes if a dark current cancellation technique is employed. A possible technique for cancelling the dark current is to use a dummy pixel replica with a shielded photo-transistor to subtract the dark current in the active pixel array as illustrated in Fig.9. However, a slight fill-factor loss will result from the addition of two extra transistors to the pixel circuit.

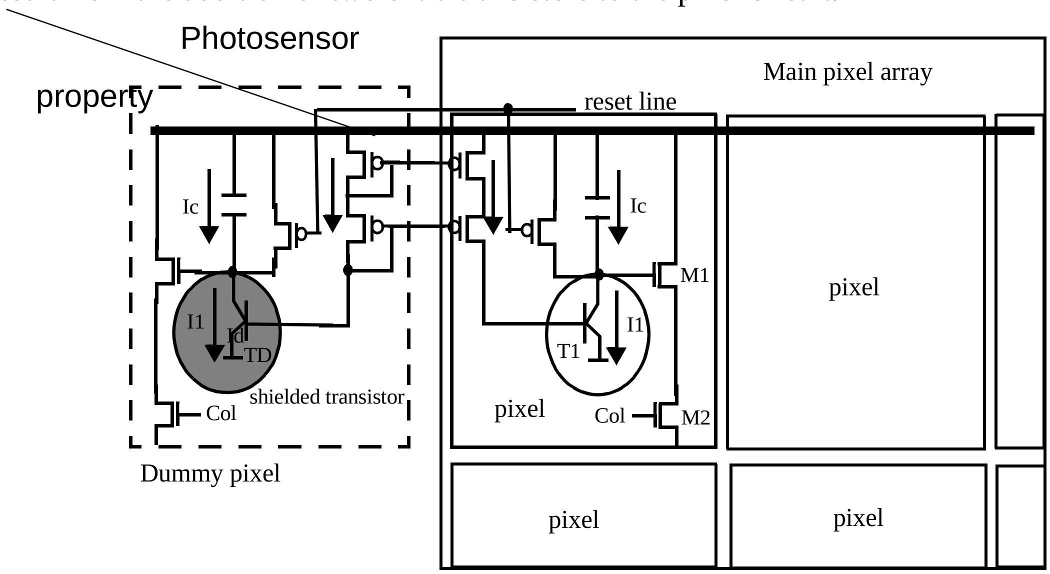


Fig.9: A schematic diagram illustrating dark current cancellation technique using a dummy pixel with shielded transistor.

The above circuit concept Fig.9 was tested in a fabricated prototype [58], and it exhibited a good dark current cancellation that was comparable to photo-diode dark signal levels (Fig.10). Complete description of the device is given in PAPER4. Further investigations on the circuit concerning improvement in the FPN and temperature effects are underway.

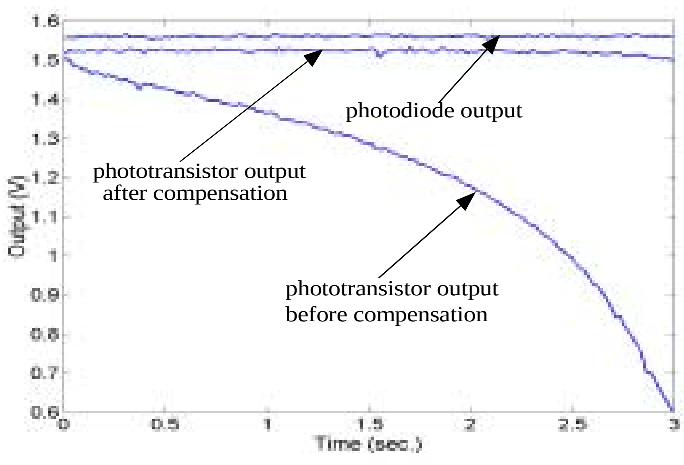


Fig.10: Dark output signal of the phototransistor before and after current cancellation compared to the dark signal from a photodiode

A brief comparison between the performance of the three CMOS pixel detectors that were coated with scintillating material described in PAPERS 2, 3 and 4, is listed below (Table 1).

Table 1: Comparison between various properties of CMOS detectors

	p-diff/n-well	n-well/p-substrate	uncompensated photo-transistor	compensated photo-transistor
optical sensitivity	low, peaks in the blue light range	high, broad spectral response	higher, broad spectral response	higher, broad spectral response
dark current	low	low	high	low
direct X-ray detection	very low	low	high	high
array FPN	low	low	low	high
MTF [85]	good	lower	best	
sensitivity to bulk noise	not sensitive	sensitive	sensitive	sensitive

3.2. Pixel Design for Detector flip-chip bonding

In this type of integrating pixels, semiconductor detectors formed in a separate material, such like a GaAs wafer, is hybridized to a CMOS readout chip by flip-chip bonding [46][47]. In the design of this pixel an in-pixel charge storage is essential to accommodate the large amount of charge generated from the detector. This is essential for radiation imaging because it is desired to collect as large a number of photoelectrons as possible to get a lower statistical noise. For a dental X-ray imaging system where GaAs pixel detector array is to be flip-chip bonded to a CMOS read-out chip, we have fabricated a prototype circuit that can accommodate up to 100 Millions electrons which is twice the charge capacity specified for this application. A high poly/poly capacitor CMOS process was used together with a special layout technique to maximize the capacitor area. Since the readout speed is a major concern, the design employed segmentation buffering of the clock lines to reduce the spread RC effect of the long clock lines. Clocking speed up to 100MHz was measured. The chip was not flip-chip bonded to the pixel sensor because the chip size was much smaller than the size which could technically be handled. Thus, no X-ray measurements were done. The design is demonstrated in PAPER1 appended to this thesis. Other methods to increase the dynamic range could employ external circuitry to connect additional capacitors [59].

3.3. Pixel Design for Wire-bonded Detectors

In pixel designs for wire-bonding, the pixel array size in an ASIC design is usually limited because of the limited packaging and wiring capability. The need for high charge storage capacity is much more crucial in applications where a much higher input signal current is received. The ion beam current in an ion implanter is an example of such applications. Beam currents in the order of microamperes are measured, which means a charge capacity in the nanofarad range is demanded. In PAPER8 we are presenting a novel idea for real-time ion beam profiling. The method employs a graphite detector matrix that is wire-bonded to a remote electronic readout. The graphite matrix replaces Faraday cup that has conventionally been used for the beam current measurement. In the context of the subject of this thesis, this system represents high energy particles imaging.

We have designed a 100 pixel array CMOS ASIC readout, for ion implanters, where the chip was wire-bonded to the graphite pixel array. In this design a current mirror has been used to isolate the output nodes of the detector from the input nodes in the CMOS pixels. This connection insures that the detector node will be insensitive to the variation in capacitor voltage in situations when the high energy beam is decelerated before it reaches the target, Fig.11. A low-voltage cascode current mirror [60] was used because it has low minimum saturation voltage. This will minimize the influence on the input dynamic range (capacitor voltage). The pixel size was 520x520μm^2. The readout mode is started by the Reset signal that charges the integrating capacitor to 5 volts by the parallel transistor. Simultaneously the gate transistor is switched on by the Gate signal allowing the beam current to flow into the pixel through the current mirror. The capacitor is thus discharged with a current magnitude equals to the beam current. After a predefined integration time the Gate transistor is switched off and the entire array is then read out using a row-column addressing circuit. During the Gate off the currents from the all pixel detectors is directed towards a summing node where it will be available for measurement. The ASIC chip photograph and a magnified micro-photograph of some pixels are shown in Fig.12. Fig.13 shows a picture of the graphite pixel detector and a real time sequence of images representing the ion beam movement across the detector. A modified chip design, that should allow flexibility in dynamic range control, and adequate circuitry for measuring the total beam current, is in fabrication stage. The flexibility in dynamic range will be achieved by a binary-weighted current mirrors in each pixel.

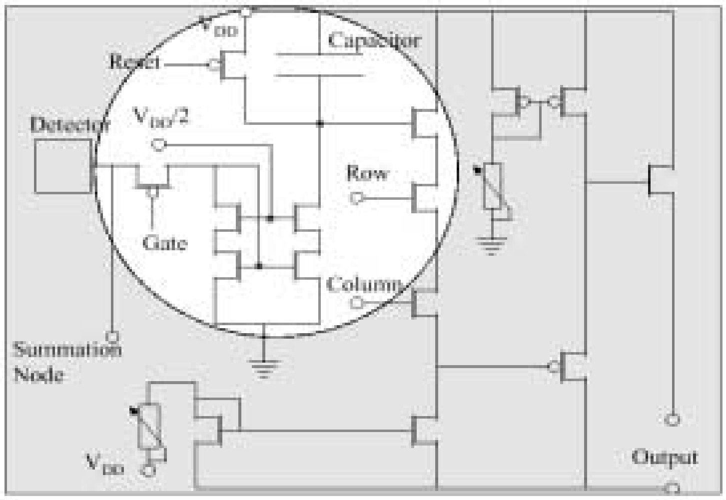


Fig.11: Schematic diagram of the pixel circuit of the ion beam profiler.

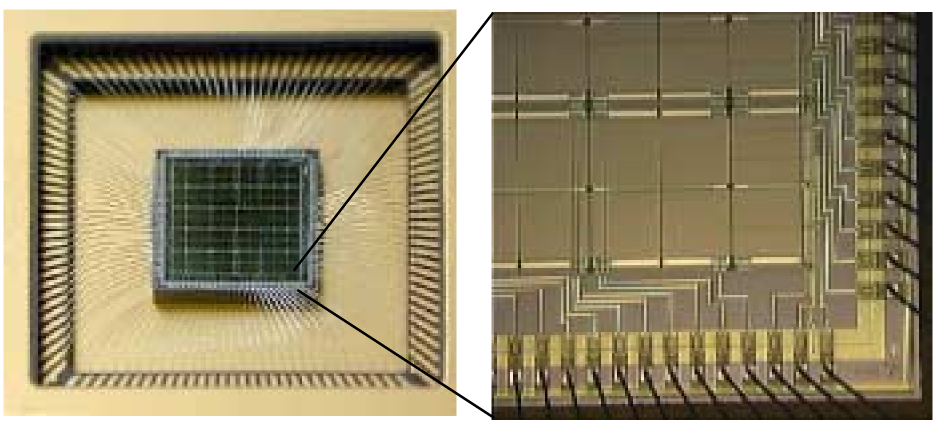


Fig.12: Chip photograph of the ion beam profiler ASIC (left), and a microphotograph of part of the chip (right)

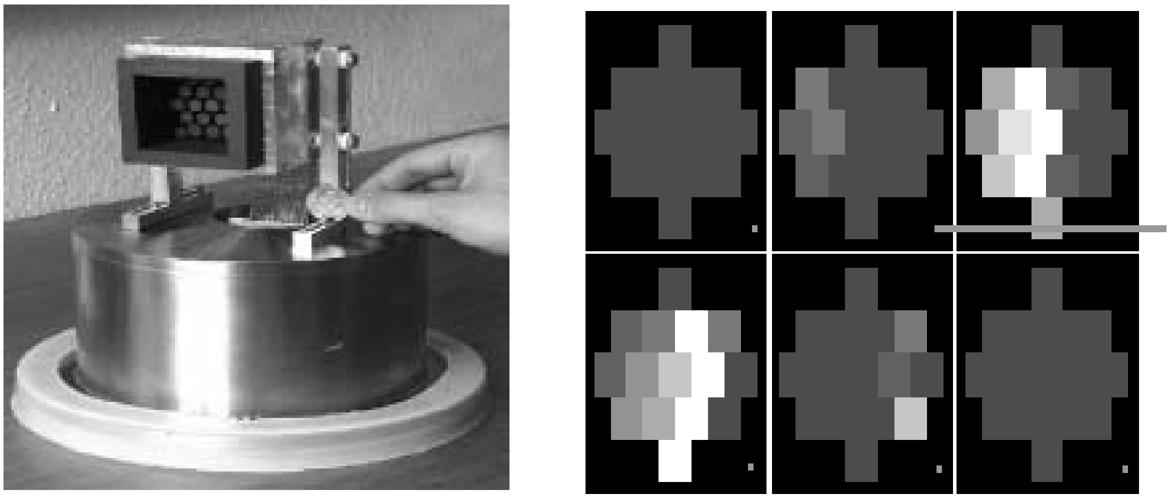


Fig.13: The graphite detector mounted to the flenge (left), and a sequence of images showing the ion beam moving across the detector. The images taken by the ASIC chip (right).

4

PIXEL ELECTRONICS FOR SINGLE PHOTON COUNTING

An important property of pixel detectors is the very small pixel capacitance (50 - 150 fF, depending on the design of the pixel sensor). This allows implementation of a low-noise amplifier (<100e$^-$) with small power consumption (about 50 μW/pixel). Thus a detector can easily be operated in true single photon counting mode at room temperature.

X-ray imaging using hybrid pixel detectors in single photon counting mode is a relatively recent and exciting development.The photon counting mode implies that each pixel has a threshold in energy above which a hit is recorded. The advantages of hit counting with real-time processing compared to traditional film-based methods are: its linearity and it has, in principle, infinite dynamic range, very good contrast performance and intensity analysis, intensity-independent detection efficiency (no haziness), multiple exposing capability and time-resolved detection (film sequence), and low-dose capability. The ideal detector for medical imaging should have good spatial resolution and be as safe as possible in terms of risk to the patient (as efficient as possible to minimize the dose necessary for good diagnosis). These requirements can be optimized by choosing the best detection techniques. However, unlike the integrating type pixel architecture, the photon counting pixel design is a complicated task where the skill for mixed analog and digital circuits design techniques is required. VLSI chip sets, rather than a single integrated system-on-chip, do exist [61][62]. The front-end electronic readout for single photon counting typically contains an analog signal processing channel and a digital circuitry. The analog part consists of a preamplifier, a shaping amplifier and it can include other circuits like base-line restorers and pileup rejecters. The digital part, on the other hand, consists of a pulse discriminator and a counter in addition to a readout logic. Because of this large number of electronic components, that could reach several hundreds of transistors, the pixel design is driven mainly by area and power consumption in addition to mixed mode operation constraints. A most pronounced readout chip for single photon counting was developed as part of the Medipix project[1] [63]. The block diagram of the Medipix pixel cell, shown in Fig.14, is a typical example of photon counting pixel design. The design has the smallest pixel area reported so far (170x170µm^2).

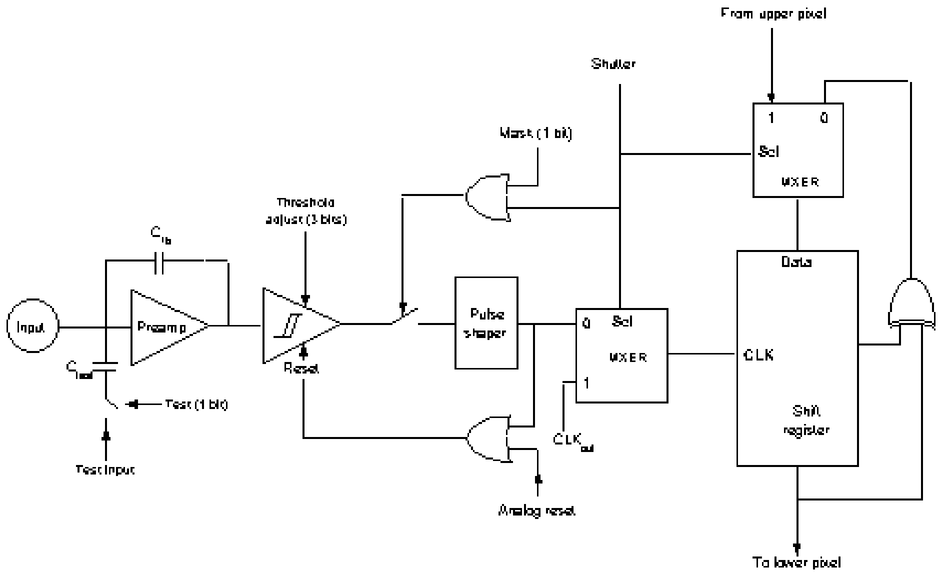


Fig.14: The Medipix pixel block diagram [63]

4.1. The Analog Section

The analogue part of an event counting pixel cell starts with the preamplifier followed by a pulse shaping amplifier. A block diagram of a generalized analog pulse processing channel is displayed in Fig.16. The preamplifier design is critical since it should match the detector interface. The circuit should be fast with low noise performance. The charge-sensitive amplifier (CSA) is the most common configuration in use because its conversion gain is independent of the detector anode capacitance variation (Fig.15a).

1. Medipix project[1] is a common development between CERN, University of Freiburg, University of Glasgow and INFN.

It is very often that the preamplifier stage is implemented as a single-ended folded cascode (Fig.15b). This circuit topology offers a better performance in terms of stability and gain compared with a standard cascode. It allows the cascode transistor to operate at a lower current, which gives higher output resistance. This is in contrast to the conventional cascode where the current required for the input transistor must flow in the whole chain. The choice of a PMOS input transistor gives a better result in terms of flicker noise compared to NMOS transistors.

If $A >> (C_i+C_f)/C_f$

$V_{out} = -A.V_{in}$

$= - A.Q/(C_i+(A+1)C_f)$

or: $V_{out} = -Q/C_f$

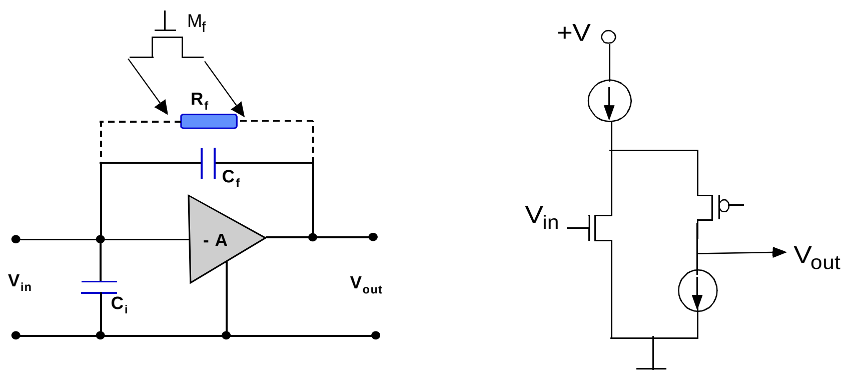


Fig.15: (a) A simplified CSA , and (b) a typical folded cascode topology

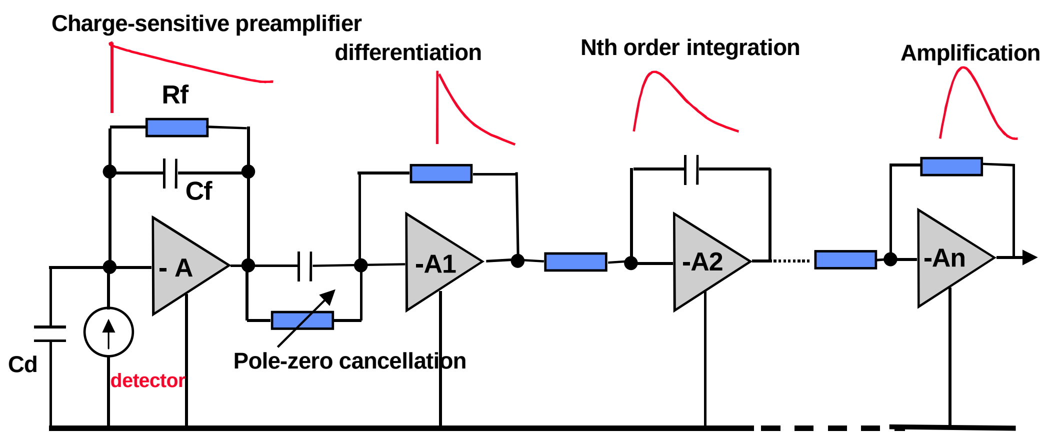


Fig.16: A block diagram of a general analog pulse processing channel

CMOS technology offers many advantages over other technologies in implementing these amplifiers because of the high input impedance of the MOSFET transistor and its readily available inexpensive prototyping. However, one constraint in implementing small area circuits for many CSA applications is the on-chip feedback resistor (R_f). The value of this resistor is typically in the Megaohm range for a minimal parallel noise contribution. To reach that high resistor value using MOS transistors the transistor should be operated in its non-saturation region, which is a challenging objective since its R_{ds} is subject to fluctuations due to the variations in its threshold and bias conditions. Various methods are employed to use a MOSFET as a feedback resistor in CSAs. It can be connected as a parallel switch which could then be closed by a reset pulse. However, this method costs more complexity in the system and adds reset noise. CMOS CSA employing differential transconductance are also used [64]. This, however, suffers from the variation in effective transconductance with leakage current and hence affects the gain. Other CSAs implement CMOS fully compensated continuous reset systems where the feedback MOSFET is biased by a special circuit using N replicas of the feedback transistor [65][66]. This technique offered good performance at the cost of a large number of replica transistors (N) that could reach tens of transistors. Such a number of transistors results in an increase in the layout area and parasitic capacitance as well as power consumption. Various preamplifier techniques are reported in the literature [67][68][69].

The preamplifier output signal is usually an exponential pulse (if a resistor feedback is used) with a long time constant. The exponential pulse is applied to the first stage of the shaping amplifier, which is a differentiator to filter out the exponential pulse tail; thus filtering the low frequency components of the input signal. A pole-zero cancellation network is implemented to eliminate the amplifier's output over-shoot that results from the differentiation. The differentiator is then followed by a number of integration stages that work as a low-pass filter. In this stage the high-frequency components in the signal, including the high frequency noise, are filtered out. An amplified semi-Gaussian pulse is then produced at the shaper output that is ready for digital processing.

We have proposed and published a new biasing technique to implement a MOS transistor as a feedback resistor in CSAs with a stable operation against the unavoidable process variations and other variable conditions. Our circuit exhibits a full tuneable control on the MOS transistor resistor together with additional features like automatic pole-zero cancellation, shaping time tuning and voltage gain tuning capability. A complete circuit description and simulated results are displayed in PAPER6.

4.2. The Digital Section

The semi-Gaussian pulse that is output from the preamplifier-shaper is input to a discriminator. The discriminator can either be a simple integral discriminator or a window discriminator. In the integral discriminator type, a comparator outputs a logical signal every time the input pulse amplitude exceed a preset threshold. This is the main feature characterizing the single photon counting pixel in comparison to integrating pixels. That is because the pulse amplitude carries a unique information about the radiation source, namely the radiation energy. On the other hand, it is obvious that the threshold voltage can always be set above the noise level. In the window discriminator type, two comparators are used to output appropriate pulses when the input exceeds a lower threshold and lying below an upper threshold (Fig.17). In this latter type single energy selectivity is feasible and thus better imaging capability is obtained.

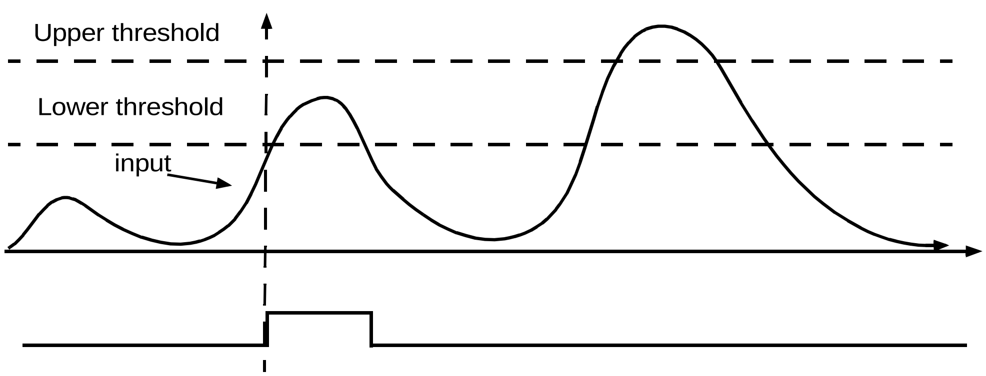


Fig.17: The window discriminator outputs a logic pulse when the input pulse lies between the lower and upper thresholds.

Several design techniques are employed to implement a reliable window discriminator. One inherent problem with window discriminators is the time-walk which is a result of variations in the pulse amplitude. The discriminator output pulse timing depends on the input pulse amplitude since lower pulses reach the discrimination level later. As a result, this time-walk might generate spurious pulses that can lead to faulty counts. Although pulse height discrimination is the main technique in radiation imaging, pulse shape discrimination finds its applications when timing information are required. Several design techniques are described in the literature [70]. However, pixel applications force many limitations including area constraints [71]. P. Fischer et. al. in their design [72] of the window discriminator, eliminated the time walk problem completely by implementing two counters per pixel. The contents of the counters are then subtracted off-chip to obtain the resultant count that represent the events those fell within the discriminator window. However, this design is inefficient as far as pixel area is concerned. In addition, the digital switching activity can significantly degrade the noise performance and increase the power consumption. The read-out speed will be halved since two counters are read out instead of one. As an alternative we have designed a window discriminator that is all-digital using asynchronous logic. The circuit is area efficient which makes it suitable for pixel circuits. Moreover, it does not rely on internal time references and it eliminates the need for a double counter [73]

In radiation imaging, digital signals produced by the in-pixel discriminator must be counted simultaneously in all the pixels. Standard synchronous or asynchronous binary counters are usually used for this purpose. In order to read out the contents of these counters, a bus system is needed or the counters are transformed into shift registers that are then read sequentially. Both solutions require additional chip area and circuitry. The simplest way to implement an N bit state machine with a maximum number of states is probably a linear feedback shift register (LFSR) [74]. This consists of an N bit shift register and an exclusive OR gate (or exclusive NOR) with two inputs, that feeds back the output and an intermediate tap to the input of the shift register (Fig.18). The obvious advantage of this counter is its simplicity. It allows a very compact and regular layout and can reach very high counting speeds due to the short logic delay in the feedback path. Furthermore, the complexity of such counter does not increase with increasing the length N. For the sequential readout almost no additional circuitry is needed. Moreover, an up/down counter can be implemented fairly easily by constructing a shift register that can shift in both directions (left/right). This requires an additional multiplexer per flip-flop. A second exclusive OR gate is used to supply the input signal for the second shift direction. A minor disadvantage is the randomness of the generated bit pattern, which has to be decoded off-line in order to reconstruct the number of counting pulses.

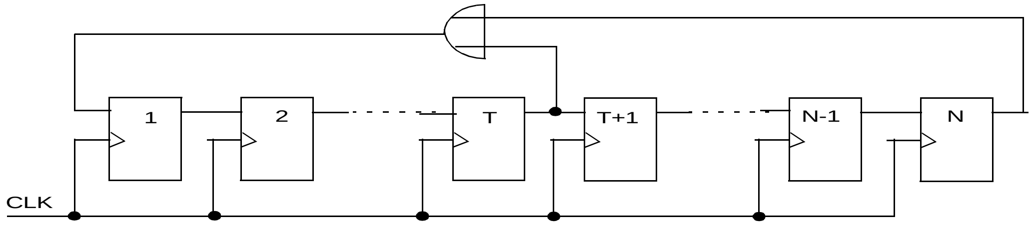


Fig.18: A simple LFSR

As have been pointed out earlier, the digital switching of this synchronous LFSR can severely degrade the noise performance since the analog and digital circuits are closely integrated in the pixel sensor. We have developed a design technique for a low interference counter that is described in details in PAPER7. The method introduces the idea of a prescaler (ripple counter) that scales down the digital switching of the whole counter by up to seven times. The counter has the same dynamic range and hardware cost as an LFSR. Design methods have also been devised for the counter that enables the designer to weigh the counter performance against the digital switching in the pixel.

To get rid of the digital circuitry in the pixel, we have proposed an all-analog single-channel analyzer (SCA) in PAPER6. With the presence of high poly/poly capacitance CMOS processes, an in-pixel capacitor can be implemented to store a signal that is generated by the window discriminator in a charge/discharge fashion. Thus, all the digital part in the pixel circuitry will be eliminated. In this proposed design an input pulse amplitude that exceeds the lower-level discriminator threshold triggers a monostable multivibrator that, in turn, switches on a constant current source to charge the capacitor for a fixed period (p). If the input pulse reaches its peak below the upper-level discriminator threshold, the capacitor remains charged an amount of charge equivalent to one pulse count. If the input pulse amplitude exceeds the upper threshold level, then another monostable multivibrator will be triggered which, in turn, switches on a current sink for the same period (p). This will discharge the capacitor an amount of charge equals to the previously supplied charge by the current source, and hence removing the off-window pulse. The operation is analogous to the up/down counter described in the previous paragraph. Using an in-pixel 10pF capacitor will yield an estimated area cost of 5650μm^2 in the AMS[1] CYE 0.8μm CMOS process. A corresponding 15-bit digital counter (a typical counter length for most imaging applications) in the same process will occupy ~45000μm^2. An area saving close to an order of magnitude will be obtained with the implementation of this an-analog SCA. Furthermore, with a proper design of the current source values and the monostable periods, the 10pF capacitance will result in a dynamic range that is much higher than a 15-bit counter.

1. Austria MicroSystems CYE CMOS process has a poly/poly capacitance equals to 1770pF/mm^2

5

INFLUENCE OF PIXEL DESIGN ON IMAGE PROPERTIES

Basic definitions of imaging theory in medical and biological applications, and their importance for detector development is demonstrated in [75].

5.1. Dynamic-Range and SNR

Dynamic range (DR) and signal-to-noise ratio (SNR) are two characteristic measures of pixel performance. In integrating pixels, the dynamic range quantifies the ability of a sensor to image highlights and shadows; it is defined as the ratio of the largest non-saturating current signal i_{max}, i.e. input signal swing, to the smallest detectable current signal i_{min}. This is typically taken as the standard deviation of the input referred noise when no signal is present. Using this definition and the sensor noise model it can be shown that DR in dB is given as [76]:

$$DR = 20\log\frac{i_{max}}{i_{min}} = 20\log\frac{q_{max} - i_{dc}t_{int}}{\sqrt{(\sigma_r)^2 + qi_{dc}t_{int}}} \qquad (8)$$

where q_{max} is the well capacity, q is the electron charge, i_{dc} is the dark current, t_{int} is the integration time and $(\sigma_r)^2$ is the variance of the temporal noise. For a voltage swing V_s and photodetector capacitance C the maximum well capacity is $q_{max} = CV_s$. Equation (8) shows that the DR increase roughly with the square-root of the pixel charge capacity. Several efforts have been done for the DR enhancement. Some techniques implement well capacity adjusting schemes as described by Knight [77] and Sayag [78], or by current-mode background suppression [59].

Photonic noise is caused by the statistical distribution of the flux of X-ray photons. It is the fundamental noise limit for any system. On the other hand, fixed pattern noise (FPN) is caused by sensitivity variation over the sensors and it is directly proportional to the signal. FPN can be eliminated by individual calibration of each pixel using gain maps as described in [46][79]. Electronic noise is caused by factors like dark current in the sensor, the amplifier noise and externally induced noise. It is not directly related to X-rays, but the exposure time increases the influence of the dark current. SNR is the ratio of the input signal power and the average input referred noise power. SNR in dB as a function of photo-current, i_{ph}, is given by [80]:

$$SNR(i_{ph}) = 20\log\frac{i_{ph}t_{int}}{\sqrt{(\sigma_r)^2 + q(i_{ph} + i_{dc})t_{int}}} \qquad (9)$$

Equation (9) also shows that the SNR increases roughly with the square root of the pixel charge capacity. Several experiments have shown that the photonic noise in a semiconductor is Poisson distributed, where the standard deviation, σ, in the signal is the square-root of the mean number of detected X-ray photons [81].

However, in a scintillator coated X-ray image sensor the signal is generated both by the X-ray photons captured in the scintillator and by the X-ray photons captured in the semiconductor sensor. The latter is usually referred to as *direct X-ray absorption*. Since the amount of generated charge per absorbed X-ray photon differs significantly from the scintillator to the semiconductor, the image properties is influenced by both sensors. Theoretical analysis shows that direct X-ray absorption significantly reduces the SNR. If n_1 and n_2 are the average number of photons absorbed in the scintillator and photosensor respectively, and k_1 and k_2 the their respective generated signal then [82]

$$SNR_{tot} = \frac{k_1 \cdot n_1 + k_2 \cdot n_2}{\sqrt{k_1^2 \cdot n_1 + k_2^2 \cdot n_2}} \qquad (10)$$

Equation (10) is valid for all hybrid sensor systems. The best result is obtained when all the signals in the detector are generated by one process only (suppressing the direct absorption). For a sensor with ideal noise performance the SNR is only determined by the photo-current. The only way to achieve that is to improve the sensitivity, thereby causing more photons to contribute to the image. Both equation (8) and (9) dictate the choice of a large charge signal capacity and hence a large pixel size. However, this requirement contradicts the requirement of high spatial resolution as described in section 3.2.

5.2. Pixel Size and Resolution

Image resolution is the primary parameter for performance characterization of any imaging system. Various definitions have been introduced for quantification of image resolution. Generally speaking, image resolution, or resolving power, is the ability of an imaging system to separate two localized signals. Well-known resolution measures include the Rayleigh criterion, the bandwidth of modulation transfer function (MTF), the cut-off-frequency of MTF, the full-width-at-half-maximum (FWHM), full-width-at-tenth-maximum, and standard deviation of the point spread function (PSF) of the system. Depending on which definition of image resolution is used, a better image resolution may correspond to either a larger value (for example, the wider bandwidth, the better resolution, or a smaller value (the smaller FWHM, the better resolution) of resolution measure.

Spatial resolution is defined as the modulation obtained when exposing a pattern of a certain spatial frequency. It could either be defined as MTF assuming a sine wave pattern, or CTF (Contrast Transfer Function) assuming a bar (square wave) pattern [83][84].

As mentioned in section 5.1, a large pixel size is desirable to achieve high dynamic range and SNR, but this choice opposes the resolution requirement, which requires smaller pixel sizes. Because of contradiction between the DR and SNR at one hand and the MTF on the other hand, there must exist a pixel size that compromises these criteria. A methodology for determining an optimal pixel size is proposed in [80]. The MTF for scintillator coated pixel sensor is not only affected by the lateral dimensions like the pixel size and fill-factor, but also by the vertical dimensions like the scintillator thickness and the photosensor sensitive depth. A thick scintillating layer degrades the spatial resolution because of the light diffusion in the scintillator over the neighboring pixels. On the other hand, a larger sensitive depth of a photo-diode in a pixel array degrades the MTF because of diffusion of carriers into neighboring pixels. It has been shown that the MTF of a pixel detector with a thinner sensitive depth was higher than that of the larger depth [85]. A monte Carlo simulation of pixels with different structures in a 0.8μm CMOS process resulted in the MTF plots in Fig.19. The MTF was simulated as the Fourier transform of the carriers distribution that is generated by X-ray exposure from a point source at a central distance between the first and tenth pixel for 10 adjacent pixels. It is clearly seen that the p-diff/n-well photo-diode pixel has a higher resolution than the n-well/p-sub photo-diode pixel. This is because, in the case of the n-well/p-sub photo-diode the carriers generated in the bulk from direct X-ray absorption contribute to the signal in the neighboring pixels (cross talk) by diffusion. However, the photo-transistor (as sketched in Fig.3-a without the n-diff. layer) showed the highest MTF because the emitter-base junction length, that contribute to the amplified main signal, is rather short and it is centred in the pixel. This makes the diffusion length for charged carriers to travel before they can be swept through the base-collector junction rather long, and hence the number of carriers diffused to the neighboring pixels is consequently small.

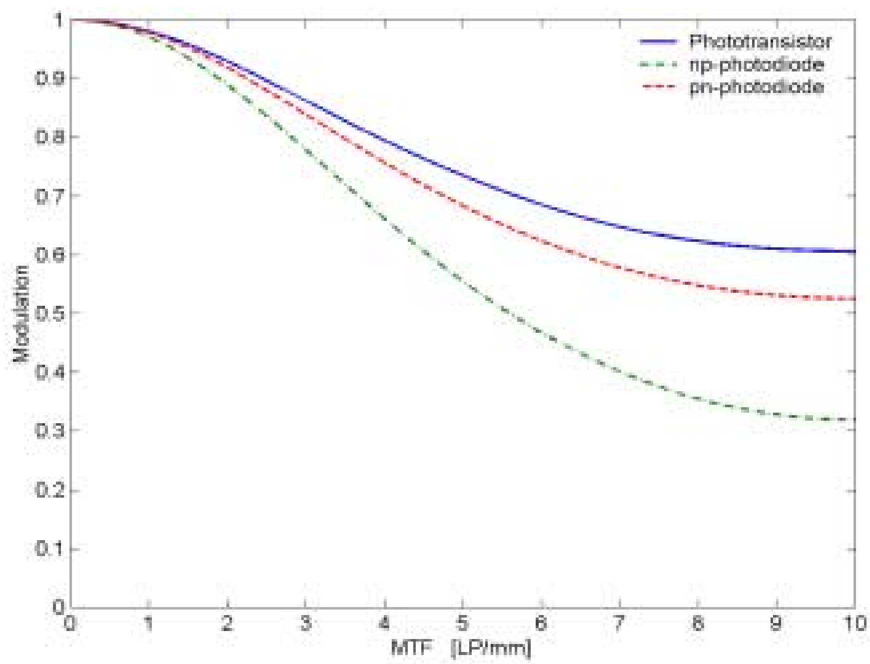


Fig.19: A plot of MTF for three types of scintillator-coated pixel sensors.

5.3. Layout and Cross-Talk

When designing a scintillator coated pixel sensor, careful layout design has to be considered to achieve good performance. A pixel layout should achieve a maximum fill-factor by, e.g., designing pixel patterns using mirroring and flipping the core pixel [47]. W. Snoeys et. al. demonstrated layout techniques to enhance the radiation tolerance [86] by designing all NMOS transistors in enclosed geometry and introducing guard-rings wherever necessary.

Cross-talk is an important opto-electronic phenomenon which can seriously degrade the performance of an imaging system. Carriers absorbed in the substrate under a particular pixel can diffuse to neighboring pixels, contributing to photo-current in those neighbors, and create a “spreading effect on the image”. Cross-talk is a strong function of both wavelength of the incident light and the diffusion length of the minority carriers. Therefore, longer wavelength light and increased packing density both degrade cross-talk performance. Singer and Kostelec and Ohba, et. al. discussed a methods for reducing diffusion of carriers from pixel to pixel in [87]**.** One method is to include a reverse-biased diode between each two pixels. Guard-rings around each pixel can also be used to reduce the cross-talk.

6

SUMMARY AND CONCLUSIONS

6.1. Thesis Summary

The leading detection materials for high energy photons (X-rays and gamma-rays) are semiconductors with silicon being the most mature and inexpensive for soft X-ray imaging. GaAs and CdTe are emerging semiconducting pixel detectors because of their high stopping power for ionizing radiation. However, due to the immature technology for these materials for monolithic detector/readout fabrication, they are usually hybridized to silicon readout circuits. Hybridization of wafers from different materials has become feasible with the readiness of the state-of-the-art flip-chip bonding technology.

The limitation of using silicon in harsh environment (high radiation doses and high temperatures) triggered research in diamond and silicon carbide pixel detectors. To reach higher energy resolution than that with conventional semiconductor detectors, superconducting detectors are being developed because of their gap energy that can be as small as meV.

Scintillating materials are used as radiation detectors as a coating layer on readout circuits. CMOS active pixel sensors are the promising devices to replace charge-coupled devices (CCDs) due to their various advantages, particularly system-on-chip capability.

In this thesis the two readout electronic techniques for radiation imaging were discussed. Namely; integrating type pixel electronics, and photon counting electronics. Photon counting image sensors are the most common in use because of the numerous advantages compared to integrating type sensors. The main features being energy resolution and dynamic range.

The design approach for a CMOS X-ray pixel detector coated with a scintillator depends on several factors starting with the application specifications. There are many contradicting alternatives that necessitate trade-offs to be taken. In the scintillating material choice, the detection efficiency demands a thick coating layer, which opposes the resolution requirement of a thin scintillating layer. The various photo-detectors in a CMOS pixel offer different alternative properties that make them feasible to implement. The shallow sensitive depth photodiode (p-diff/n-well) is relatively immune to direct X-ray absorption, which is a great advantage over the n-well/p-sub photodiode and photo-transistor as far as SNR is concerned. Moreover, its optical sensitivity peaks in the blue light range, which means the rest of the light range is optically filtered in the detection process. However, its optical sensitivity is far below an efficient level. To make an efficient pixel sensor using this photodiode an in-pixel preamplifier should be included. The n-well/ p-sub photodiode, on the other hand, have better quantum efficiency but higher X-ray absorption (lower SNR), and higher cross-talk (reduced MTF). In addition, its anode is subject to direct noise pick-up from the substrate. The third choice from the CMOS photo-detectors is the various photo-transistor structures. Despite the photo-transistor high sensitivity (higher than photo-diodes by a factor β) its high dark current limits its use in high performance image sensors since it significantly degrades the dynamic range. Moreover, the high variations of β value over a pixel array, in addition to its dependence on temperature, severely worsen the noise performance by introducing a high FPN. Nevertheless, if circuit techniques, e.g. in Fig.9, is applied, a significant improvement in DR can be achieved. Moreover, the high optical sensitivity of the photo-transistor implies that higher MTF can be achieved by allowing smaller pixel sizes and lower radiation doses.

In a high energy particle imaging effort, we have devised a novel real-time ion beam profiler for ion implating machines. The concept uses a pixellated graphite as the detection medium that is readout by a remotely wire-bonded CMOS ASIC chip. The device has been tested in real implanters and successful real time ion beam profiling was performed.

Pixel electronics for single photon counting is a recent development which is mainly driven by small area and power consumption constraints. Circuit techniques for single photon counting pixel electronics and design challenges have been discussed. A design technique dealing with the implementation of a monolithic high resistance utilizing MOS transistors has been introduced. Significant simulation results have been obtained as far as high value resistor stability, pole-zero cancellation tunable shaping time and gain, are concerned.

The mixed mode design considerations for single photon counting pixels have been addressed in two PAPERS (6 and 7). In the first approach we proposed an all analog pixel where all the digital part in the pixel is replaced by a high value in-pixel capacitor. A significant improvement in the pixel performance is estimated regarding reduction in pixel area, power consumption and noise in addition to increasing the readout speed. The second approach addresses the design of the in-pixel digital counter for the least digital switching interference and power consumption and improvement of readout speed.

6.2. Papers Summary

In our work that is related to the title of this thesis, we have designed and fabricated several prototype pixel sensor devices to investigate the performance of new sensors for dental X-ray imaging: 1) CMOS APS for flip-chip bonding to GaAs pixel detectors, and 2) Scintillator-coated CMOS APS. The results from this work is summarized bellow (PAPERS 1, 2, 3 and 4).

Different circuit techniques were developed for the enhancement of single photon counting pixel readout. These include analog and digital circuits addressing bias, area and noise constrains. The summary is given in PAPERS 5, 6 and 7.

In the high energy particles imaging field, an ion beam profiler for an ion implanter has been designed based on a novel graphite pixel matrix (PAPER8).

PAPER-1) Design of a CMOS readout electronics for dental X-ray imaging: The design of a CMOS APS for bump-bonding to a semiconductor pixel detector is presented. The results form the prototype chip is reported. The functional overview and circuit operation is shown. Techniques for maximizing the input dynamic range and readout speed are demonstrated. The advantages of implementing dental X-ray image sensors in a standard CMOS technology to replace the CCD imaging systems was proved.

(The author designed and simulated the circuit and constructed the circuit layout, in addition to the test and verification of the fabricated chip).

PAPER-2)A CMOS APS for dental X-ray imaging using scintillating sensors: The paper discusses the results from a CMOS APS coated with a scintillation layer The performance of this hybrid sensor has been investigated through the evaluation of the different photo-sensing elements in CMOS technology. For the p-diffusion/n-well, the n-well/p-substrate photodiodes and p-diffusion/n-well/p-substrate transistor, the paper characterizes the devices for dark current, direct X-ray absorption, optical sensitivity and array uniformity. Conclusions were drawn about the best possible photosensor to suit this type of X-ray sensor.

(The author designed and simulated the circuit and constructed the circuit layout, in addition to test and verification of the fabricated chip).

PAPER-3) An integrating CMOS APS for X-ray imaging with an in-pixel preamplifier: This paper presents the performance of the p-diffusion/n-well photodiode when designed with an in-pixel preamplification stage. An improvement in the optical sensitivity was shown together with very low absorption to direct X-rays. The various properties of this design was shown with their impact on the system noise performance and patient dose reduction. The measured results were compared to pixels based on n-well/p-substrate photo-diodes.

(The author designed and, simulated the circuit and constructed the circuit layout, in addition to test and verification of the fabricated chip).

To better investigate the imaging properties of the various scintillator coated photodiode types in a standard CMOS technology, a Monte Carlo simulation of the SNR and MTF of pixel sensor structures was performed [76].

PAPER-4) A scintillator-coated photo-transistor pixel sensor with dark current cancellation: In this paper an integrating photo-transistor pixel sensor in CMOS technology was also investigated with a technique to cancel out its dark current. The dark current was successfully reduced to photodiode dark current levels, but the design suffered from a higher fixed pattern noise. Measured results on dark current and direct X-ray detection were compared to photodiode results. The paper mentioned non-uniformity in the array readout (FPN), and suggests methods to improve the performance in future designs.

(The author designed and simulated the circuit and constructed the circuit layout, in addition to test and verification of the fabricated chip).

PAPER-5) A new biasing method for CMOS preamplifier-shapers: The paper is in the area of single photon counting readout electronics addressing the demand of small pixel area together with low noise performance. A new biasing technique for the in-pixel preamplifier-shaper using a mirror circuit was reported. The new bias enables the implementation of a monolithic stable, high value, in-pixel resistor using the smallest feature MOSFET. The MOSFET resistor is used as the feedback resistor in the CSA. Tuning capability of the resistor value, shaping time and gain was achievable. Moreover, pole-zero cancellation was automatic. The design and simulation are described in.

(The author devised the concept, designed and simulated the circuit).

PAPER-6) An all-analog time-walk free SCA fir event counting pixel detectors: A proposal of new concept of an all-analog photon counting pixel is introduced. The significant reduction in the pixel area, power consumption together with a high increase in readout speed and dynamic range were discussed. The design is fully described with simulations.

(The author devised the concept, designed and simulated the circuit)

PAPER-7) Low digital interference counter for photon counting pixel detectors: In this paper we have addressed the high sensitivity of the single photon counting pixel performance to switching operations in the digital part that designers in CMOS technology have been experiencing. It was shown how the design of a prescaled LFSR would significantly reduce the digital switching and hence the electronic noise. Design methods for the LFSR were presented. Simulated results on a case study is shown a mean current peak reduction by a factor of seven, and the readout speed is expected to double.

(The author described the targeted problems, contributed to the noise model and proposed solution, in addition to contributing to the paper editing).

PAPER-8) Active pixel detector for ion beam profiling: A novel graphite pixel detector that is wire-bonded to an ASIC readout has been demonstrated in this paper. The results from real time imaging of an ion beam implanter, using our sensor system operated in place of the conventional Faraday cup, were presented. This design will help control accurately implanters performance to increase the yield in the ever increasing device density in silicon technology.

(The author designed, constructed and verified the operation of the ASIC chip. He also participated in the experimental setup and the ion beam measurements on the ion implanter).

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