DEVELOPMENT OF A NEW SILICON BASED DETECTOR MODULE FOR PET

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By

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CERTIFICATION

I, P.R. Hooper, declare that this thesis, submitted in fulfilment of the requirements for the award of Master of Science (Hons), in the Department of Engineering Physics, University of Wollongong, is wholly my own work unless otherwise referenced or acknowledged. The document has not been submitted for qualifications at any other academic institution.

Peter R. Hooper

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**SYMBOLS & ABBREVIATIONS**

CMRP | Centre for Medical Radiation Physics, University of Wollongong
UOW | University of Wollongong
UOM | School of Physics, University of Melbourne
Abstract

In the quest to develop high spatial resolution Positron Emission Tomography (PET), the goal of this work is to characterise the innovative silicon detector array that forms the foundation of a new PET detector module design. The new detector module circumvents the current requirement for photomultiplier tubes by the implementation of its solid-state detector analogues.

The detector module is based upon pixelated scintillation crystals optically coupled to custom designed silicon 8x8 photodiode arrays and read out by a 128 channel low noise charge sensitive preamp. Each element pad of the array has an active area of 3x3 mm² and, for the work described in this thesis, is coupled via a fan-out connection system to the testing instrumentation. The testing instrumentation included I(V), optical excitation and nuclear spectroscopy (alpha particle and gamma ray spectroscopy). This work fits in as part of a larger project to provide proof of principle of this new PET detector module design.

The dark current of individual pixel elements of an unbonded array was found to be 460 pA at an operating bias voltage of 40 V.

To investigate uniformity of response across the array, three different methods have been used. Spectroscopy response on the photopeak of ¹³⁷Cs in the case of optical coupling of each pixel with 3x3x3 mm³ CsI(Tl) showed uniformity of collected charge of ±28%.

Investigation of direct response of pixels across the array with 59.9 keV X-rays from ²⁴¹Am source demonstrated excellent response.

To investigate the variation in uniformity of the detector pixel, the optical response of the array was measured using a laser diode (spot size of ~50 µm) and was found to be ±12%. The uniformity of a single pixel element response using scanning optical beam was measured to be ±1%.

Initial characterisation of 3x3 mm² diode taken from the same wafer used to create the array was optically coupled to a 4x4x10 mm³ LSO scintillator and then to a 3x3x3 mm³ CsI(Tl) scintillator and irradiated by...
511 keV and 1.2 MeV gammas from $^{22}$Na and 662 keV gammas from $^{137}$Cs sources. The energy resolution for the LSO was shown to be 28.8% FWHM for 511 keV and 16.5% for the 662 keV gammas. And the energy resolution for the CsI(Tl) scintillator crystal was 9.5% FWHM for the 511 keV gammas and 7.7% FWHM for the 662 keV gammas.

The time properties of this array were tested with coincidence techniques using current pulses from the front and back – side of the array. For modelling of charge induced in a pixel from LSO photons, in the case of 511 keV photopeak, we used low energy alpha particles generating the same amount of charge. The FWHM of the time spectra was found to be about 700 ps resulting mostly from noise and statistics of charge collection.

This specially developed Si pixel array photodiode was demonstrated to satisfy the critical technical parameters required for high resolution PET scanners. Further work has been recommended for stabilisation the sensitive area of pixel to avoid increasing reverse current due to optical coupling. A special optical adhesive has been proposed to avoid this affect.
1.0 Introduction

Positron Emission Tomography (PET) is an imaging tool that is capable of quantifying in-vivo physiological and biochemical processes using short-lived cyclotron-produced radiotracers. Indications for the use of PET are the diagnosis and assistance in treating many of our leading causes of death, such as heart disease, cancer and brain disorders like Alzheimer’s and Parkinson’s disease.

1.1 New Demands For High Resolution PET

The main physical mechanism behind the operation of a PET scanner is that when a positron interacts with an electron it converts its combined mass (of the positron and electron) into the energy of two equivalent photons. These annihilation photons are simultaneously emitted in opposite directions. The task required by the PET scanner is to detect each annihilation event (or as many as possible) and then accurately determine the locality of where the radiotracer emitted the positron that starting the annihilation event. In this manner, the resultant image formed by the localisation of the radiotracer uptake quantitatively demonstrates how the subject under study distributes the radiotracer. From this knowledge, a physician may be better able to understand the staging of the disease and its treatment.

One advantage PET possesses over other imaging modalities like MRI and CT is that the PET scan reveals a quantitative distribution of radiotracer uptake within the region of interest as opposed to the anatomical “snapshot” imaging as given in MRI and CT. Thus PET is redefining the approach to how some deadly diseases are being diagnosed and treated.

However, when compared to these anatomical imaging techniques, the resolution is currently quite poor, giving only approximately 4-6 mm full width half maximum (FWHM) versus 1 mm (FWHM) for MRI. The new PET detector module now being developed at the Centre for Medical Radiation Physics (CMRP) in collaboration with the University of
Melbourne (UOM) is aimed at significantly improving spatial resolution over the whole field of view (FOV). High resolution PET would allow studies for example, in PET of the brain, where spatial resolution of 2 mm FWHM is desired to enable dynamic studies of blood flow in the cerebral arteries [1].

One significant area where high resolution PET may yield great results is to the imaging of small animals. Animals are studied using PET imaging to evaluate drug responses before they are trialed in humans. Animal models of human diseases are widely used in basic biomedical research to elucidate disease mechanisms and to develop and test new treatments [2]. While autoradiography is already used to study these biochemical mechanisms in animals and obtains a resolution that approaches 100 µm, it is very labour intensive and therefore costly [2]. And because cost is the major determining factor, advances in PET should offer decreased overall operating costs, give faster acquisitions and shorter interpretation times [3].

Also, a great deal of important information is lost when subjects are sacrificed, as in the case of autoradiography. Animal studies in which the same animal is observed over the long term can make for expeditious and safer drug trials because repeated measurements of the same subject may reveal facts that would have otherwise remained hidden.

1.2 Small Animal PET

Clinical PET scanners are less than ideal for small animal PET. They are very expensive, not widely available and fundamentally more important, they do not have adequate spatial resolution or sensitivity for small animal applications [2]. In this case, spatial resolution can be thought of as the distance separating two radiative point sources that are considered as two distinctive points in an image. Sensitivity is the relationship between the total number of radiations emitted from the radiotracer and the count-rate of the scanner.
Small animal PET should ideally possess sufficient spatial resolution to be capable of depicting drug and tracer bio-distribution in the organs of rodents, cats, small simians and other animals [4]. This *in-vivo* characterization of the biochemical processes in new drugs would conceivably simplify and make drug evaluations less costly, in turn accelerating the time needed to bring drugs to market [4]. In addition, this scanner should be of benefit to the development and characterization of new drugs, and in the *in-vivo* study of biochemical processes [4]. Even though PET is an ideal technique for making available *in-vivo*, non-invasive functional imaging in animals, applying existing PET technology to this science has limitations. The greatest of these to overcome is the need for much improved spatial resolution [2].

Given the smallness of the animals to be studied, the imaging environment is more favourable because the scatter and attenuation effects are lower due to there being less depth of animal for the photons to penetrate. And as well, the lowered dose rate required to image smaller animals directly reduces the deadtime losses caused by the lower count rates. The smaller imaging volume of the scanner in combination with the lower count rate enables fewer detector modules to cover given solid angle [2], therefore the cost of smaller PET scanners is less.

### 1.3 Demands For New PET

For PET to become more valuable, new PET systems must gain in becoming able to precisely measure the point of interaction (where the annihilation event takes place), that in turn, will bring about higher spatial resolution. And also, gains in sensitivity and count-rate capability will yield more optimal counting statistics obtainable in a shorter scan time without using an excessively high radiation dose [5], thus further improving spatial and timing resolution.

An improved count-rate capability increases the rate at which the PET scanner is able to process each radiation event that is detected. A PET
scanner requires a certain time to pass so that a detected event is processed. If another event occurs within this time (known as deadtime), this event will not be counted. Therefore, the animal or person under the PET study will receive a dose for no improvement in image quality.

A scanner with low efficiency will detect a lower number of events and this directly translates to a noisier reconstructed image. For this noisier image to be presented in a diagnostically acceptable form, it needs to first undergo smoothing process that results in a loss of spatial resolution [5].

While spatial resolution of PET systems has been progressively improving over the past decade, there has not been a corresponding advance in sensitivity [5]. Higher sensitivity is an important parameter to attain because it assures the most accurate results while using the lowest scan time and dose. Therefore to optimise the PET detector module’s design, improvements must be made in spatial, temporal and energy resolution, underpinned by gains in sensitivity and count rate capability [3].

As a rising tide lifts all boats, achieving gains in just one of these areas will significantly enhance the value of PET. However, the value of PET will become dramatically greater when accuracy in the measurement of the point of interaction is accomplished. For this is a fundamental problem to be overcome when obtaining high spatial resolution in PET.
2.0 New Approaches To PET Detector Modules

This chapter presents a review of the advantages and drawbacks of various PET detector module designs.

2.1 Classical PET Module Design

Early PET detectors made use of single crystals of thallium doped Sodium Iodide (NaI(Tl)), individually coupled to PMTs. They were quite versatile and are still found in operation today. However, with the discovery of the scintillator Bismuth Germanate (BGO), with its high density and high atomic number and its strength to resist hygroscopic deterioration (water absorption), most clinical PET systems in use today make use of the modular block detector design that can multiplex as many as 64 BGO crystal elements to four PMTs [4,6].

This block design takes advantage of BGO’s high density and high atomic number, which directly enhances the system’s sensitivity [4]. However, this design also limits the system’s intrinsic spatial resolution to greater than half the individual crystal’s width [4]. Intrinsic resolution is the best directly obtained resolution, but it is rarely utilized because unfiltered images are practically too noisy [7]. Current scanners have intrinsic resolutions of less than 5 mm (taken at the centre of the field of view), however, because reconstruction algorithms trade off resolution for reduced image variance, the final resolution is usually greater than 8 mm [7].

During the PET scanning process, an annihilation photon is absorbed within a scintillator crystal that is connected to four adjacent photomultiplier tubes (PMT) thus forming a “block” detector as shown in figure 2.1.1. In the process of converting a 511 keV photon to scintillation light photons, the amount of scintillation light detected among the four PMTs is proportional to a specific position in the crystal. In other words, the scintillator light generated as a 511 keV photon is absorbed in the crystal block and is
detected by the four PMTs and the location of this interaction point is determined by the relative responses of the PMTs.

![Figure 2.1.1](image.png)

**Figure 2.1.1** PET block detector module currently in use. (Figure taken from Phys. Med. Biol. 47 (2002) R85–R106.) This illustration shows the cuts in the scintillator block and how Anger Logic determines the position of interaction by relative pulse on the four PMTs.

As the output of each PMT ideally represents the number of scintillation photons the PMT has detected, by comparing the relative response to the annihilation photon of the four PMTs using simple Anger Logic, a mathematical calculation that compares the difference in the response of each PMT allows the location of where the interaction took place in the crystal to be reasonably approximated.

Using this particular light sharing method to determine the interaction position leads to statistical uncertainties and can often result in mispositioning of these interaction events [8], consequently translating to a loss in spatial resolution.

In PET, where spatial resolution is strongly dependent on the size of the detector element, new technologies have resulted in an obvious trend toward smaller and smaller detector elements with the aim of further improving spatial resolution [9].
Saw-cutting the crystal into smaller sections is one method to reduce the detector element’s size. This method is successful in improving the spatial resolution but it comes at a high cost, especially with respect to impaired detector sensitivity. Also, the cost of scintillator material lost when saw cutting is not negligible.

An interesting phenomenon with BGO scintillation crystals is that its light output and energy resolution indicates an underlying dependency on the width of the BGO crystal. The energy resolution quickly degrades with BGO elements thinner than 2 mm even though the decrease in the total light output from the BGO is not large [10]. In thin BGO crystals, scintillation light evidently undergoes multiple reflections on the reflector walls before escaping from the crystal [10].

This accordingly leads to the degradation of the intrinsic resolution because the light attenuation is dependent upon scintillation position, whether it is at the top, middle or bottom inside a thin crystal [10]. Furthermore, the coincident timing resolution, which is the time required for a pair of annihilation photons to be depicted as “coincident,” is found to be virtually independent of the crystal width [10]. This means that there is no timing advantage to be gained by narrowing down the BGO crystal detector elements.

In addition to the lower sensitivity [11] brought about by there being less scintillator material in the saw-cut detector elements to stop the impinging annihilation photons, the process of extracting the scintillation light from long and narrow crystals becomes increasingly difficult.

On top of this, the use of smaller crystals leads to a bigger proportion of inter-crystal cross-talk because of Compton scattering [9], that in turn diminishes spatial resolution. The inter-crystal scattering is reasoned to be the primary cause of event mis-positioning in high resolution block detectors used in commercial PET systems and this contributes to a decrease in tomography image contrast [9].
To reduce this effect, extremely thin (0.3 mm) lead sheets for passive inhibition of inter-crystal cross talk, i.e., the use of inter-crystal septa, has been tested [9]. The inter-crystal septa effectively decreased the requirement for active energy thresholding used for scatter rejection and thus there is an overall increase in the PET scanner’s efficiency [9]. Traditionally, energy thresholding on the photopeak is used to lower the effect of Compton scattering (CS) [9]. When the Energy resolution is poor as in the case when quasi-discrete BGO elements are used, the detection efficiency of the system is lowered [9]. Therefore, any method that reduces inter-crystal cross-talk has the potential to improve the sensitivity of the scanner.

In PET detectors using BGO scintillator elements, only approximately 44% of the 511 keV gammas interact photo-electrically (PE), this means that the important bulk of photons are scattered within the block [8]. This causes a higher pulse pile up effect due to the deadtime associated with the processing of any photon impinging on the scintillator, regardless of whether it interacts by PE or by CS and in short, lowers the counting statistics [8].

Currently PMTs are the photon detector of choice. Their very high gain (more than an order of magnitude greater than PD), and their fast response times, gives PMTs many advantages. However, their relatively low QE (~20-25%) is a significant drawback [5].

Furthermore, to necessitate adequate spatial resolution, many detector elements are coupled to a smaller number of PMTs [5]. Thus the ability to distinctively identify individual elements that are involved in the interaction is strongly dependent on the light output of the scintillator. In the case of BGO, which has a relatively poor light output, this translates to a weakness in accurately identifying the point of interaction [5].

In the current generation of PMTs using the BGO scintillator, the effect of light sharing is estimated to add 2 mm in quadrature to the resolution of the system, relative to the individually coupled crystals [5].

In PET scanners that determine the position of interaction by relying on the distribution of light among several PMTs, higher light output from the
scintillation crystals will naturally lead to improved spatial resolution [12] and better energy discrimination for enhanced sensitivity.

A smaller crystal size results in poorer statistics because of the finite number of scintillation photons, making it difficult to attain the high spatial resolution wished for [13]. However the availability of a new scintillator, cerium doped, lutetium oxyorthosilicate (LSO), possesses the desired characteristics of high light output per scintillation. LSO with its higher light output at a faster rate has the potential to enable advances in resolutions to be made. This is especially true for detectors that rely on sharing scintillation light for determining the position of interaction within the crystal [4].

Reducing the scanner’s detector ring diameter will enhance the sensitivity because the solid angle of detection is correspondingly increased. That is, when the detector face is brought closer to an active source, from geometric considerations, more photons will obviously be detected. However, reducing the ring’s diameter increases the effect of radial elongation, which is an artifact (see figure 2.12) brought about by the annihilation photons deeply penetrating into a scintillator crystal to the point where they actually travel into the neighbouring crystal and make their point of interaction. This causes scintillation photons to be collected in the wrong crystal and this leads to mispositioning of the lines of response (LOR).

![Figure 2.1.2](image.png)

**Figure 2.1.2** Example of how lack of DOI information introduces an error between the reconstructed line of response (LOR) and the true path of the annihilation photons. This radial elongation error worsens closer to the edge of the field of view (FOV).

These misaddressed events cause blurring of the spatial resolution as a result. This radial elongation effect is most pronounced when photons
impinge upon the crystal at oblique angles, causing the attenuation length through the incident crystal to become less.

To eliminate this blurring effect, it is necessary to position the point of interaction more exactly within the scintillator crystal. The measurement of this position is known as the Depth of Interaction (DOI). For more precise DOI measurements, the more the LOR error must be trimmed down.

In summary, to develop a high resolution PET scanner with a small axial coverage, traditional PET design methodologies must be rethought. The current block design used in most commercial PET scanners suffers from several drawbacks that impose limits on the block detector approach to very high resolution PET systems [8]. For these reasons, it becomes clearly worthwhile to examine new approaches to PET instrumentation.

The classic design of clinical PET scanners put a stop to the fundamental limits of efficiency and resolution from being realised [14]. However, with new technologies changing the set of compromises imposed on the PET camera design [14], the ability to manufacture a small volume PET scanner with excellent spatial resolution while maintaining or also improving sensitivity is on its way.

To enhance the medical capability of a PET camera with a high intrinsic spatial resolution, a high overall sensitivity, uniformity of characteristics throughout the FOV, adequate sampling without detector motion and large throughput capabilities for high rate dynamic studies, are important design criteria [11].

Therefore, for all the obvious reasons, new technologies and approaches must be considered, since no current design based on PMTs can achieve all these goals simultaneously [15].

### 2.2 New Approaches To Measure Depth of Interaction (DOI)

Traditionally, silicon photodiodes (PDs) have been noisy and this along with their unity gain has ruled them out as being possible replacements of PMTs as scintillation light amplifiers in PET. However, recent developments
in the design and manufacturing of silicon PDs have radically changed their inherent characteristics. It is now feasible for custom-made low noise PDs to be able to replace PMTs in PET scanners.

**2.2.1 Advantages of Si Photodiodes**

For Si photodiodes to replace PMTs in PET detector modules, their advantages must by far outweigh their drawbacks. In recent years there have been great strides taken in the ability to optimise the characteristics of Si photodiodes to closely match the requirements sought after.

The small physical size of PDs enables two major features to be explored. Firstly, PET detector modules that have PDs instead of PMTs are much less bulky so the configuration of modules can be more closely packed. And secondly, because the charges generated within the detector have only a correspondingly short distance to be collected (in an appropriately designed device), their response time is comparable to PMTs [16]. In short, Si photodiodes allow the constraints of classical PET scanner design to be relaxed and this allows necessary space for innovation to occur.

Other advantages of Si PDs in their application to PET detector module are: they are robust, require usually less than a 100 $V$ operating voltage in so doing away with more expensive high power supplies; unlike PMTs, they are insensitive to magnetic fields; their very high QE (50-80% in visible spectrum) strengthens the signal bandwidth product; the gain of photodiodes is fairly independent of temperature and they are relatively inexpensive to produce compared to PMTs [17].

On the other hand, photodiodes have drawbacks such as their unity gain; current commercially made PDs have low QE at visible and UV wavelengths so at the scintillator emission wavelengths involved in PET, this makes them very sensitive to electronics noise and places limits on energy and timing resolution when measuring gamma radiation [18]. Also, another critical factor to be considered in the design of the PD device is the entrance window. To maximise the transmission of the scintillation light into the...
active volume of the detector, this window must be kept as thin as possible [19]. And the decision to use anti-reflective-coating (ARC) to tune the incidence wavelength (of the scintillator) to that of maximal spectral response for the PD is part of the design process for an optimal detector.

In comparison to PDs, PMTs have low noise and are fast. However, their gain is temperature dependent and drifts with time. They require kilo-voltage power supplies and are sensitive to relatively small magnetic fields more than a few gauss. They possess low QE at around 20% but perhaps their biggest negative is that at approximately 8 cm deep, they are bulky and place a troublesome size constraint on the PET design making it difficult to read out closely packed arrays of small scintillator crystals, thus limiting the spatial resolution improvements that are constantly demanded [18].

2.2.2 Novel Attempts To Measure DOI

One advanced PET module design that attempted to accurately encode the DOI measurement that would eliminate radial elongation blurring involved the use of a silicon photodiode placed in front of the BGO scintillator. Eliminating this blurring would considerably improve the spatial resolution. With the recent availability of lower capacitance photodiodes has allowed the opportunity to tentatively measure for DOI in the PET detector module, without, most importantly, introducing unacceptable levels of electronic noise.

The Moses et al PET detector module (as shown below in figure 2.2.2) takes advantage of these low noise photodiodes. By processing a portion of the detected scintillation light by the PMT and (ideally) the remainder of the scintillation light by the PD element coupled to the crystal [20], this detector attempts to derive an accurate DOI.

The PMT provides a timing pulse (accuracy ~ 2 ns FWHM) and initial energy discrimination (typical Energy resolution for 511 keV is 17% FWHM and threshold is 250-350 keV) [20] and the photodiode (PD) identifies the crystal of interaction [21].
In the earliest and most simple version of the detector module, the PDs are only used to identify the crystal of interaction [20]. Although in the later, more advanced module, the PDs also assess the position of the 511 keV interaction in the BGO crystal (in the 30 mm long dimension) by comparing their pulse heights to those observed in the PMT [20].

This detector module is comprised of 64 (3x3x30mm³) BGO (and then later with LSO on the later model) scintillation crystals coupled on the front with a 64 pixel array of 3 mm² Si PDs and on the rear to a single PMT [21].

The PD generates its signal from the scintillation photons that strike it. As the signal is dependent on the crystal depth and the noise is independent of depth, the SNR and therefore the DOI resolution is dependent on depth [21]. The best SNR is necessary for accurate positioning and is obtained when the scintillator used provides the most steady and the highest output possible [4,21].

Scintillation light from a 511 keV annihilation photon in a BGO crystal makes approx 1800 e-h pairs (0.28 fC or 0.28 x 10^{-15} C) in the PD [22]. This is a small signal and requires amplification by charge amplifier with a very low noise FET input stage [22]. The Si PIN PD array has a 300 µm depletion
thickness, 300 pA dark current per pixel and each element has a 3 – 6 pF capacitance (measured at the perimeter of the array) with elements towards the centre having higher capacitance due to their longer lead length and 73% quantum efficiency (QE) at 415 nm [21].

Summing the PD and the PMT provides the total energy pulse and the ratio PD/(PD + PMT) theoretically allows the DOI to be calculated [21]. Due to its high detection efficiency, correct identification of the crystal of interaction is carried out only 79 ± 4% of the time [21]. To accurately determine the DOI this percentage needs to be greater.

For a single LSO detector module, the timing resolution is 750 ps FWHM its pulse height resolution is 24 ± 3% FWHM for 511 keV photons and its DOI measurement resolution is 8 ± 1 mm FWHM [21]. Although this design realised a spatial resolution improvement over the conventional PET module (due its smaller 30 mm x 3 mm² crystals) [29], it did not achieve a precise way of obtaining the DOI because of the statistical uncertainty in measuring the depth of interaction in each crystal.

### 2.2.3 PET Detector Modules With Avalanche Photodiodes

The foremost reason for the growing interest in the development of new detectors and detection schemes for high spatial resolution PET is that the bulky size of currently available PMTs makes it difficult to achieve close packing of detectors in arrays [24]. Since reducing the size of the detector helps to enhance spatial resolution [25], the inability to pack detectors closely together rules out the opportunity of taking this pathway to enhanced spatial resolution – thus new detector designs are continuously being considered.

Improvements in the spatial resolution of PET systems are being attempted by reducing the size of the detector. However, because of their added complexity and their high cost per channel, most of these designs are shelved [24]. If smaller detectors that allow a close packing fraction can be
implemented and still fall within the energy and timing requirements for PET, then breakthroughs in improving spatial resolution would be more likely.

One possible device that can replace PMTs in certain applications is the avalanche photodiode (APD), because of its internal gain, fast timing and high photon detection efficiency [26]. Although current commercially produced Si PIN PDs demonstrate sufficient energy resolution for PET, they are unfortunately not able to offer the sufficiently high coincidence discrimination required for PET [24]. However, small discrete scintillator crystals based on APDs have been found to attain uniform, isotropic and high spatial resolution and when linked with parallel processing enable low deadtime and high count rate capability [27]. Si APDs with their high quantum efficiency, low dark noise, fast response time and small size offers the possibility of challenging the constraints that limit spatial resolution in current PET systems [24].

APD Principal Operating Mechanism

Radiation incident on the device generates e/h pairs in the same way as a photodiode. The relatively small numbers of e/h pairs created in the radiation event are then internally amplified by an avalanche process that takes place in the junction because of high voltages involved [28].

High electric fields accelerate the charge carriers to the point where their collisions within the semiconductor create additional e/h pairs along the collection path [28], see figure 2.2.3 below. Thus, the avalanche photodiode boosts the signal much further above what is possible using conventional photodiodes thus enabling good energy resolution even at low radiation energies [28].
When using scintillators, APDs offer a number of advantages over PMTs, namely their compactness, ruggedness and flexible geometrical configuration [17,26]. They have low dark current noise and highly effective QE in the 400 - 600 nm range [29]. They have fast response times, insensitivity to magnetic fields due to their short electrical carrier paths [30], low power consumption, have simple low power biasing requirements [24] and are relatively less costly [17]. The internal gain possessed by APDs is desirable for use in converting low energy scintillation photons into an electrical signal because this gain amplifies only the signal and not the noise produced by the electronics [28].

Another advantage of the APD mechanism is the noise due to surface dark current, which comprises the largest percentage of the dark current noise, is not multiplied, while the multiplication of the bulk dark current takes place, this is regarded as a negligible amount [31]. Thus this internal gain is an attractive way to overcome the problem of amplifier noise and deriving an excellent SNR [32]. This benefit also comes at the cost of initiating statistical noise due to the multiplication processes [28].
The time resolution of large area APDs have been found to be only three times worse than a fast PMT and forty times better than with Si photodiodes, while also achieving good energy resolution [26]. This good time and energy resolution allows the use of a small coincidence window that decreases the number of random events in PET [25]. The energy resolutions of APDs at certain wavelengths are found to be superior to those acquired with PMTs [30]. APDs achieve good timing properties because the increased electric fields speed up carrier collection [28]. APDs have an advantage over PDs in that faster shaping times can be used because of their lower noise [26].

**APD Drawbacks**

To attain uniform gain throughout the detector volume, the design of the detector geometry is a foremost consideration [28]. Careful detector design is also required to reduce the likelihood of breakdown due to high surface fields [28].

Because small variations in applied voltage have an amplified effect on the gain, well-regulated high voltage supplies are essential to maintain an even voltage [28]. APDs have also been found to exhibit internal gain dependent on temperature [17]. The variation in gain can be as much as 10-20% per degree Celsius [32]. Temperature stability can be ensured by the use of thermo-electric modules [27], however this adds to costs and makes further complications.

Lecompte et al [14] built a PET scanner that attained a nearly isotropic volumetric resolution of the order of 0.015 cc FWHM. This fact verifies that replacing PMTs with solid-state photo-detectors to undertake the read-out of scintillator crystals and achieving fairly uniformly high resolution PET is now a firm reality. The overriding question to be asked though: is this development head and shoulders above what current clinical scanners can produce and what cost?
2.2.4 Attempt To Measure DOI With Position Sensitive PMTs

Further attempts to improve accuracy measurement of the DOI involved the use of a “Position-Sensitive” PMT coupled to BGO arrays having small grid segments.

The advent of PS PMTs naturally led to their application to PET detector modules so their capability of determining the DOI could be evaluated. One other PET design attempted to maximize both the amount of light and the effect of collection while retaining the advantages of a compact design in that they assembled a detector module composed of LSO crystals coupled directly to a miniature metal can PS PMT [33].

Principal Of Operation Of PS PMTs

Positioning of the scintillation flash is undertaken by collecting current from the two crossed anode layers, each layer consisting of four side-by-side anode plates separated by 0.5 mm [33]. The current distribution across the first layer establishes the x-coordinate and the second layer – the y-coordinate. In addition, an additional signal from the last dynode plays a part in timing and energy discrimination [33]. Thus, the comparative magnitudes of each layer determine the position of the scintillation flash within the FOV [33]. In this way, the crystal of interaction can be identified but this method is not successful in measuring the DOI.

Drawbacks With PS-PMTs

Unfortunately, PS-PMTs are notorious for their “edge effects” that results in significant photocathode non-uniformity [5]. On the periphery, there is a gain drop invoked by electron loss at the edge of each dynode because of electrons spreading out during the cascade multiplication [10]. The gain at the periphery can be raised, however, it involves mounting grids for local gain control in the tube [10].

The other edge effect causing non-uniformity is due to the response of the peripheral wire anodes differing to that of the central anodes and without
complicated electronics to read each anode individually, a strong distortion occurs [10].

As well as having a large non-photosensitive areas at the perimeter, PS-PMTs have significant problems with cross talk and are much more expensive than conventional PMTs [17,18]. In addition, one of their greatest drawbacks is that they have a large dead space surrounding the photodetecting surface and this brings about packing fraction problems [5].

While these approaches offer some improvement in spatial resolution, the bulky size of PMTs greatly restrict designs that enable the ability to measure the DOI. In short, even when expensive and complex PS PMTs are integrated into the PET module, the DOI that is necessary for high resolution PET, is still not achieved.

2.2.5 TOF Positron Tomography

Scintillators with extremely short decay constants offer the prospect of Time-of-Flight (TOF) tomography, enabling the positioning of the annihilation event’s location. By using the speed of light constant (3x10^8 m/s), the known distance and line between two detectors and the time difference in the detection of the photons, the location of where the annihilation event took place can be determined [6,34]. Measurement of the difference in the arrival times of the detected 511 keV photons has the potential to enable a PET scanner to restrict the location on the annihilation event to a subsection of the LOR [35].

For example, a 1ns time difference in the detection time corresponds to a 15cm spatial difference in the position of the annihilation event’s localisation [34]. The positional information determined in this way has an associated uncertainty dependent upon the statistical limitations of the detector [34]. Thus we acquire a Gaussian-shaped distribution for a positron source located between two detectors, with the mean located at the source of the positrons [34]. The positioning uncertainty is measured at FWHM and
reflects the width of this distribution of statistics [34]. If this uncertainty is in someway reduced to less than 1cm then direct three-dimensional positioning is possible. Significantly, this enables the reconstructive process to be bypassed, therefore removing the tendencies of the process of current commercial PET systems that amplify the noise in the image [34]. With the automatically decreased noise in the image, its advantages are obvious.

Using BaF$_2$ and CsF scintillators, several TOF PET cameras have been built. These CsF cameras typically obtain approximately 500 ps FWHM coincidence time resolution [35], and gives an spatial uncertainty of 7.5 cm [34]. This resolution is not sufficient for direct 3D positioning but it is nonetheless effective in aiding the reconstruction process for an improved image quality [34].

However, TOF PET cameras based on these scintillators have not seen widespread use specifically because the UV emissions of BaF$_2$ require the use of relatively expensive quartz windowed PMTs. Furthermore, the lower density and PE fraction causes a degree of misaddressing of positions of interaction brought about by the penetration of the annihilation photons into the neighbouring detector crystal before interacting. This contributes to the radial elongation effect and causes deterioration of the spatial resolution [35].

The spatial resolution of an annihilation event is presently limited by the size of high quality PMTs necessary for sub-nanosecond timing desired for TOF PET [1]. The advent of the scintillator LSO has at least offered hopes for its use in TOF PET because it has a rapid decay constant and gives good stopping power. However experimental evidence demonstrates its coincidence time resolution in one experimental set-up measured between two 3x3x30 mm$^3$ crystals of 475 ps FWHM, which is much worse than the predicted 300 ps FWHM [35].

This degradation in timing resolution is speculated as being caused by the scintillation light undergoing multiple reflections at quasi-random angles within the crystal [35]. The effect of these multiple reflections is thought to
reduce the effective light propagation speed within the crystal to an effective refractive index of between 3.9 and 5.3 and the randomness of the reflection angles (and thus its path lengths) brings about a dispersion effect and therefore a 10 – 90% increase which adds about 1ns to the optical signal [35].

Thus, interactions that take place at different locations within the scintillator lead to differences in the time taken for the scintillation light (carrying information about the event) to travel from the annihilation event to the PMT and directly increases the coincidence resolving time [35]. The corresponding magnitude of this effect is proportional to crystal length [35].

As scintillator photons propagate more slowly than the incident annihilation photons, this causes TOF tomographs to suffer from a loss in timing resolution. As a result, the timing resolution for long crystals is significantly worse than for shorter crystals. Consequently, there is a resulting trade off between timing resolution and detection efficiency unless the DOI can be accurately measured. When an accurate DOI can be obtained, the timing resolution can be naturally improved for longer, more efficient (has more opportunity to stop photons) crystals [1].

In summary, it is quite evident that as the accuracy of the DOI measurement improves so does the capacity to undertake precise TOF measurements. This will in turn make for less noisy images with higher spatial resolution. With currently used scanners, the application of the TOF technique to PET scanners is fairly non-productive. However, with the novel PET scanner under review herein may offer very accurate DOI measurements and the TOF technique would give in this case the potential to remove noise from the image without deteriorating the intrinsic resolution.

2.2.6 Scintillator Read-Out With Photodiodes

Higher energy radiations can be measured by employing scintillators to convert input radiation into lower energy photons that can easily by detected by a photodiode where it creates a current pulse. We can only approximate
how many scintillation photons were actually created within the crystal – this current read off the scintillator is the only hard quantity we can derive from the incident radiation [36].

Inorganic scintillators are used in PET to introduce annihilation gamma photons, emitted from the patient, into the detection system. Thus the functioning of the PET camera is strongly coupled to the physical and scintillation properties of the crystals [6].

Without there being a scintillator with high stopping power and high density to detect the 511 keV gammas used in PET, the photodiode by itself would have very poor detection efficiency.

Figure 2.2.6. Scheme to illustrate activated scintillator mechanism (such as that in NaI(Tl) and LSO(Ce)).

A typical scintillator is a transparent single crystal in which a band gap of 5 eV or more separates the valence and conduction bands [6]. See figure 2.2.6. Incident gamma rays are absorbed by the bulk of the scintillator and a fraction of this energy localises on the activation ions [6]. This activation site eventually relaxes and so emits scintillation photons that typically (with NaI(Tl), for example) have about 4eV, which corresponds to visible blue light [6].

The conversion of incident radiation to emitted scintillation photons takes only a matter of nano- to micro-seconds [6]. The intensity of the
generated pulse is ideally proportionate to the incident radiation energy. It becomes clear that when the scintillation pulse is more intense and is emitted at the wavelength in which the photodiode is most sensitive, then the optimal signal is produced. A shorter decay constant also allows better energy distribution and gives the option of improved count rate capability and tighter coincidence timing [6]. Because of the importance of this decay characteristic with regards to PET a figure of merit of number of photons emitted from the scintillator per nanosecond is used [6].

In PET, a scintillator ideally requires:

(a) Excellent 511keV photon detection efficiency to reduce radial elongation, shorten scan times and maintain low tracer activity and to detect as many of the emitted gammas as possible [6].

(b) Good detection efficiency also ensures fine coincidence measurements, good energy resolution to decrease events put down to Compton scatter in the body and good timing resolution of reduce random coincidences and dead time [6].

(c) Low cost because PET scanners may use up to 10,000 cc of scintillator crystals [6].

(d) And the quality of possessing the mechanical ruggedness to make manufacturing of the crystals easier and presumably cheaper.

For high resolution PET, the decay constant of the scintillator emission becomes important because to minimize the effect of singles count rates that are typically high, the coincidence resolving time ideally should be as small as possible to reject unwanted random events thus a scintillator with short decay constant is advantageous in both cases [6]. See table 2.2.6 below.

Annihilation photons principally interact with matter via photoelectric (PE) absorption and by Compton scattering (CS). One basic parameter for scintillator performance is its photofraction. This is defined as the number of counts in the photopeak region divided by the total number of counts [12] and is important in obtaining the best image because it is better to have the interaction between the gamma photon and the scintillator take place at one
unambiguously localised centre to allow a truer LOR to be formed and thus enable high spatial resolution [8].

For example, Lutetium Oxyorthosilicate (LSO) has a lower atomic number ($Z_{\text{effective}}$ is 66) than BGO ($Z_{\text{effective}} = 75$) [12]. This means that for both annihilation photons to be recorded in that type of scintillator, BGO has a 60% higher probability of registering two PE interactions than LSO [8].

Although LSO has near four times the light output of BGO and a five times faster decay constant, Cherry et al [8] speculate the reduced likelihood of PE interactions in LSO would lead to worse spatial resolution and or lower efficiency.

<table>
<thead>
<tr>
<th>Scintillator Material</th>
<th>BGO (Bismuth Germanate)</th>
<th>LSO (Lutetium Oxyorthosilicate)</th>
</tr>
</thead>
<tbody>
<tr>
<td>chemical formula</td>
<td>Bi$_4$Ge$<em>3$O$</em>{12}$</td>
<td>Lu$_2$SiO$_5$:Ce</td>
</tr>
<tr>
<td>atomic number</td>
<td></td>
<td></td>
</tr>
<tr>
<td>decay time (nS)</td>
<td>300</td>
<td>40</td>
</tr>
<tr>
<td>Light output (photons/MeV)</td>
<td>8,200</td>
<td>30,000</td>
</tr>
<tr>
<td>Attenuation length (cm)</td>
<td>1.1</td>
<td>1.2</td>
</tr>
<tr>
<td>Photoelectric Fraction</td>
<td>43%</td>
<td>34%</td>
</tr>
<tr>
<td>Emission Wavelength (nm)</td>
<td>480</td>
<td>415</td>
</tr>
<tr>
<td>Index of Refraction</td>
<td>2.15</td>
<td>1.82</td>
</tr>
<tr>
<td>Radioactive background</td>
<td>no</td>
<td>yes</td>
</tr>
</tbody>
</table>

Table 2.2.6. Comparison of the scintillator crystals LSO and BGO.

However, in spite of the lower atomic number of LSO, the photofraction of small crystals (2x2x10mm$^3$) of LSO and BGO were measured (with PMTs) to be equal due to the greater light output that allowed higher energy resolution of LSO [12].

Whenever radiation creates light photons within the scintillator, there are many effects that can bring down their number before they are finally
converted to e/h pairs in the Si PD [37] and these factors play a major part in obtaining good statistics.

### 2.2.7 Extracting Light From Scintillators

Factors that affect the efficiency of output of the scintillator are: the physical dimensions of the crystal and what levels of self absorption take place; the quality of the reflector material and the surface surrounding the scintillator; how well the emission spectrum of the scintillator is matched to the photo-sensitivity reflectivity of the photodiode and how efficiently the generated e/h pairs are collected [37].

To have the greatest output it becomes necessary that the scintillator should have minimal absorbance at its emission wavelengths so the scintillation photons can easily exit the crystal [38]. Then the transmission of the scintillation photons will be clearly optimised when the refractive index of the crystal is as close to that of the photodiode entrance window (and coupling glue) [6].

### Crystal Size And Geometry

When determining the optimal crystal geometry, there is an substantial trade-off between spatial resolution and the desired volume of scintillation light to provide the necessary ingredients for a good SNR [8].

The shape of the crystal and the energy deposited within it are factors that influence the timing property of the scintillator [35]. Whenever multiple reflections take place, the time required for that information to be emitted is dispersed because of the varying path lengths that scintillation photons take, even though they may have been created at the same location [35,39]. The magnitude of this effect becomes greater with increasing crystal size [35]. However, if the height were to be decreased, the diameter of the dispersion of scintillation photons upon the photo-detector would consequently decrease leading to the possibility of better DOI measurement resolution.
It has also been found that rounding the edges of square or hexagonal crystals have improved light output by ~5% [21].

**Qualities Of Crystal Surfaces And Use Of Reflectors**

Surface preparation of the crystal plays an important role in the complicated problem of light collection efficiency and the gaining of maximum light output necessary for an improved SNR [21,40]. The results obtained by a Monte Carlo (MC) simulation by Bird et al suggest the most advantageous surface preparation is to polish the sides and roughen the face opposite the PD [40]. They confirmed this with laboratory tests and then discovered even better results then the MC predicted when the crystals had been polished on all sides [40]. This result was put down to the MC assumption that the wrapping material had a lower reflectivity then what it physically possessed [40].

Light collection efficiency is dependent upon how thoroughly the crystal surface is polished. Very polished crystals tend to trap more scintillation light because of internal reflections against the side and refraction mismatches at the interface [8]. Efficiency can be enhanced by the providing ground surfaces and by the use of external reflectors but there are losses involved with the interaction of the scintillator light and the reflector, consequently gaining only a slight improvement in collection [8]. Cherry et al predicted the optimal light collection efficiency would occur using a crystal that has highly polished side, a ground top and is wrapped in a diffuse reflector [8]. The concept behind this is the ground surface breaks up the internal reflection from the polished sides and redirects the light to the face of the photodiode [8], consequently improving the SNR.
The differences between simulations and observations are most likely due to the perfect plane surfaces assumed in simulations while in practice polished surfaces will exhibit microscopic irregularities [8]. These tiny irregularities found in practice may be enough to reduce internal trapping [8].

2.2.8 Innovations Now Possible With Lutetium Oxyorthosilicate (LSO)

LSO(Ce) was discovered in 1991 [41] and offers possibly the best combination of properties of any scintillator today [6]. With high light output of 25,000 photons/MeV [18], unlike BGO, LSO allows the application of many small detector elements per photo-detector [6]. LSO has a fast decay rate of 12 and 42 ns, with relative intensities of 35 and 65% respectively [25] enabling a more accurate timing signal to be generated [42]. In effect, this allows LSO to offer a narrower coincidence-timing window of 4-5 ns [25] in PET leading to an increase in the counting rate [42] and a decrease in the random to true coincident rate [12]. This high number of scintillation photons per nanosecond (25,000 photons/30 ns) makes LSO very attractive when coupled with Si PIN photodiodes [43] even though its emission spectra (420nm) is not an ideal match to Si [6].

LSO has a high density (7.4 g/cm), and an atomic number (Z_{effective}) is 66 [12] that give it short attenuation length (1.2 cm) and high PE fraction (34%) [18] resulting in good annihilation photon detection efficiency [6]. Being mechanically robust and non-hygroscopic also allows construction of LSO detectors to be relatively easy [6].

The presence of naturally radioactive $^{176}$Lu in LSO results in an intrinsic count-rate of 300 s$^{-1}$cm$^{-3}$ [48] but since this number is small compared to the typical counting rate from injected tracers in PET, this is not give a great affect to accuracy [6].

Compared to other scintillators, LSO has low refractive index of 1.82 [38]. This in turn causes a relatively low reflection level at the crystal
detector interface and results in a higher fraction of the scintillation light to reach the detector and be converted into a current signal [38].

LSO(Ce) has the high price of ~US$100 cm$^{-3}$ due to Lu and integrating LSO into a large PET scanner will easily take costs to the US$10^6$ level [41]. Unless this scintillator or one similar to LSO, can be manufactured more cheaply then the outright cost of LSO alone may prohibit its use in PET scanners.

There are inconsistencies in the volume of scintillation photons generated in LSO ranging from 20,000 – 25,000 ph/MeV [44]. It is speculated that this excessive variance in photon numbers is responsible for their disappointing energy resolution [29]. An energy resolution for LSO of 4.5% FWHM has been obtained from a $^{137}$Cs (662 keV photons) source, it is more regular to attain energy resolutions in the range of 10-20% FWHM [41].

One chief reason for poor energy resolutions is that it is hard to manufacture large size crystals with homogeneous scintillation light output [41]. There is strong non-linearity in light yield as a function of incident gamma ray energy [29]. Thus it becomes difficult to select LSO crystals with the desired decay time and high light output because of these excessive variations in scintillation characteristics from crystal to crystal [43].

In summary, the short attenuation length, high light output and fast decay of LSO, despite the somewhat poor energy resolution, permit the use of smaller detectors with improved timing properties and increased count rate for enhanced PET applications [38].

### 2.2.9 Error Analysis In PET

A high resolution PET detector module preferably incorporates:

1) Maximum detection efficiency [17]

2) High spatial resolution (less than 5 mm FWHM) [17]

3) Low parts cost [14]
4) Minimal deadtime [14]
5) Excellent timing resolution (< 5 ns FWHM) to bring down the accidental coincidence rate that is proportional to the square of the single photon rate [17] and
6) Good energy resolution [17]

Detection Efficiency

Usually expressed in terms of the percentage of incident photons of given Energy (or per MeV) detected by the module, the detection efficiency is the likelihood that incident photons interact and are partially or fully absorbed within the scintillator crystal consequently generating an electrical signal pulse at the output of the detector [6,17].

Typical numbers of 511 keV gammas impinging on each detector pair of detectors in a whole body scanner may be as high as 50,000 per second [34]. And of the 50,000 gammas, usually not more than 0.2% (ie. 100 gammas) are detected in coincidence by the electronics [34]. Coincidences that are detected are not all “true” coincidences and have two other components called scatter and randoms [34].

Scatter is produced by the Compton interaction of the gamma with the tissue matter in the torso and the outcome is a secondary gamma that has lost its direction and an amount of its energy [34]. Randoms are brought about by the chance detection of two unrelated gammas in the coincidence electronics.

Typically, in conventional systems the contribution of the scatter and randoms to the total coincidence events detected may be 20% for the scatter and 25-30% for the randoms. Therefore, only about 50% of the total coincidences detected are “true” coincidences [34].

The detection efficiency of “true” coincidences from a typical patient for one pair of detectors could be as low as 0.1% of all the gammas impinging on each detector [34]. Statistically then, a PET scanner is literally starved for true coincidence photons [14]. To improve these statistics it is found that adding several layers of detector rings around the subject to be studied such
that the number of detector pairs in coincidence is greatly increased has been one solution to this low detection efficiency [34]. Since the radioactivity that is injected in a patient may be diffused throughout the organ of interest, it is important in dynamic studies to quickly collect information about the studied organ, preferably all of it simultaneously. Heightening the overall detection efficiency allows reduction of the total scan time or the lowering of the dose to the patient [34]. This improvement in overall detection efficiency is however offset by a possible increase in the scanner because many more detector modules may be required.

**Spatial Resolution**

Ideally the spatial resolution would not vary over the whole FOV of the tomograph. Non-variation requires accurate depth and position of interaction within the crystal of interaction [17].

The radial elongation artifact is caused by the 511 keV gammas penetrating into adjacent crystals before undergoing an interaction [39]. The narrower the scintillation crystals are cut, the more this blurring artifact deteriorates the reconstructed image [39]. The image blurring prevents the edges of larger structures in appearing as clean lines and may impede the representation of smaller ones as distinct objects [3]. Image blurring also smears neighbouring areas and averages them together resulting in loss contrast [3].

The capability of being able to measure DOI and having very high count rate performance becomes more important in smaller animal/brain PET scanners. Due to the smaller diameter, the count rate is proportionally higher and because the subject fills virtually the entire FOV the geometrical effect of radial elongation is more dramatic. Placing the detector modules closer to the patient port requires detectors with very high count-rate performance to minimize dead time losses [14]. The patient port nearly fills up the detector ring, so severe radial elongation artefacts will be present unless the DOI can be measured [14].
Degradation in spatial resolution for sources distant from the centre of the FOV in characteristic of all ring based PET detector systems [45]. This non-symmetric broadening of the point source profile is a result of assigning events to the wrong LOR [45]. This spat broadening is primarily a function of crystal length and the angle of incidence of the photons [45].

To attain a more uniform resolution throughout the FOV, it is necessary for the detector system to provide accurate DOI information so the correct positioning of these events takes place [45]. Correctly measuring the DOI also improves the performance of a PET detector system by enhancing the sampling characteristics and thus allowing the introduction of smaller ring diameters to increase coincidence sensitivity [45].

PET designs that permit DOI measurement capability allow the detector diameter to be significant with no corresponding degradation in spatial resolution.

**Effect of Detector Deadtime**

Each time a 511 keV photon interacts within any crystal of a module that is connected to a PMT, that module is “dead” for the time it takes for the amplifier to shape that signal [7]. This means the time required to process a single event limits the PET scanners count-rate. Thus deadtime is the principal factor restricting the injected dose [7].

While randoms are negligible at low count rates, the number of randoms increases as the square of the activity until the number of true events starts to saturate the detector. As the detector saturates, this causes deadtime to become significant [7].

**Timing Resolution**

Events due to unrelated positron decays occurring within the coincidence window or randoms are the major source of noise in PET. To minimize this component, the coincidence time window must be made as small as possible
to filter any events due to unrelated positron decay. The time resolution of a
detector is then an important parameter in keeping noise to a minimum [36].

**Detector Bulkiness Problem**

When a positron interacts with an electron, a pair of annihilation photons
is the ultimate result. In accordance to the law of conservation of energy, the
two emitted photons from this annihilation, endowed with energy equivalent
to the combined mass of the positron and electron, are spat in a collinear,
back-to-back manner with energy of 511 keV each. However, due to the pre-
annihilation particles not being at a complete standstill and therefore
containing momentum (that must be also conserved), the angles at which the
annihilation photons are emitted are not exactly at $180^\circ$. Thus, the main
difficulty in designing a PET scanner with very high spatial resolution lies
not in the statistically varying range of the positrons or in the non-
collinearity deviations in the $180^\circ$ back-to-back annihilation photon
emission, or in the detection properties of small scintillator crystals but the
difficulty is due to the size of the available PMTs [1]. Thus some novel
approaches have taken form using various configurations of light pipes and
new styles of PMTs [1] and other photo-detectors.

2.2.10 Overcoming Classical PET’s Drawbacks

Various approaches to gain accurate DOI measurement from PET
detector modules have been attempted but still no current PET system offers
this important feature. One major weakness in a number of the proposed
methods is the requirement of additional detector electronics [45].

The design being built by CMRP clears away many of the obstacles
hindering DOI measurements in currently proposed PET scanner systems.
Implementing low cost, small size, high QE and rugged Si PDs makes them
worthy of testing in high resolution PET that demands good spatial, energy
and timing resolution [18].
The CMRP’s new scanner incorporates to a high degree a strong solution to advance high resolution PET in these areas:

a. high efficiency
b. has DOI for improved spatial resolution
c. lower costs by doing away with PMTs, except for LSO
d. low dead time due to onboard Viking signal processing
e. good energy resolution with high SNR
f. good timing resolution: from LSO fast decay and also because of the detector’s unique bonded, front loaded PD/electronics set-up (Parallel Connection of two or more detectors that is not a great concern since each PD will be connected to its own preamp on the Viking IC chip [46].

The small ring diameter, septa-less operation, and the low dead time of LSO and the capability to accurately measure DOI give this design the ability to perform very high resolution PET with high sensitivity.

Importantly, sensitivity and count rate performance have not kept pace with the improvements in spatial resolution, and many clinical PET studies are statistics rather than res limited [5]. Ultimately, key performance parameters such as lesion detectibility or quantitative accuracy and precision depend on the SNR or total ERROR [3]. Obviously, a focus on SNR implies equal weighting to both components [2].
3.0 CMRP Approach To High Spatial Resolution PET Modules

This project presents a complete modular design that can be used to realise a variety of PET camera configurations. Innovations that make this new PET scanner feasible are the low noise, high efficiency Si PIN photodiodes and the novel use of the high speed Viking IC readout.

The custom-made photodiodes used in this scanner are designed to take advantage of LSO’s evidently ideal characteristics (as discussed in the previous chapter) for PET. And the Viking IC provides the signal processing capability that is required for the scintillator readout. As timing and energy resolution is of main concern for the detector modules, the electronic noise produced in the readout module (which includes the Si PD and the Viking IC) must naturally be kept to a minimum.

3.1.1 Si Pixel Detector Design

One approach to obtain accurate 2D positional information from a single sided Si photodiode is to construct the top electrode as a pattern of individual small elements (pixels) that are each electrically isolated. There are advantages but also challenging drawbacks that are to be traded-off when fabricating such a device.

The advantage of smaller pixels is their relatively small leakage current and capacitance due to their physical size [47]. However, It must be kept in mind that the greater the density of electrical contacts and wiring used, the smaller the active area there is available to use for detecting events. As an electrical connection must be made to each pixel with a readout connection for each pixel, it directly means that the more pixels there are, the more wiring is necessary, along with the greater need for more powerful processing electronics to cope with the larger volume of recorded data.
3.2.1 Theory Of Photodetectors

When n-type material is doped (in this case by the ion-implantation technique) to produce a p-type region, a p-n junction is formed.

Majority carriers in the p-type material (holes) flow due to diffusion towards the n-type material that has a lower concentration of holes. This diffusion down the hole concentration gradient exposes the holes to excessive numbers of electrons in the n region and thus recombination takes place there. Likewise, electrons, the majority carrier from the n-type material, will similarly diffuse towards the p region and recombine with holes.

The result of this charge diffusion is the p-region becomes depleted of charge accepting holes and the n-region depletes of excess electrons. This effect brings about a layer of fixed negative charge on the face of the p-layer and similarly a layer of fixed positive charge form in the n-layer. The ionised centres (both donors and acceptors) fixed in the junction forms a space charge region (depletion layer) [48].

Equilibrium is then maintained by equal number of donors and acceptors present in this “depletion layer,” in consequence of the Fermi level being constant throughout the entire material.

The electromotive force between these layers invokes an electric field across this depletion layer, as shown in figure 3.2.1 below. Here in this region, the contact potential stops any majority carriers from flowing.

Although some of the more energetic holes and electrons may diffuse across the depletion region, the overall current balance is zero. This is because minority carriers compensate for the majority carrier’s drift by minority holes drifting through the electric field from the n-region and minority electrons drifting from the p-region [23]. Now there are few (if any) free carriers in this presently high-resistivity region.
When incident light strikes the detector, electrons within the crystal structure become stimulated. If their energy is greater than the band gap energy $E_g$, the electrons are elevated to the conduction band and holes are consequently left in the electron’s place in the valance band [32]. Ideally, the number of electron-hole pairs generated per ionisation event is proportional to its incident energy (mean energy $E$ at room temperature to produce an e/h pair = $3.5 \pm 0.1$ eV in Si. For example, the number of pairs produced by a 5MeV particle in Si is $1.43 \times 10^6$ [49]. The electric field brought about by the potential difference latent in the p-n junction, causes the separation of generated e/h pairs and freights the charge carriers towards their respective electrodes.
For a radiation detector to function nicely in practical settings, the depletion layer is widened by applying a reverse bias across the p-n junction and this subsequently increases the detector’s sensitive volume. The reverse (or back bias) voltage strengthens the electric field between the layers and by lowering the energy levels of the n-type region with respect to that of the p-type region, thus pulling it wider. While there is no leakage of majority carriers under reverse biasing, the diffusive flow of minority carriers across the depleted region stimulates a leakage current. This leakage is fairly independent of the applied voltage [49].

3.2.2 The Depletion Layer Thickness

To determine the extent of the depletion layer for a given bias we can solve Poisson’s equation to relate electric field potential $V_x$ at any point $x$ in the depletion layer with the volume charge density at that point $\rho_c$, the derivation evaluates it to, for n-type Si [23]:

$$d = 5.5 \left( \frac{\rho_n V}{V} \right)^{\frac{1}{2}}$$

where: $d$ is the depletion depth in $\mu$m and is limited by the physical thickness of the wafer. $\rho_n$ is the resistivity of the n-type material in $\Omega$m and $V$ is the bias voltage applied.

The field strength rises linearly toward the front contact where it attains its highest value. When the detector has an applied bias in excess of that required for full depletion, that is when $x = W$ (see figure 3.2.2 below), it is said to be “over-biased.” Over-biasing the detector creates a more uniform field. Therefore to ensure a high uniform field that covers the entire sensitive volume of the detector, the detector must be over-biased [56]. However, there are trade-offs with over-depleting the detector, especially with respect to timing applications. This aspect is discussed later in this section.

If the incident radiation event takes place outside the depletion layer, the efficiency in the collection of the e/h pairs produced is very poor because many of the charges will recombine or be trapped. However, some charges may be collected and will contribute to the signal. Therefore any entrance
window, in which an interaction is likely before the depletion region, will act as a “dead layer” and must be thin [53].

![Electric field configurations in a p-n junction detector. Taken with modifications from Delaney & Finch [23].](image)

**Figure 3.2.2** Electric field configurations in a p-n junction detector. Taken with modifications from Delaney & Finch [23].

Within the depletion layer there is effective capacitance formed because each side of the junction has an opposite charge. This capacitance is called the junction or terminal capacitance and is a major determinant of the response speed of the photodiode detector [32] and also has a direct bearing on detector noise.

The capacitance per unit area is [53]:

$$C_A = \{(\kappa \varepsilon_0 e N_D)/2V\}^{3/2}$$

where: $e$ is the electronic charge; $V$ is the applied bias voltage

And for $p^+$-n detectors (as in our case)
\[ C_A = 19/(\rho_n V)^{\frac{1}{2}} \text{ (in pF mm}^{-2} \text{ with } \rho_n \text{ in } \Omega \text{m}) \]

These capacitance formulae determine the charge per unit area when there is a small increment in voltage. It is a practice to use the highest resistivity material available with the highest reasonable applied voltage to achieve a deep sensitive region and to keep capacitance to a minimum.

The n-type silicon used in the manufacture of these photodetectors has resistivity of approximately 50 Ωm. Thus when biased at 60 V this detector would be depleted to 300 μm and would have a capacitance of 0.35 pF.mm\(^{-2}\) and for a 3x3 mm\(^2\) area gives the pixel a capacitance of 3.15 pF.

### 3.2.3 Detector Properties And Spectroscopy

Fundamentally, the important properties of semiconductors are the bandgap and the net impurity doping [50].

Between the valence band (VB) where electrons are bound and the conduction band (CB) where they are free, exists an energy range where there are no allowed energy values for an electron. So for an electron to elevate from the VB to the CB, the energy value (or greater) of this bandgap \((E_g)\) has to be delivered to the electron when it is in this state. As electron energy is dependent upon the inter-atomic distances, sufficient thermal energy can be delivered to elevate the electron via vibrations of the semiconductor crystal (phonons). Also, incident radiation with adequate energy (>\(E_g\)), to break the valence bonds will put more electrons into the CB. The hole left behind in this process is considered a positively charge electron [50].

A pure (or intrinsic) semiconductor is one in which the number of electrons and holes are equal. Intrinsic semiconduction takes place when number of electrons in the CB equals the number of holes in the VB. In this case, thermally generated electrons and holes lead to the so-called intrinsic carrier concentration \(n_i\) where \(E_g >> kT\) is:

\[ n_i = N(E) \exp(-E_g/2kT) \]
where \( N(E) = (N_v N_c)^{\frac{1}{2}} \), the function for the density of the states and the number of vacant states, \( p_i \), in the VB is equal to \( n_i \) [51].

Applying an electric field can cause net momentum changes to the electrons in the CB and to those in the VB, thus generating two distinct contributions to the total current [51]. This means that the total current is a sum of separate contributions; part from electrons and part from the holes. The lower edge of the leakage current is governed by the number of intrinsic carriers [50].

**Impurities and lattice defects**

Unlike an ideal semiconductor crystal with its atoms at rest in a perfectly periodic structure, the real semiconductor strays from the ideal because of thermal vibrations, chemical impurities and lattice structure defects and these flaws dramatically affect charge flow throughout the device [52].

The lattice structure of the semiconductor crystal can be modified by the introduction of localised energy levels into the forbidden energy gap [53]. These elemental impurities of the 3\(^{rd}\) and 5\(^{th}\) group (group III impurities being acceptors and group V impurities donors) of the periodic table of isotopes can be intentionally substituted into the lattice sites in the forbidden gap and can cause the semiconductor to conduct more strongly [50]. The new localised centres created in this gap can also be ionised by donating an electron to the CB or accepting an electron from the VB and the energy required for these processes is less than the width of the energy gap \( E_g \) [54].

When an ion, such as boron, impinges a material, as what occurs in the ion-implantation technique used to dope this silicon, a great portion of its incident energy is imparted towards the end of its pathway. This large absorption of energy just before the ion comes to rest causes clusters of point defects to form. Often the total number of these induced defects well exceeds the number of implanted ions [55] and is a consideration when manufacturing the detector.
Recombination and trapping

When a incident radiation with energy \( \geq E_g \) interacts in the detector, electrons in the VB can absorb enough energy to be spurred up into the CB and in so doing, creating pairs of excess electrons and holes, resulting in a non-equilibrium concentration of charge carrier in the device. Thus according to the properties of the detector, natural processes then eventually allow equilibrium to be regained. The effective carrier lifetime is determined by the recombination rate of the excess electrons and holes created as the system moves towards equilibrium and the number of traps the carriers encounter.

As a charge is carried across the depletion layer it can be trapped at defects in the crystal lattice [49]. Defects in the semiconductor can trap electrons or holes and leads to recombination and annihilation of both carriers. Trapping also stops carriers from contributing to the conductivity [52]. Thus to achieve efficient charge collection, the sensitive volume must be relatively free of deep traps and have a high electric field to ensure carriers are quickly collected to minimise recombination.

Spectroscopy

The ionising radiation produces a number of e/h pairs in proportion to the stopping power of the material [49]. These charge carriers are freighted to their preferred electrode at a velocity given by the product of their mobility and the electric field [49].

\[
V(x) = \mu E(x)
\]

where: \( E \) is local E field; \( \mu \) is the mobility of charge carriers and \( \mu_e \) is \( \approx 3 \) times as large as the \( \mu_h \) [56].

The instant an e/h pair is formed, the charges begin to separate because of the electric field, thus a change in the charge is induced on the detector electrodes and this corresponds to the current that is observed on an external circuit [56]. The externally measured output signal is the accumulated charge on the feedback capacitor of a preamplifier that supplies a shaping amplifier.
and its shaping time is greater than the time needed for charge induction and accumulation at any electrode [57].

The sum of all contributions to the signal from all charge as they progress to the electrodes as a function of time, yields the total current pulse and the induced charged is obtained by integrating this current over the collection time [56]. In sum there are three components to the anode current: the 1st to the rapid birth of a point charge; the 2nd is caused by the motion of the point charge and the 3rd is the contribution due to the accumulation of trapped charge behind the moving point charge [57].

After the collection of all the free carriers at the contacts has been accomplished, the charge Q observed should be ideally proportional to the incident radiation energy imparted into the sensitive volume of the detector [49, 50]. The failure of any generated charge carrier not reaching the electrodes is detrimental to the detector’s resolution since trapped charges induce variations in the output pulse. Provided all the carriers are collected, the amplitude of the induced charge will be unconstrained by the depth where the charges were created.

When a negative charge is trapped in the detector, a positive charge is induced on the anode and this positive charge induces a negative charge across the preamp input terminals (and conversely for a trapped positive charge) [57]. Since the measurement is performed in the depleted region of the diode where the carriers are rapidly swept out by the existing $E$ field, recombination through localized levels is highly improbable. Thus trapping is therefore predominant in determining loss of signal pulse height through decreased collection efficiency [52].

### 3.2.4 Resolution And Noise

Spectra broadening is caused by: variations in the birth number of e/h pairs due to a specific incident energy being split between being deposited into the crystal lattice as phonons or into the ionization of the atoms which make the e/h pairs and because of noise in the detector and its amplifiers.
Noise is important because it sets a practical limit on the smallest signal that can be resolved. In electronic circuitry there are two main types of noise: thermal noise and shot noise. Thermal noise occurs even when there is no current flow through the detector due to active electrical components like resistors and shot noise is associated with the flow of any current, even if it does not contribute to the net flow.

Increasing the signal-to-noise ratio (SNR) can be achieved by reducing the noise reaching the output of the amplifier by narrowing its bandwidth. This means to bring the amplifier’s rise-time ($T_R$) and fall-time ($T_F$) together [58]. This process can be experimentally undertaken by plotting the shaping time against the detector’s capacitance to bring together the optimal SNR amplification setting by determining the “noise corner.”

The noise corner, where the noise is at a minimum, occurs at the peaking time $\tau_0$ ($\tau$ is when $T_R$ equals $T_F$). Experimentally (unless the diode’s specifications are present), the amplifier’s shaping time is adjusted until the noise of the system is lowest by observing the output signal on the oscilloscope. Because a PET detector needs to be capable of high counting rates, it must have low noise at shorter peaking times [59].

3.2.5 Initial Developments With Si PD And CsI(Tl)

In 1987, Sakai undertook a comparative study of gamma ray spectroscopy using eleven types of scintillators coupled to a Si PDs and then to a PMT [60]. Sakai found that the pulse height distributions of photodiodes can be compared to those of the PMT and the differences found in the two measurements are caused by the inherently different wavelength-dependent quantum efficiencies of the Si PD and the PMT.

His experiment nicely demonstrated that in the case of CsI(Tl) scintillator, it was possible to achieve 6.86% FWHM on the Si PD and 5.23% FWHM on the PMT with 898.1 keV gammas – a somewhat close result. The 540 nm peak wavelength of CsI(Tl) happened to have matched the strong spectral response (the relationship between the photoelectric
sensitivity and the incident wavelength) of the Si PD at this wavelength to create such good counting statistics and thus energy resolution. This result showed the great promise of Si PD and how they are capable of replacing PMTs for gamma spectroscopy in certain circumstances. [60]

### 3.2.6 A New Solid State Approach To PET Detector Module

Sadly CsI(Tl) is an unsuitable scintillator material for PET. To date, the new scintillator, LSO, evidently has the closest to ideal characteristics for its application to PET (except for its high cost). The obvious thought comes to mind: how effective would it be to build PET detector modules based upon the scintillator LSO attached to a Si PIN photodiode?

Unfortunately for LSO, its scintillation light peaks around 420 nm. And at this wavelength, commercially produced Si PDs have quite poor quantum efficiency because the charge carriers generated by the incident light are poorly collected and thus the signal is weak. Therefore it is necessary for specially designed Si PDs to be manufactured if solid-state devices are going to play a great role in revolutionising PET.

What's more, to have high quantum efficiency is not our only desire, as this would be reasonably easy to achieve. For PET, we need more than just high QE to bring out a higher spatial resolution, we also need a fast, noise-free signal. And obtaining this is where the greatest challenge lies.

To this point in time, only a few groups worldwide have made reported inroads into developing very high spatial resolution PET scanners based upon solid-state detectors. And even fewer have proposed using unity gain Si photodiodes: Moses et al of the Berkeley Group in the United States and this work carried out by our group here at the CMRP in collaboration with UOM.

The Moses et al approach [14] was to improve the performance of block detectors by attaching an array of Si PDs onto the front face of the scintillator block and a PMT onto its rear. The PD was to determine the crystal of interaction and the PMT would provide energy and timing signals.
While the concept of pixelisation of the array aids to improvement in timing resolution because the variations in the signal propagation to the electrodes is significantly less than for a detector with a larger area, the Berkeley detector concept was not feasible for high resolution PET because of statistical uncertainty in the DOI measurements within the scintillator crystal.

The CMRP approach discussed next is more elegant in terms of high resolution PET (see figure 3.3) design and does away with some disadvantages faced by previous designs.

### 3.3 New Design Of Si Pixel Detectors Modules For PET

To reveal a superbly precise and noise-free PET image, the PET scanner needs to combine high measurement accuracy of DOI and a high count-rate detection to minimize errors due to Poisson statistics [44]. Improving count-rate statistics, in order to achieve the optimal signal – simultaneously reducing noise – it becomes vital to efficiently match the scintillator emission wavelength to the photodiode’s absorption spectrum [44]. This matching is especially important because photodiodes have only unity gain.

![Figure 3.3 Illustration showing the layout of the CMRP PET detector module.](image-url)
Also, by entirely rejecting the use of PMTs in the detector modules, the bulkiness factor of past detectors is avoided. Thus a greater variety of options are now available to suit the characteristics required for the particular need.

### 3.3.1 Blue Enhancement Of Photodiode

Ideally, the most efficient charge collection takes place when incident photons interact and generate the e/h pairs in the depletion layer. As the attenuation length of LSO (420 nm photons) in silicon is relatively short (0.1 – 0.2 µm), it is desirable for the p⁺-layer to be finished well before this depth. Thus the design of the front surface of the device is critical.

Since photodiodes offer no internal amplification in the conversion of optical scintillation photons to an electrical signal, it becomes critically important to obtain the highest QE and collect every possible charge carrier while at the same time keeping noise (from Auger recombination) to the minimum. In other words, a strong SNR is fundamental to obtaining good statistics underlying good energy, timing and spatial resolution.

The intended goal of the doping profile of the p⁺-layer is to use a special boron implantation technique to heighten the electric fields of this region. The greater electric fields subsequently force an increase in the carrier drift velocities of charge carriers (holes) from where they are born to their collecting electrode. This increased conductivity leads to faster collection times, more efficient charge collection because recombination of carriers is less likely and correspondingly less noise. In addition, because the efficiency of charge collection is strong and similar numbers of electrons and holes are collected, the nearest to maximal signal (SNR) is formed and offers the promise of an improved SNR at shorter peaking times required in the case of PET.

As higher energy light photons have a shorter absorption length in silicon, most of them will be absorbed in the p⁺-layer that is usually around 0.5 – 0.7 µm in the photodiode. Absorption in the dead layer before entering the depletion region causes inefficient carrier collection due to
recombination thus generating noise (from electrons carriers generated in this region will likely recombine) and not contributing to the net signal collected.

By use of a special ion-implantation technique, strong electric fields become embedded very near the surface of the device. These regions of heightened electric fields ensure that charges generated are more likely to be swept towards their collection contacts. This technique allows the shaping of asymmetric nearly abrupt p-n junctions, very close to the surface [61], ensuring a fast and effective charge collection, leading to a fast rise time of the detector signal for improved timing resolution [62] and good energy resolution. The dead layer can also be minimized with this ion-implantation method.

The ion-implantation applied to this diode is designed to provide a very high electric field at a depth of less than $0.1 – 0.2 \mu m$, which is the absorption length of LSO’s 420 nm emission wavelength, to approach very close to the active surface of the pixel detector [62].

Another advantage of the pixilated design of this detector is that it naturally enhances the signal collected by making best use of the Shockley-Ramo effect. This is where there is a “proportionally higher contribution to induced charge on the pixilated electrode because the induced charge is proportional to the difference in weighting potential between the point of origin and collection” [63]. As the sensitive region is very near the front electrode, this means that any motion of the charge carriers taking place near the pixel surface will cause a proportionally stronger contribution to induced current on that electrode, i.e., the rise time is much shorter.

This means that each generated charge, when moving at a velocity $\dot{v}$, contributes a current $e.\dot{v}/d$ in the detector, where $d$ is the distance between the electrodes and this induces an identical signal in the external circuit [63, 64]. This generation of charges at a very shallow depth (under a strong electric field) adds to the impulse force put on the carrier (hole in this region) that boost the velocity of the charge and therefore its effect on the
induced charge over the preamp. This is another advantage of using pixelation in the proposed detector.

The reason to use small pixels is because the smaller area of each element means that capacitance is subsequently less. As by the general relation of capacitance that varies non-linearly with applied voltage \( dV = \frac{dq}{C} \), the smaller the capacitance, the greater the potential voltage.

This concept can be extended further by the incorporation of \( p^+ \) strips into the front face of the photodiode. By implanting an even higher concentration of Boron atoms in strip formation on the front, the electric field in that strip region is heightened. The concentration gradient formed by the \( p^+ \) strips over the doped \( p^- \) region theoretically hastens the movement of hole charges more quickly towards the contact, thereby enhancing the induced charge over the external circuit and improving timing properties. However, this is outside of our work and is a subject of another project.

To keep the noise to a minimum and for fastest charge collection, this detector is operated at just fully depleted to ensure lowest possible capacitance because of its relationship on noise contribution from the preamp and the corresponding decline of preamp rise time [56]. This full depletion enables better timing because of low capacitance of pixel.

Besides creating a more uniform electric field, there is no real benefit to over-depleting the detector in an effort to enhance speed of carrier collection. When over-biasing occurs, the signal waveform begins to deteriorate along with a loss in signal resolution. Therefore the fastest timing measurement is made when the detector is just fully biased [56].

### 3.3.2 Other Design Considerations

A passivating layer of thick SiO2 protects the outer surface of the photodiode against external influences such as dust and moisture that may especially affect the dark current. Ions coming to lie on the surface may cause formation of charge inversion layers beneath the surface and generate changes that cause deterioration in the reverse current and add to the noise
The effect of the deposition of a few ions onto the surface will produce a strong effect because the impurity levels in this device are very low and thus it takes few charges to create relatively large field distortions or surface channels [50]. As the critical area of our device is very near the surface, these surface channels can be ruinous because they lead to incomplete charge collection near the surface. The placement of a guard ring on each pixel element attempts to keep these surface currents in check.

The design of the photodiode arrays improves the QE at 420 nm with the assistance of special antireflective coating based on a SiO$_2$ layer with thickness $\lambda/4n$ (where $n$ is the refractive index of SiO$_2$ and $\lambda$ is the incident photon wavelength). In the current model of the PD silicon array, the thickness of antireflective coating was 155 nm.

This detector module employs a parallel readout of each pixel by means of a “fan-out” wiring system that connects each pixel photodiode to a Viking read-out chip via a connecting pad with a 90 $\mu$m pitch. The Viking chip is a low-noise (RMS noise is 200 e$^-$ for 5 pF capacitance, as shown in Figure 3.3.2), high-speed 128 channel chip used for strip detectors high energy physics (HEP) applications [66].
Figure 3.3.2. Noise scheme of the Viking read-out chip (Taken from O. Toker et al NIM A 1994 [67]).

Figure 3.3.3. The Viking chip reads out the signals from up to 128 Silicon detectors, with very low noise contribution (Taken from O. Toker et al, NIM A 1994 [67]).

Combining the Viking chip (with block diagram as shown in Figure 3.3.3) to the photodiode array designed in the CMRP enables the construction of detection module that can overcome the noise associated with the PD and the associated electronics with its potentially very high SNR that would ideally suit its application in PET.

3.3.3 Speculation Of New PET Detector Module’s Parameters

Estimation Of Capacitance

In detector, with resistivity of approximately 50 $\Omega$ m and when fully depleted at 60 $V$, the sensitive depth should be near 300 $\mu$m.

using 
\[ C_D = \frac{19}{(\rho_n V)^{1/2}} \] 
(in pF mm$^{-2}$ with $\rho_n$ in $\Omega$ m)

= 0.346 pF/mm$^2$

The detector pixel capacitor with area 2.8 x 2.8 mm$^2$, gives a detector capacitance of 2.72 pF (approximately).
The leads from the detector to the preamp also contribute an estimated 2pF, bringing the total of the detector and the leads to ~5 pF.

**Estimated Noise Of Read-Out hip (Viking)**

With an estimated 5 pF capacitance of load on this device, it is seen that the corresponding electronic noise charge is ~200 e⁻ (rms). However this noise value assumes the detector is fully depleted to 300 µm.

**Estimation Of Detector Response Using LSO And CsI(Tl) Scintillators.**

Assume that LSO light yield is 30,000 ph/MeV peaking at wavelength 420 nm [37] and CsI(Tl) outputs 60,000 ph/MeV at 540 nm wavelength [37], using less than the maximum absolute light output yield.

The incident radiations on PET detectors are annihilation photons of energy 511 keV. For LSO, the scintillator yield should be ~15,300ph/511keV and CsI(Tl) the incident energy should yield ~30,000ph/511keV.

The QE of the CMRP photodiode version 1 (SPAD 1) that was used here was approximately 68% at 420 nm and 50% and 540 nm.

Assuming 100% charge carrier collection efficiency and scintillator photon emission efficiency, ie no self absorbance:

The LSO scintillator per annihilation photon should create $15,300 \times 68\% = 10,404$ e/h pairs in the photodiode and CsI(Tl) should create $30,000 \times 50\% = 15,000$ e/h pairs.

(This corresponds to $3.6 \times 15 \times 10^3$ (eV x e/h pairs)

deriving a 54 keV signal).

Since the noise of the system is expected to be 200 e⁻ (rms) at full depletion, the SNR could be as high as 34:1 for LSO and up to 75:1 for CsI(Tl). Even if the light collection efficiency for these scintillators is only 50%, a strong SNR should be expected from this design. This estimation was
taken for the situation where all light photons are approaching the sensitive volume of photodiode.
4.0 Detector Fabrication And Characterisation

Various experiments are carried out on this detector array with the aim of determining some of its characteristics so further improvements can be made. Firstly, the device is described in detail along with the environment where the experiments were carried out. The following experiments were completed on the detecting device: current-voltage function of the detector; X-ray spectroscopy using $^{241}$Am 59.5 keV gammas; Gamma ray spectroscopy by exciting a CsI(Tl) scintillator with $^{137}$Cs 662 keV gammas; then the effect of cross-talk between pixels, the effect of neighbouring pixels bias on pixel response, and finally, the determination of the jitter/walk effects on the timing resolution was undertaken.

4.1 Array Description

The device studied in this project was created from a high resistivity (3-5kΩ.cm) n-type silicon was doped with boron using the ion-implantation method. This implantation created a p⁺ region very near the front surface of the photodiode. Onto this front face, aluminium contacts on a perimeter of each pixel were evaporated to effectively enable charge collection produced in the device to be put through to electronic read-out instruments.

The front side of the device (as shown below in figure 4.1) is a pixelated array of 64 detector elements with dimensions 2.8 x 2.8 mm², with the array dimensions 25 x 25 mm². The detector is approximately 300 μm thick and requires a reverse bias of approximately 60 V to deplete it to this depth.

A continuous metallic aluminium evaporated film covers the rear (n⁺) of the array and acts as an ohmic contact to collect negative charges (when operated in reverse bias mode). This rear electrode sits upon a Kapton testing board that protects and gives a working rigidity to the device.
To characterise this device experimentally, the pads of the Si array were wire bonded to each electrode to an interconnection device called a fan-out and each pad was individually hooked to the preamp. This fan-out (shown in figure 4.1) architecture is thus able to allocate for various neighbouring subsets of the pixels to be electrically connected together and this in turn allows the formation of larger composite pixels ranging in size from 3 mm$^2$ to 25 mm$^2$. Therefore, this fan-out enables the neighbouring pixel to be read-out while the primary pixel is irradiated as well as having the ability to form a bundle of pixels into a single detector unit with the particular area required.

**Experimental Conditions**

The experiments involved in this project were undertaken inside a specially designed radio frequency (RF) interference-free room that keeps a constant $23^0 \pm 1^0$C temperature. This room is housed within the CMRP laboratory, University of Wollongong.

Many of the experiments make use of a special X-Y table with attached pixel array that has the capability to finely and accurately reference distance.
measurements to within micron dimensions of the interaction point’s location. This X-Y table is situated inside a “light-tight” aluminium box within this RF free room because when the detector is not in absolute dark, the photosensitivity of the device causes a photoelectric current to take place. This environment ensures that experiments are undertaken with the knowledge that environmental factors are reasonable constant, especially with regard to undesired ambient light contamination.

**General comments:**

Wires with 2 mm diameter were soldered to the contact pads of the fan-out to outside circuitry (pre-amplifier, etc) via BNC plugs on the light tight box’s wall. Soldering these plastic insulated aluminium wires to the detector’s contact pads required Baker’s Flux® to ensure a proper electrical junction. The particular soldering process involved is a very delicate operation that gives quite variable results in the quality of the joint.
Unexpected Experimental Development With Photodiode (Version 1)

Observations indicated that the dark current of the device was deteriorating for bias \( \geq 15 \, V \). This early avalanche breakdown of the detector, believed to be due the deposition of ions on the sensitive area, meant that the pads of the array could not be biased to the full depletion depth. We found a method to stabilize and to stop further breakdown in the detector. This following remedy was applied in an attempt to block this deterioration: the open face of the photodiode array was coated with a layer of *Norland Optical Adhesive #65*. The theory behind the application of this special high transparency optical adhesive was to hopefully impede the further breakdown of the array. This coating process also introduced new characteristics into the device. In addition, the thick application of epoxy to the surface lamentably prevented further alpha spectroscopy of the PD from being carried out. However, regardless of this drawback, the application of this protective layer was still deemed necessary.

4.2 Investigation of Current–Voltage I(V) Characteristics Of A Single Pixel In Array

Electronically, this detector type is a reverse biased semiconductor diode that exhibits a leakage current whose fluctuations produce noise [68]. This leakage current quickly avalanches once the device reaches its breakdown voltage. This point of breakdown places an upper limit on what electric field strength the diode can withstand and to what depth it can be depleted and is primarily due to its given resistivity [56].

Ideally, the conductivity of the fully depleted detector should be zero but in reality, the reverse currents in quality p-i-n pixels is determined by recombination in the depletion region and is proportional to \( \sim V/\tau \). (Where \( \tau \) is the recombination lifetime for minor charge carriers and \( V \) is the volume of the depletion region.) For detector characterisation, I(V) curves are fundamentally important because leakage currents are due to impurities and
defects in the crystal [69]. The magnitudes of the surface currents may substantially contribute to the leakage current. This has drastic effects on the energy resolution of the device. The relationship between bias voltage and the corresponding leakage current allows the determination of a stable range of voltages where any variance in the voltage does not cause great fluctuations in the leakage current [69]. This relationship also aids in determining the exact current flowing so the voltage drop over the load resistors can be subtracted from the supply voltage and allow an accurate calculation of the true bias on the detector [56].

These pixel detectors also incorporate a guard ring in their design to minimise the effects of surface currents and as well, the whole array had a guard ring with p⁺-strips around the array’s 25 x 25 mm² perimeter.

**Experimental Method**

The current-voltage curve measurements were taken on the arrangement shown in figure 4.2.1 where we investigated the role of the guard ring and the biasing conditions. The current was read as a function of reverse bias voltage under three conditions. Firstly, without the guard ring of the detector element biased. Secondly, with the guard ring biased and finally with the element’s guard ring and the eight immediately neighbouring pads surrounding the studied detector biased at the same value.

![Figure 4.2.1 I-V Curve experimental set-up](image)

**Results**
Discussion

Figure 4.2.2 clearly reveals the significant effect of biasing the guard ring (b) when above $10 \ V$ reverse bias. After the immediately neighbouring pads were biased along with the guard ring, there was a further reduction in the dark current of ~20%.

The specific behaviour of the I-V characteristics, as indicated by (a), can be observed by the abruptly increasing leakage current soon after 10 Volts bias. This effect can be explained by the increasing surface current due to inversion of surface under MOS capacitor that is created by Al-SiO$_2$-Si junction on the perimeter of the sensitive pixel. On the other hand, as indicated by (b) and (c), if negative bias is applied to the guard ring, this effect fails to occur because this surface current is blocked.

While the reaction of optical grease with the photodiode was nearly fatal with respect to the dark current, a layer of optical adhesive over the photodiode noticeably stabilised the dark current. Over time there was still some degradation in dark current so an attempt was made to restore the Si PD to its original specifications. To do this the device was annealed at moderate temperatures but this only caused minor improvements.
In operational condition all pixels will be biased and we can rely upon our current voltage character (see figure 4.2.2 (C)), ie., the reverse current of approximately 0.4 nA corresponds to a (nearly) fully depletion bias of \( 50 \, V \).

### 4.3 Variations in Uniformity of Response Across Array

To gauge the uniformity of charge collection across the entire array a scintillator is optically coupled to each pixel element and then the spectrum of each pixel is compared to its neighbours.

The scintillator used was a plain cut 3x3x3 mm\(^3\) CsI(Tl) crystal. The light output of CsI(Tl) is 50,000 photons per 1 MeV and its sides are unpolished. As CsI(Tl) is not a fast scintillator (\( \tau \sim 0.9 \, \mu s \)), the amplitude of signal from this crystal will be strongly dependent on which shaping time constants are chosen in the successive stages of amplification. With CsI(Tl) crystals it is clear that in a low noise amplification system, improvements in performance can be improved by employing longer shaping times [43] and this was shown to be the case with the fairly lengthy shaping time of 3 \( \mu s \).

The spectroscopic energy response uniformity of the photodiode is determined by principally studying: how the response varies over the entire array of pixels; how the response varies across an individual pixel and by observing the effects on a particular pixel while the neighbouring pixels are irradiated, ie., cross-talk. This energy spectroscopy with scintillator establishes the strength of the pulse signal output.

#### Experimental method

Gamma ray spectroscopy experiments were undertaken using the readout set up shown below in figure 4.3.1 with the scintillator crystal placed on top of the detector element studied. All except one side of a 3 mm\(^3\) CsI(Tl) scintillation crystal was tightly wrapped in 10 layers of Teflon tape and optically coupled to the PD element face using Bicron\(^{TM}\) optical grease. As discussed earlier, this optical grease was unfortunately found to react with the Si photodiode surface in the first version of antireflective coating.
technology and caused a near fatal increase in the dark current. However, when a layer of *Norland 65* optical adhesive was applied to the photodiode, this was found to initially raise the dark current but then stabilised. After the application of the optical adhesive it was found the optical grease had no further effect on the deterioration of the dark current. (Further versions of PD array technology took into account this problem and implemented a special multi-layer anti-reflective coating to protect the surface.) This improvement was a result of our investigation in a project. The crystal was excited by a $^{137}$Cs source and the scintillation pulse was read out by the Si pad array element connected via an *Ortec*™ 142A preamp to a *CANBERRA™* 2021 spectroscopy amplifier (with 3µs shaping time), with the data collected with an *AMPEK™* 8000A multichannel analyser in ADC units. Each of the 64 pixel elements was measured and spectra for Cs$^{137}$ were recorded.

![Experimental set up for spectroscopy studies in Si photodiode array.](image)

Additionally to spectroscopy measurements, we investigated the array response to the laser beam for $\lambda = 700$ nm. Although the absorption depth for this wavelength was deeper in comparison with LSO’s 420 nm, this data is a good test of response. The laser measurements shown in this section and the next were taken in collaboration with the University of Melbourne and are shown here for completeness. The optical uniformity of the array was
measured by mounting a pulsed laser diode (\( \lambda = 700 \text{ nm} \)) in an X-Y table jig. The diode is fitted with a special collimator lens and has a spot size of nominally 50 \( \mu \text{m} \). The computer controlled X-Y table was designed and built by the CMRP and has a step size of 5.2 \( \mu \text{m} \) and 10.4 \( \mu \text{m} \) in the X and Y direction respectively. The accurate positioning of the X-Y table allowed for the uniformity of the pad array as well as individual elements with very high precision. The pad array was positioned parallel to the XY-plane of the X-Y table at a distance of \( \sim 1 \text{ mm} \) from the head of the laser diode lens. The readout of the pad array via the VIKING chip [79] was triggered from the pulsed laser. Each optical uniformity data run incorporated 1000 events with all data stored in the data acquisition computer and then averaged. We are not presenting here the full experimental set-up of read-out system since these measurements are shown only to demonstrate the comparison with \( \gamma \)-spectroscopy.

**Results**

It should be emphasised that all experiments carried out in this project were done on the first version of the technology, ie., where only the SiO\(_2\) layer was above Si sensitive area.

Before discussing the array experiment, it is worthwhile to bring to the attention of the reader some preliminary results involving an individual pixel detector so the development of the array can be followed.

While manufacturing the Si PD arrays at the microelectronics factory BIT Ltd, a run of 3x3 mm\(^2\) (named SPAD 1) photodiode pixels were manufactured on the same wafer utilizing the same manufacturing process. These diodes were used for preliminary aspects of characterization of the technological quality and modelling. The 3x3 mm\(^2\) pixel was mounted on a small Kapton (PC) testing board that provides easy access to take electrical and optical measurements.
The single 3x3 mm² Si photodiode was coupled to 3x3x3 mm³ CsI(Tl) scintillator (shown in figure 4.3.2) and to LSO 4x4x10 mm³ (figure 4.3.3) and irradiated by $^{137}$Cs.

**Figure 4.3.2** Room temperature, pulse height spectrum of a 3x3x3 mm³ CsI(Tl) crystal excited by I$^{22}$Na and II $^{137}$Cs read out by the SPAD1 PD. Spectrum III is the 32 and 36 keV X-ray spectrum of $^{137}$Cs directly detected by the PD.

**Figure 4.3.3** Room temperature, pulse height spectrum of a LSO crystal excited by I $^{22}$Na and II $^{137}$Cs. The LSO crystal was read out by the SPAD1 PD. Spectrum III is the 32 and 36 keV X-ray spectrum of $^{137}$Cs directly detected by the PD and is scaled by a factor of 10 beyond channel number 275 for clarity.

Figure 4.3.2 demonstrates spectra obtained with individual Si photodiode when coupled to 3x3x3 mm³ CsI(Tl) scintillator. The energy resolutions of 7.7% FWHM for the 662 keV gamma rays from $^{137}$Cs and 9.5% FWHM for the 511 keV gammas from $^{22}$Na have been achieved. Figure 4.3.3 shows the energy resolutions when the Si photodiode was coupled to LSO and
irradiated. It must be noted that in this case LSO scintillator had the size 4x4x10 mm$^3$; this is larger than the 3x3 mm$^2$ sensitive area of Si photodiode. Such mismatching leads to losing source photons from scintillator to Si PD and effected energy resolution. The energy resolution in this situation achieved 28.8% FWHM for the $^{22}$Na 511 keV gamma rays and 22.7% FWHM for the $^{137}$Cs 662 keV gammas.

The data collected from this single pixel establish that the applied coupling technology and the manufacturing technology of PD are working well and that it is likely that application of this technology to an array of pixel detectors should also be expected to work.

Table 4.3.1 shows obtained spectroscopy results for each pixel in the array. Described above is the experimental spectroscopy approach applied to each pixel on the array to study uniformity of response. This response was affected by uniformity of optical coupling, charge collection and can be affected by neighbouring pixels and guard rings.

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Table 4.3.1 shows obtained spectroscopy results for each pixel in the array. Described above is the experimental spectroscopy approach applied to each pixel on the array to study uniformity of response. This response was affected by uniformity of optical coupling, charge collection and can be affected by neighbouring pixels and guard rings.

Spectroscopy Response Variations Over The Array
Table 4.3.1 Charge collection efficiency (given from the photopeak channel) varies over the 64 silicon pad detector elements by ±28%. Data is acquired for $^{137}$Cs.

The variation in charge collection efficiency of ±28% over the 64 pixel elements can be explained by variations in the uniformity of the optical coupling. As the optical coupling in this experiment was done by hand, the application of the optical grease is naturally uneven. Also, it is easy to lose photons at this stage of the detector and the result is quite large variations in final energy resolution. This issue has been addressed and very recently developed optical coupling technology has lead to 13% FWHM energy resolution for 511 keV gammas incident on LSO but this is not included in this thesis.

Figures 4.3.4 and 4.3.5 show distribution of photopeak amplitudes across the pixel array and a typical spectrum from a pixel of the photodiode array respectively.

![Uniformity of Array (Spectrographic)](image)

Figure 4.3.4 Bar graph showing relative distribution of photoelectric peak positions (channel numbers) of the 8 x 8 Si photodiode array.
Figure 4.3.5 Example spectrum (array pad element 27) showing the $^{137}\text{Cs}$ 662 keV peak centred on channel number 215

**Discussion**

Results from the uniformity experiments performed over the whole 8x8 silicon pad array, using the two different methods outlined above, are presented in Figure 4.3.6. In this figure, each column represents the output from an individual pad element in ADC channel units. Figure 4.3.6(a) shows the response of the array to a laser diode. The noise in the mean output of each pad element was 3% FWHM of the mean value. A clear trend is observed in laser diode response data across the array. The variation is ±12% from the mean optical response has been observed, such variation is quite acceptable, as it is standard practice to calibrate each detector in a detector module if necessary.
Figure 4.3.6. Comparative response of the array using optical (a) and gamma ray spectroscopy (b). Note: the ADC channel is channel number of MCA.

When the detector module is used under the circumstances discussed above, the final spectrographic response uniformity (shown in figure 4.3.6(b)) between the array pads was found to vary by ± 28%. This has influence on energy resolution. The factors causing variations are: the effect of light build-up in the scintillator, the thickness and uniformity of the optical grease between the scintillator and the array pixel photodiode. In both cases Figures 4.3.6 (a) and (b), the observed gradient of response of the pixel array is clearly observed. The reduction of the signal by ~ 10% corresponds to the pixels with longer connection pads that produced additional capacitance to the PD pixel. Capacitance of fully depleted pixel is about 3 pF, whereas contact pads can have up to 5 – 6 pF for pixels opposite the readout side of the array. Additional capacitance is reducing signal due to capacitance charge division between pixel connection pad and input capacitance of readout circuit.
4.4.1 Uniformity Of Charge Collection Across Pixel

Experimental Set-Up And Method

The uniformity of charge collection across an individual pixel element was determined by reading out the spectrum directly from a collimated $^{241}\text{Am}$ alpha source. Note that only one pixel element was chosen at random for this experiment. The array was secured on the X-Y table and a collimated (0.5 mm diameter) $^{241}\text{Am}$ alpha source was placed as near as possible to the face of the pad element.

Then the source was incrementally scanned across pixel with step of 0.5 mm using a computer control unit. The same spectroscopic electronics set-up used in experiment 4.3.1 was continued on for this procedure.

This procedure was then repeated using a mounted laser beam ($\lambda = 700$ nm) instead of the alpha source.

Results And Discussion Of Charge Collection Uniformity

Across An Individual Pixel

As each pixel element has a sensitive detecting area of 3 x 3 mm$^2$, this spectrographic analysis, taken of $^{241}\text{Am}$ under free air geometry with 0.5 mm collimation (shown is figure 4.4(a)) indicates that charge collection is quite uniform to near the pixel’s edge. Figure 4.4(a) shows counts taken of the area under the peak of $E_{\alpha} = 5.3$ MeV for the different positions of the collimated $\alpha$– beam.
Figure 4.4(a) Alpha peak counts read out as a collimated (0.5 mm) $^{241}$Am alpha source is incrementally moved across an individual array element.

Figure 4.4(b) The uniformity of an individual pixel element when optically excited with a scanning laser beam from the laser LED.

The acquisition time for each position was the same. Taking into account the geometry of the $\alpha$ – collimator, the spot size diameter was ~ 1 mm. This is why degradation of the number of counts is observed for the $\alpha$ – beam position ~ 0.5 mm from the edge of the sensitive area of the pixel. The much higher spatial resolution for the optically stimulated pixel PD (figure 4.4(b)) indicates that the uniformity is very strong until (within microns) the very edge of the detector pad. The variation in the response across an individual pad element is less than 1% across the entire pixel.
The laser beam data was much more uniform within the sensitive area of the pixel. This is due to the much smaller light spot of ~ 50 µm in comparison to the collimated alpha beam and measurements in air. The depth of interaction by α – particles was determined to be ~ 17 µm and the 700 nm photons to 10 – 20 µm so data is related to the testing of the same sensitive area.

The obtained data is important for performance of the PET detector because the photons from the scintillator are approaching the sensitive area of pixel in different positions and under different angles. The uniformity of the magnitude of collected charge demonstrates uniformity of the entrance window, ie., it shows the absence of dead layers. And as well, it shows the advantages of the developed technology that gave a uniform and efficient electrical field across the sensitive of the pixel at shallow depths.

4.4.2 Determination Of Cross-talk Between Pixels

Accuracy of PET imaging depends on the ability of measuring the DOI for each event. In the proposed design of PET module the DOI is determined by scintillator voxel 3x3x2 mm³ sitting above each pixel in which the interaction occurs. Assuming that each voxel is optically isolated from others, the accuracy of DOI measurements depends on the ability of response from each single pixel that belongs to a particular voxel.

If each pixel is considered to be a separate detection element in itself, it is thus important to realise to what extent the neighbours are affected if a radiation event occurs in a single voxel above pixel.

Two different approaches are possible for determinations of the two dimensional position of interaction of 511 keV photons:

1) measurement of centre of mass of photon flux from scintillator using signal from four neighbouring pixels in the case of continuous scintillator above the array or

2) using a voxelated (100% optically isolated) scintillator to measure response in one pixel only.
In both situations it is important to avoid cross talk between neighbouring pixels. As was discussed in section 2.2.4, PS – PM tubes have a problem with cross talk that deteriorates the quality of image. The effect of cross talk in pixelated Si detectors has not been thoroughly investigated for nuclear medicine applications.

The nature of cross talk in pixelated detectors is related to inter-capacitance between neighbouring pixels and capacitance of the depleted pixel. The charge induced in a pixel due to photons from the scintillator will be shared with other pixels through charge-capacitance division on $C_{\text{depleted}}$ and inter-pixel capacitance.

In this project we investigated cross-talk for the CMRP designed Si photodiode array.

**Experimental Set-Up and Measurements**

By utilising an experimental set-up similar to that in which each pixel of the Si PD array is characterised, a collimated $^{137}$Cs source is directed upon the 3 mm$^3$ CsI(Tl) scintillator optically coupled to the pixel and the spectrum from its neighbouring pixels was read out with a traditional spectroscopy set up as shown in figure 4.4.3 below. The experiment was done with biased and not-biased neighbouring pixels.
Figure 4.4.2 Experimental set-up for study of cross-talk between pixels.

No charge signal was read out from neighbouring pixels (above the noise level) during the irradiation of the particular pixel of array with 662 keV photons from $^{137}$Cs source.

This experiment demonstrated that the design of Si PD pixel array is free from cross-talk in such experimental conditions for different biasing of neighbouring and irradiated pixel.

4.5 Characterisation of Noise Characteristics of Pixel Array

In our previous spectroscopy measurements, with a single 3x3 mm$^2$ PD taken from the same Si wafer as the array, we compared response of optically coupled single PD with CsI(Tl) and LSO scintillators in terms of amplitude of charge produced and the comparison of this charge to direct deposited energy in bare Si PD. It was demonstrated that response from 662 keV photons (CsI(Tl)) corresponds to the 45 keV direct energy deposition in Si PD and for the case of 511 keV upon LSO corresponds to 8 keV of direct energy deposition.

The small amount of deposited energy in the Silicon PD pixel from scintillator photons generated by 511 keV gamma events require outstanding signal-noise (SNR) performance of the Si PD array under room temperature.

For measurement of SNR we investigated the spectroscopic response of a particular pixel of Si PD array using X-ray source $^{241}$Am. Comparison with single 3x3 mm$^2$ PD also has been done.

For this experiment we used direct X-ray radiation that produces similar number of e-h pairs as in case of coupled LSO-photodiode irradiated with 511 keV gamma photons. The $^{241}$Am X-ray source was chosen for this investigation.

Experimental Set-Up
The X-ray energy resolution was measured using an $^{241}$Am source that was placed directly onto detector element 36 of the array. The photodiode at the time still had a coating of optical grease over its front surface. Because of this, alpha particles from the $^{241}$Am are stopped but its X-ray emissions would still penetrate into the detector. The experimental electronics arrangement used is like the one shown in figure 4.3.1.

![Figure 4.5 Pulse height spectrum of photodiode detector element 36 directly excited by $^{241}$Am X-rays. Peak is of 59.5 keV gamma emission.](image)

Figure 4.5 shows typical spectra obtained from a particular (number 36) pixel of Si PD array.

While it is demonstrated that some detail has been lost to noise, for example, the $^{241}$Am source has a few low energy X-rays ($E < 30$ keV) that are totally lost. However, the 59.9 keV gamma ray is still clearly depicted and has a FWHM resolution of 8.2%. The primary reason for the high noise is due to the fact that the device was not fully depleted with the reverse bias for this experiment of $25\ V$ (explained in earlier section – due to early breakdown). This means there is a much greater capacitance across the junction and thus there is a greater series noise that primarily is caused from...
sources in the input stage of the pre-amp. The measurements were taken at bias $V = 25V$, i.e., not under full depletion condition that minimises capacitance. Under full depletion energy resolution will be even better.

This experiment demonstrated that for even a large detector array (25 x 25 mm$^2$) at room temperature, the developed technology derives a strong signal well above noise, which is expected to be $\sim 50$ keV from deposition of energy into the LSO voxel.

4.6 Timing Characteristics Of Pixel Array

It is important to understand the distribution of triggering signals using current pulses from the back-side of the pixelated detector because for any incident light on a particular pixel, there will always be an observed pulse from the back of the array. The investigation of jitter in such situation provides an orientation for possible high resolution in further PET detector modules utilizing a LSO – Si PD array. Additionally, the modelling of light absorption conditions as in the case of 420 nm photons near the surface of the p$^+$ region will be advantageous.

For modelling this situation we have used $\alpha$ – particles from a $^{201}$Po source. Irradiation was incident on the p$^+$ side as this will be the case in real situation with scintillator – PD coupling in this PET module. The range of $\alpha$ – particles in Si was estimated $\sim 17 \mu m$, taking into account absorption in air.

For modelling the effect of 420 nm light absorption near the surface of the PD, we irradiated Si PD with $\alpha$ – particles incident through mylar film. In this case where the ionisation is brought closer to the surface, the effect of energy deposition in this near-surface region can be studied and provides a model for our time resolution.

It is desirable that the timing signal always be triggered at the same fraction of the output pulse. This is important because the pulse amplitude has a distribution due to energy straggling of the $\alpha$ – particles and the finite energy resolution of the detector. To avoid this situation (see figure 4.6.1)
we fed both signal channels through constant fraction discriminators (CFD 935). A delay was added into the back end pulse after the CFD to cover sufficient time for walk effect. Therefore the front side p+ signal pulse registers start on TAC and the rear n+ signal registers the stop signal.

**Electronics Settings:**

The timing Filter Amplifiers (TFA) were both set with a gain of ~20, and an integration and differentiation time of 10 ns. The output pulse height was adjusted to generate a ~200 mV signal at the output of the TFA with these same conditions.

Because minimum jitter occurs when the trigger threshold is set at the point of maximum slope dV/dt of the pulse transition [56], the delay time for each CFD was thus matched to the system and set at 10 ns, and the width of the CFD output pulse minimised.
The input of the CFD threshold (activator) was 100 mV. Biasing of the detector was 26 V (13 V on bottom and 13 V on the top). The output of the (Ve biased) substrate was delayed by ~25 ns after the CFD and the output of the Time-to-Amplitude Converter (TAC) was fed into the MCA. The TAC range is 50 ns, where 10 V ≡ 50 ns delay).

Timing Resolution Results

Figure 4.6.2 TAC of 1 mm collimated $^{210}$Po 5.3 MeV alphas onto photodiode face (in free air). Frontside-start, backside-stop.
Figure 4.6.3 TAC of 1 mm collimated $^{201}$Po 5.3 MeV alphas onto photodiode face (in free air) traversing 17 microns of mylar. Frontside-start, backside-stop.

**Discussion**

Figure 4.6.2 shows timing jitter when 5.3 MeV $\alpha$ – particles are incident directly onto the front $p^+$ surface of a particular pixel. And figure 4.6.3 shows timing jitter when a 17 $\mu$m mylar film is placed between the PD and the 5.3 MeV $\alpha$ – particle from $^{201}$Po source. The study of these variations in detector rise time illustrates the effect of depositing ionising energy nearer the photodiode surface. The unperturbed, ie., non-mylar case (figure 4.6.2) has a FWHM of about 200 ps while the mylar blocked photodiode (figure 4.6.3) set-up the FWHM is around 700 ps.

The cause of the higher resolution of figure 4.6.2 may be due to a higher electric field at the depths of the sensitive volume of the PD where the 5.3 MeV $\alpha$ – particles deposit a substantial portion of their energy. However, the worse resolution of the mylar-covered PD could be due to the weaker electric field near the surface of the PD and possibly more importantly, to the fact that the counting statistics are around ten times less than for the directly irradiated PD.
Conclusion

A new custom made silicon PiN photodiode has been characterised with the aim of forming a PET detector module to improve the spatial resolution in PET allowing Depth Of Interaction (DOI) measurements.

This module consists of a silicon PD array optically coupled to LSO scintillator emitting photons with main emission wavelength of 420 nm. These photons are absorbed effectively near the surface of a Si PD and this creates the challenge in their detection. Also, the finite number of photons reaching the surface of the PD under different angles is an additional challenge for obtaining an appropriate S/N ratio for the PET detector module.

The major innovation of this development is a Si PD array with efficient near-surface charge collection capability thus maximizing the response for the LSO scintillation photon spectra. These low-leakage-current PDs capable of good energy and timing resolution may enable the replacement of traditional PMTs in PET.

The work outlined in this thesis indicates that these photodiodes could be used to develop a new PET detector module with DOI capability. Obtaining accurate DOI information is one of the main criteria that must be met if the spatial resolution of PET is to be improved.

During this project we investigated electrical, spectroscopic and photo-characteristics of the first version of a photodiode array with the aim of developing recommendations for further improvement in them for application to PET detector modules. Investigations were carried out on the newly designed 3x3 mm$^2$ Si detector PD in 8x8 array format with SiO$_2$ antireflective coating (AR) corresponding to the minimum reflection of photons with wavelength 420 nm. This PD was designed at CMRP and was produced at SPA BIT Ltd. Initial investigations were carried out on a single 3x3 mm$^2$ PD produced in the same Si wafer as an array in the same technological batch.
Study of I-V characteristics of single photodiodes and pixels on the array demonstrated full depletion of PD under 50 V, reverse current corresponding to full depletion was ~ 450 pA per pixel and was ~ 250 pA for single diode. We investigated the effect of the guard ring on a single PD using an I-V curve. Without the guard ring positively biased we observed a step-like increasing of reverse current under reverse bias of ~ 10 V that was associated with a MOS capacitor created by passivated silicon oxide and Al metallisation on a perimeter of the sensitive p-region. Analysis demonstrated that the increasing of leakage current was due to the surface channel created due to inversion of Si under the MOS capacitor. With biased guard ring applied the reverse current follows the $V^{2/3}$ law up to breakdown and this is confirmation that reverse current is a bulk current only.

The quality of the first version of the Si PD technology and optical coupling was tested on a single 3x3 mm$^2$ PD coupled to 3x3x3 mm$^3$ CsI(Tl) and then to a 4x4x10 mm$^3$ LSO scintillators, with both scintillators wrapped with a teflon ribbon to increase light collection. The response when irradiated with 511 keV photons from $^{22}$Na and 662 keV photons from $^{137}$Cs was compared to direct photopeak deposition from $^{137}$Cs X-rays with energy around 20 – 30 keV. Good photopeak response for both scintillators that far exceeded the noise of the detector was observed. Comparison with direct X-ray spectra showed that 662 keV photopeak from CsI(Tl) coupling corresponds to deposited energy 51 keV and in case of 511 keV, it was 62 keV. For LSO coupling it was 9 keV and 14 keV respectively, this is far above the noise of PD under the room temperature. In the case of LSO, where the scintillator crystal was larger then sensitive area of PD part, photons were lost and therefore the response was not as good. In optimal geometry like with CsI(Tl) estimation showed that at least 20% improvement in response can be expected. Under 511 keV irradiation, energy resolution was 9.5% FWHM for CsI(Tl) and 28.8% FWHM for LSO.
Electrical characterization of Si PD array verified that each pixel was operational and average reverse current was 420 pA/pixel depleted under a bias of 40 V.

The prototype of the PET detector based on PD array wire bonded to the hybrid with 128 channel charge sensitive preamplifier and multiplexer chip VIKING was tested. A High Energy Physics (HEP) strip detector DAQ system has been used. The uniformity of photoresponse of pixels in the PD array was undertaken using an optical stimulation method by a pulsed laser diode beam ($\lambda = 700$ nm) and a 64 channel parallel readout with Viking IC. The results demonstrated good uniformity of response ($\pm 12\%$ variation across the array). It was observed that there was slight reduction in response of some pixels of the Si PD array that correlated with the length of connecting pads that introduced additional capacitance. However, this correlated response can easily be taken into account when the final PET module is commissioned.

Uniformity of response within single pixel of PD array was investigated with a 50 micron (spot size) laser diode beam by scanning it over the pixel in step sizes of 100 microns. Uniformity of response was better than 3%.

The response of the PD array to gamma radiation was investigated by the optical coupling of each pixel of the array to single 3x3x3 mm$^3$ CsI(Tl) and 4x4x10 mm$^3$ LSO, carried out in a like manner to how it was done on a single PD. The energy resolution and amplitude of photopeak for 662 keV photons were measured. All undamaged pixels clearly demonstrated the photopeak well separated from the noise. Quite a large spread of response ($\pm 28\%$) was observed and is related to the non-ideal optical coupling technology of the single 3x3x3 mm$^3$ scintillator to the PD array. In each process of coupling it was impossible to keep the same thickness of optical grease and adhesive. Also, measurements were taken with modular electronics (single ORTEC Charge sensitive PA) in contrast to direct wirebonding to VIKING chip and this increased length of connecting wires produced additional noise.
We investigated the cross talk between pixels of the array using coupled CsI(Tl) scintillator and collimated 662 keV photon irradiation. No cross talk was observed for passive and biased neighbouring pixels. This was an important achievement of the developed technology for PET detectors.

Timing characteristics were investigated on a single pixel 3x3 mm$^2$ Si PD when it was irradiated with alpha particles with range in silicon similar to absorption length of 420 nm photons. The readout set-up was similar as for PET detector module with timing signal from the rear N$^+$ contact of PD array. This demonstrated the possibility of achieving timing jitter not more than 0.7 ns, which is suitable for PET animal scanner.

The S/N ratio for developed PD array was investigated using $^{241}$Am 59 keV X-rays with energy deposition equivalent to the estimated response from using pixelated LSO proper coupled to the pixel array. This demonstrated that the energy photopeak was easily separated from the noise.

Characterization of the first version of PD Si array demonstrated that the developed technology and approach are satisfactory for the future PET detector module. We found some shortcomings in the first version of the detector. As a result of this research the following recommendations have been developed (and have been successfully implemented in the later technology version SPAD-3).

1) To improve the passivation of pixel sensitive area and QE of Si pixel array using double antireflective coating based on SiO$_2$ and SiN$_4$.

2) To improve the optical coupling technology by using higher refractive index glue.

The new photodiode pixel with the proposed changes implemented obtained an energy resolution of 13% FWHM when optically coupled to LSO and irradiated with 511 keV photons (so far the best found in literature) and a QE of approximately 80% at wavelength 420 nm.
References:


21. Huber J.S., Moses W.W., Derenzo S.E., Ho M.H., Andreaco M.S., Paulus M.J., and Nutt, R. “Characterization of a 64 channel PET detector


49. ibid, pp 75-76.


