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Charge collection in PET detectors

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Charge Collection in PET Detectors

By

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Abstract

Improving the spatial resolution in PET will increase the chance of accurate diagnosis of cancer. PET detectors are comprised in part of detectors such as photomultiplier tubes or solid state photodetectors optically coupled to scintillation materials, such as BGO or LSO. In this thesis a study of recently developed photodiode detectors, SPAD4 photodiodes, and LYSO crystals has been undertaken to investigate these properties.

As part of this research IV (current-voltage) and CV (capacitance-voltage) characterization measurements were carried out on the photodetectors and results indicate they are of excellent quality. Ion Beam Induced Charge Collection (IBICC) measurements were undertaken to fundamentally investigate the effect of bias on the efficiency and uniformity of charge collection within the photodetector. A reverse bias operating voltage of 50 V was eventually chosen as the optimum bias voltage for the SPAD4 photodiode for the gamma ray spectroscopy and timing resolution experiments that followed the IBICC characterisation. Charge collection uniformity was found to be excellent at this optimum voltage.

An investigation into the surface treatment of LYSO crystal and cladding was also completed. Physical measurements and simulations were conducted and the results are compared. Past simulation results have demonstrated that scintillation crystals produced a higher output when the surface was rough as opposed to the industry standard smooth polished surface [1]. Physical measurements and comparisons have also been completed with saw cut finish crystals and chemically etched crystals [2, 3]. Simulations were completed with “Scintillation Program”, a simulation code developed at the CMRP, and DETECT2000. Measurements completed for this thesis produced no significant change in the deduced scintillator light output after roughening a crystal side.

Timing resolution in commercial PET scanners ranges from 8 to 16ns. Experimentally with a photomultiplier tube coupled to a photodiode, timing resolution as low as 9.4ns FWHM has been achieved [4]. Coincidence timing resolution measurements using standard NIM (Nuclear Instrumentation Module) electronic modules were taken with a
NaI-photomultiplier tube and the LYSO-SPAD4 photodiode module. The best result obtained was a FWHM coincident time resolution of 22 ns, which is similar to past results.

The SPAD4 photodiode and LYSO crystal combination show promise as the basis for a future PET detectors module, although further work needs to be completed to improve the timing performance.
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Notation/Abbreviation
CMRP = Centre for Medical Radiation Physics
UOW = University of Wollongong
PET = Positron Emission Tomography
CT = Computed Tomography
MRI = Magnetic Resonance Imaging
IBICC = Ion Beam Induced Charge Collection
PMT = Photomultiplier Tube
PD = Photodiode
LYSO = Lutetium Yttrium Oxyorthosilicate
LSO = Lutetium Oxyorthosilicate
BGO = Bismuth Germanium Oxide
GSO = Gadolinium Silicate
FWHM = Full Width at Half Maximum
IV = Current Voltage
CV = Capacitance Voltage
SLICE = Scintillator Light Collection Efficiency
Int = Integration
Diff = Differentiation
CFD = Constant Fraction Discriminator
TFA = Timing Filter Amplifier
TAC/SCA = Time to Amplitude Converter/Single Channel Analyser
Na–22 = Sodium 22
Cs–137 = Caesium 137
I-125 = Iodine 125
NECR = Noise Equivalent Count Rate

Publications List