Evaluation of a novel inorganic scintillator for applications in LDR brachytherapy using both TE-cooled and room temperature SiPMs

Michael Martyn*[a], Wern Kam[b], Agnese Giaz[c], Simona Cometti[c], Romualdo Santoro[c], Peter Woulfe[a,b], Massimo Caccia[c], and Sinead O' Keeffe[b]
[a]Department of Medical Physics, Galway Clinic, Galway H91 HHT0, Ireland; [b]Optical Fibre Sensors Research Centre, University of Limerick, Limerick V94 T9PX, Ireland; [c]Università degli Studi dell'Insubria, via Valleggio, 11, Como, Italy.

ABSTRACT

This work considers the use of an optical fiber sensor, employing a Gd_2O_2S:Tb inorganic scintillator, for applications in LDR brachytherapy for prostate cancer. Gd_2O_2S:Tb is characterized by a scintillation decay time of ~500 μs, implying that each primary gamma interaction produces a series of single photons, requiring the use of adequate detectors, such as Silicon Photomultipliers (SiPMs). These devices suffer from a significant Dark Count Rate (DCR), undermining system sensitivity. This work reports the result of a feasibility study where identical SiPMs, but different packages, are compared. Specifically, a room temperature SiPM in a ceramic package and a TE-cooled SiPM in a TO8 package. In the former, the optical fiber is in direct contact with the sensor surface, while in the latter there is a separation of ~3 mm. The signal, measured as Photon Count Rate (PCR), in excess of the DCR, was measured in a water phantom at distances of 5 mm and 30 mm from an I^{125} source. For the TE-cooled SiPM, the DCR dropped by ~96% as expected, and the PCR dropped by ~80%, compared to the room-temperature SiPM, due to reduced light acceptance. However, incorporating an optical coupling system into the TE-cooled SiPM, to improve acceptance, resulted in sensitivity increases of 332% and 296% at distances of 5 mm and 30 mm respectively, compared to the room-temperature SiPM. It is hoped that these improvements in sensitivity, will allow for accurate monitoring of the dose-rate from LDR sources, within the clinically relevant treatment volume for prostate cancer.

Keywords: Optical fibers, SiPM, TE-cooled SiPM, Brachytherapy,

1. INTRODUCTION

Radiotherapy refers to the use of ionizing radiation to treat diseases such as cancer. At some point during their treatment, 50 – 60 % of patients require some form of radiotherapy [1]. Radiotherapy is typically delivered in the form of external beam radiotherapy (EBRT), using a linear accelerator, or brachytherapy, which involves the use of radioactive sources. Brachytherapy can be subdivided into Low Dose Rate (LDR) and High Dose Rate (HDR). Sources with dose rates of < 2 Gy/h are classified as LDR, whilst sources with dose rates of > 12 Gy/h are classified as HDR [2]. Brachytherapy treatment can employ permanent implantation of radioactive sources known as seeds (common for LDR treatment of prostate cancer), or temporary implantation of a radioactive source via a catheter (common for HDR treatment of prostate and gynecological cancers).

Precise positioning of the radiation source is crucial to ensure the target area receives sufficient dose to fulfil the objective of the treatment (curative or palliative), whilst minimizing the dose to nearby healthy tissues and organs at risk, such as the bladder, urethra and rectal wall (for prostate / gynecological cancers). Currently, imaging for brachytherapy planning and source placement / localization relies on ultrasound imaging and or pre / post-treatment imaging via CT or MRI [3]. However, these methods for imaging the patient anatomy and identifying source location do not represent real-time independent monitoring of the actual dose delivered to the patient, and as such, create an environment where treatment delivery errors may go undetected. The Horizon 2020 funded ORIGIN Project (Grant Agreement ID: 871324) aims to address these shortcomings in the current treatment delivery practices, and the urgent need to provide real-time in-vivo dose imaging and source localization methods, by developing a new optical fiber based sensor system to support diagnostics-driven therapy through enhanced adaptive brachytherapy.

*michael.martyn@galwayclinic.com; phone 00353 91 785981

To this end, this work considers the use of a novel optical fiber sensor, employing a Terbium-doped Gadolinium Oxysulphide (Gd_2O_2S:Tb) inorganic scintillator, for applications in LDR brachytherapy [4]. Gd_2O_2S:Tb powder was obtained from Phosphor Technology Ltd, UK (Type UKL65/F-R1). This scintillator has a primary peak wavelength of 545 nm and a median particle size of 3.5µm. Gd_2O_2S:Tb is characterized by a scintillation light decay time of approximately 500 µs, implying that each primary gamma interaction produces a series of single photons, requiring the use of adequate detectors.

Silicon Photomultipliers (SiPMs) are state-of-the-art devices with single photon sensitivity, however they suffer from a significant Dark Count Rate (DCR), possibly undermining system sensitivity [5]. This feature of SiPMs, combined with the low dose rate and steep dose gradients associated with LDR brachytherapy radiation sources, mean that maximizing the sensitivity of the detector is crucial for applications in LDR brachytherapy. This work reports the result of a feasibility study where identical SiPMs, but different packages, are compared to investigate possible improvements in sensitivity. Specifically, a room temperature SiPM in a ceramic package and a thermoelectrically (TE) cooled SiPM in a TO8 package are considered.

2. METHODS & MATERIALS

2.1 Optical fiber sensor fabrication

A hemi-sphere tip scintillator geometry was employed, with a diameter of 1 mm to match that of the Polymethyl methacrylate (PMMA) fiber core diameter. The hemi-sphere tip was fabricated using a mix of Gd_2O_2S:Tb scintillator powder and NOA 61 epoxy (mixing ratio 3:2) and was affixed to the PMMA fiber core and cured using an externally applied ultraviolet (UV) lamp.

2.2 Water phantom set-up

A PTW MP3-XS water phantom system (PTW, Freiburg, Germany), filled with sterile water, was employed in this work for obtaining all measurements. This set-up allows for clinically relevant measurements to be obtained, in a standardized, reproducible, and dosimetrically accurate environment. The radiation source (AgX100 I^{125} seed produced by Theragenics Corporation) and optical fiber sensor were positioned within the water phantom using custom 3D printed components. The optical fiber sensor was aligned with the I^{125} seed (activity 0.39 mCi), which remained in position throughout the experiment to reduce measurement uncertainty. The output end of the optical fiber could be connected to the SiPMs considered, without disturbing the position of the source.

2.3 Photon counting experimental set-up: Room-temperature SiPM

The CAEN SP5600E Educational Photon Kit was employed for the purposes of evaluating the room-temperature SiPM [6]. The components of this kit which were employed were the CAEN SP5600 power supply and amplification unit (PSAU) and the front mounted Hamamatsu SiPM [model S13360-1350CS], with an active area of 1.3 x 1.3 mm^2 and a pixel pitch of 50 µm. The CAEN DT5720A desktop digitizer was also included in the experimental set-up, to enable the use of the CAEN control software (i.e. this component must be connected to the PC to employ the CAEN control software). As shown in Figure 1, the optical fiber sensor is interfaced to the SiPM via an FC terminated connection. Furthermore the PSAU and digitizer are interfaced to the laptop via USB 2.0 connections, allowing for readout and analysis of measurement data.

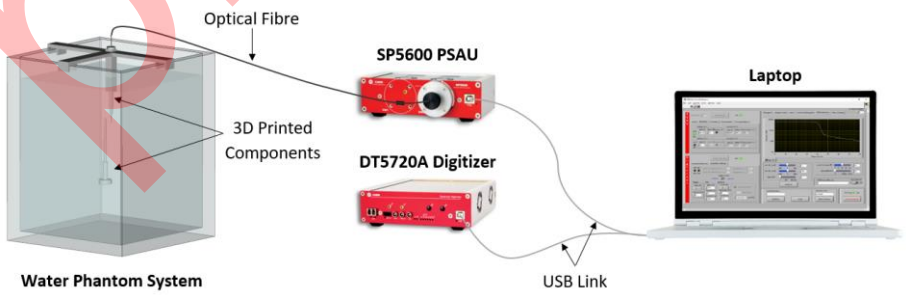

Figure 1. : Schematic diagram of the room-temperature SiPM experimental set-up.

2.4 Photon counting experimental set-up: TE-Cooled SiPM

The CAEN SP5600E Educational Photon Kit was once again employed for the purposes of evaluating the TE-cooled SiPM. This time however, a TE-cooled Hamamatsu Multi-Pixel Photon Counting (MPPC) module [model C13366-1350GD] was employed. . Where a TE-cooler describes a solid-state device which transfers heat against a temperature gradient through the Peltier Effect [7]. Once again this particular SiPM has an active area of 1.3 x 1.3 mm^2 and a pixel pitch of 50 μm. As shown in Figure 2, the TE-cooled module required the use of an external power supply (providing +/- 5 V) and a frequency counter (the CAEN PSAU was employed for this purpose).

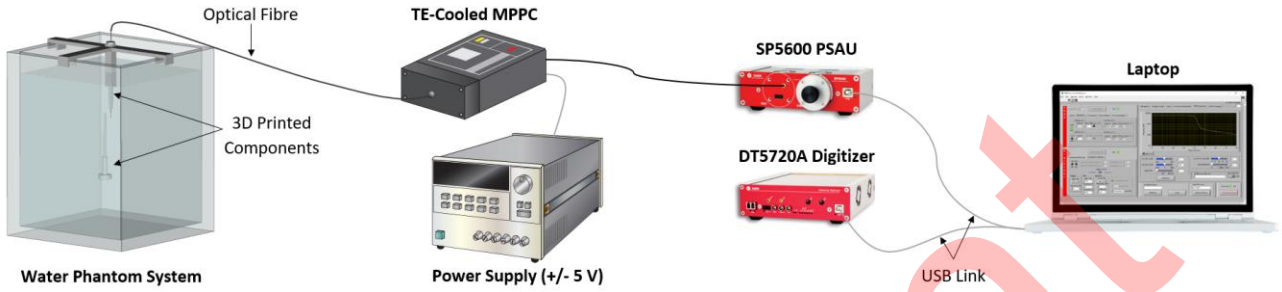

Figure 2. Schematic diagram of the TE-cooled SiPM experimental set-up.

2.5 Photon counting experimental set-up: TE-cooled SiPM + optical coupling system

The experimental set-up described in section 2.4 was once again employed for the purposes of evaluating the TE-cooled SiPM. This time however, an "optical coupling system" was employed to improve the light collection efficiency at the interface between the TE-Cooled MPPC module and the fiber output. This optical coupling system consisted of three components; an aspheric lens (C340TMD-B), a focuser (F230SMA-A), and an adapter (AD1109F).

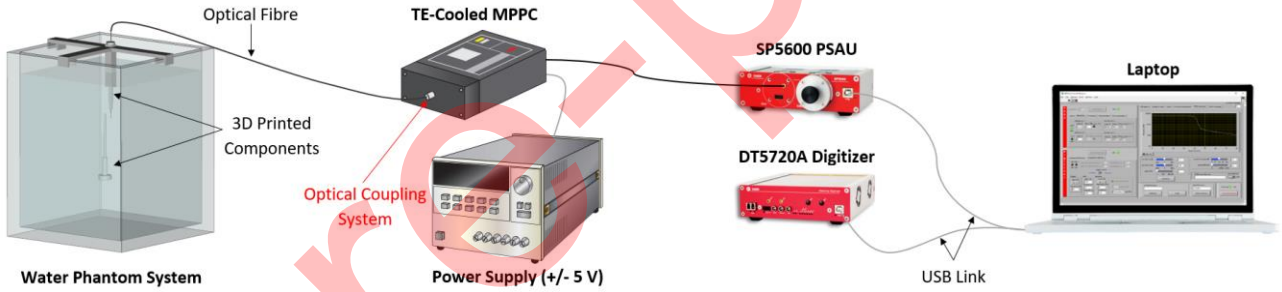

Figure 3. Schematic diagram of the TE-cooled SiPM + full optical coupling system experimental set-up.

3. RESULTS

Results obtained for both the room-temperature SiPM and the TE-cooled SiPM are provided in Tables 1 – 3. Data provided include the Dark Count Rate (DCR), which was measured at a distance larger than 100 mm from the I125 seed, and the Photon Count Rate (PCR) in excess of the DCR, which was measured at distances of 5 mm and 30 mm from the I125 seed. All counting data presented in Tables 1 – 3 were averaged over a 30 second interval. An additional sensitivity metric is also provided for both distances considered, and is calculated (assuming a Poissonian process) as $PCR/\sqrt{DCR}$ Finally all measurement data were obtained at the 0.5 photo-electron threshold level (single photon interaction level), ensuring maximal preservation of the scintillation signal from the Gd2O2S:Tb.

Table 1: Comparison of DCR, PCR, and sensitivity for the room-temperature SiPM and the TE-cooled SiPM

	DCR (Hz)	5 mm from Source		30 mm from Source	
		PCR (Hz)	$PCR/\sqrt{DCR}$	PCR (Hz)	$PCR/\sqrt{DCR}$
Room Temperature SiPM	60,278	334,374	1,362	6,056	25
TE-Cooled SiPM	2,523	65,312	1,300	1,136	23
Comparison	-95.8 %	-80.5 %	-4.6 %	-81.2 %	-8 %

Table 2: Comparison of DCR, PCR, and sensitivity for the TE-cooled SiPM, with and without the optical coupling system

	DCR (Hz)	5 mm from Source		30 mm from Source	
		PCR (Hz)	$PCR/\sqrt{DCR}$	PCR (Hz)	$PCR/\sqrt{DCR}$
TE-Cooled SiPM	2,523	65,312	1,300	1,136	23
TE-Cooled SiPM + Full Optics	2,627	301,569	5,884	5082	99
Comparison	≈	+361.7 %	+352.6 %	+347.4 %	+330.4 %

Table 3: Comparison of DCR, PCR, and sensitivity for the room-temperature SiPM and the TE-cooled SiPM with the optical coupling system

	DCR (Hz)	5 mm from Source		30 mm from Source	
		PCR (Hz)	$PCR/\sqrt{DCR}$	PCR (Hz)	$PCR/\sqrt{DCR}$
Room Temperature SiPM	60,278	334,374	1,362	6,056	25
TE-Cooled SiPM + Full Optics	2,627	301,569	5,884	5082	99
Comparison	-95.8 %	-9.8 %	+332.0 %	-16.1 %	+296.0 %

4. DISCUSSION

The results presented in Table 1 clearly demonstrate that employing the TE-cooled SiPM results in significant reduction in the DCR (~ 96 % reduction). This finding however is coupled with an observed reduction in the PCR at both 5 mm and 30 mm from the source (~ 80 % reduction), resulting in an overall reduction in sensitivity. Although both SiPMs considered in this work are identical, a reduction in the PCR is observed due to the packages within which each SiPM is housed. For the room-temperature SiPM, housed in a ceramic package, the fiber tip (output) is essentially in contact with the sensitive surface of the SiPM. The TE-cooled system however, employs a TO-8 package, to embed the cooler and seal the sensor, avoiding humidity condensation. This results in an increase in the distance between the sensitive surface of the SiPM and the entrance window (~3 mm). Therefore, when divergent light exits the fiber, it may not reach the SiPM surface and the collection efficiency drops. A method to overcome this limitation of the TE-cooled module was defined, employing an “optical coupling system” to improve the light collection efficiency. This optical coupling system consisted of three components; an aspheric lens (C340TMD-B), a focuser (F230SMA-A), and an adapter (AD1109F).

Table 2 clearly demonstrates the improvements in the PCR, and subsequently the overall sensitivity, which were achieved by incorporating the optical coupling system into the TE-cooled experimental set-up. With an average increase in the PCR of ~ 350% compared to the TE-cooled SiPM without any optimization of the light collection efficiency. Finally, Table 3 shows that although a slight reduction in the PCR remains for the TE-cooled SiPM with the optical coupling system, compared to the room-temperature SiPM, significant improvements in sensitivity were still achieved (increases of 322 % and 296 % at distances of 5 mm and 30 mm respectively).

5. CONCLUSIONS

This work compared two identical SiPMs, one operated at room-temperature [Hamamatsu SiPM model S13360-1350CS] and one TE-cooled [Hamamatsu MPPC model C13366-1350GD]. It has been demonstrated that failure to account for the reduced light collection efficiency of the TE-cooled module (due to a separation between the SiPM sensitive surface and the fiber tip) results in a reduction in sensitivity, compared to the room temperature SiPM. However, significant improvements in the sensitivity of the TE-cooled system can be achieved by incorporating an optical coupling system (aspheric lens + focuser), with the TE-cooled SiPM ultimately outperforming the room-temperature SiPM in terms of sensitivity, by 332 % and 296 % at distances of 5 mm and 30 mm respectively. It is hoped that the observed improvements in sensitivity will allow for accurate monitoring of the dose rate from LDR sources within the clinically relevant treatment volume for prostate cancer.

ACKNOWLEDGEMENTS

The ORIGIN project is an initiative of the Photonics Public Private Partnership (www.photonics21.org), and has received funding from the European Union's Horizon 2020 Research and Innovation Programme under Grant Agreement n° 871324

REFERENCES

[1] Rosenblatt, E., Acuña, O., and Abdel-Wahab, M., "The challenge of global radiation therapy: an IAEA perspective," International Journal of Radiation Oncology, Biology, Physics, 91(4), 687-689, (2015).
[2] Podgorsak, E. B., "Radiation oncology physics," Vienna: IAEA, (2005).
[3] Hellebust, T. P., "Place of modern imaging in brachytherapy planning," Cancer/Radiothérapie, 22(4), 326-333, (2018).
[4] Woulfe, P., Sullivan, F. J., Byrne, L., Doyle, A. J., Kam, W., Martyn, M., and O'Keeffe, S., "Optical fibre based real-time measurements during an LDR prostate brachytherapy implant simulation: using a 3D printed anthropomorphic phantom," Scientific reports, 11(1), 1-8, (2021).
[5] Castro, I. F., Soares, A. J., and Veloso, J. F., "Impact of dark counts in low-light level silicon photomultiplier multi-readout applications," IEEE Nuclear Science Symposium Conference Record (NSS/MIC) (pp. 1592-1596). IEEE (2009).
[6] Mattone, C., Antonello, M., Locatelli, M., Arosio, V., Malinverno, L., Lomazzi, S., and de Asmundis, R., "CAEN Educational: Nuclear and Particle Physics Experiments,".
[7] Terasaki, I., "Thermal Conductivity and Thermoelectric Power of Semiconductors," Comprehensive Semiconductor Science and Technology, 326-358, (2011).