Current-Voltage Response of Cold White LED Strip to Radiotherapeutic Electron and Photon Beams: A Preliminary Study

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Abstract. Photonic devices such as photodiodes, phototransistors, solar cells, and charge-coupled devices (CCDs) have been associated with clinical radiation detection in recent studies. Light Emitting Diodes (LEDs) are of low-cost in comparison to photodiodes and available in various flexible designs, including surface mount devices (SMDs). LEDs are purposely designed to emit light, but they can also be implemented for detecting light in a photovoltaic mode. Despite the LED-related merits, a few studies, compared to other photonic devices, especially photodiodes, have adopted them as medical radiation detectors. This research hence exploits the LED’s dual applicability by investigating a cold/cool white LED strip current-voltage (C-V) signal response to photon and electron beams. Radiation parameters such as beam energy, angle of incidence, field size, surface source distance (SSD), and monitor units (prescribed dose) were varied. According to the obtained results, the electron beams induced a 10% higher C-V signal compared to photon beams. An angular dependence was also observed, i.e., the C-V signal fluctuated due to a variation in the irradiation angles; -45°, -25°, 0°, 25°, 45°, and radiation type – photon and electron beams. The C-V signal response to SSD alterations additionally produced bimodal signal graphs, skewed towards shorter SSDs, for all the beams used in this study. Finally, the C-V signal was generally linear to the prescribed dose for both the photon and electron beams. Overall, the response of LEDs to photon and electron beams separately exhibited a similar signal trend for the beam energy, field size, angular dependence, and SSD parameter variations. On the other hand, dose linearity produced contrasting signal trends across the photon and electron beams.

1. Introduction

Radiation beams and beam energies such as 6 MeV electron beams and 9 MV photon beams are used to treat and contain tumorous cells and tissues. Nonetheless, ionising and non-ionising radiation stimulate different biological effects in human tissue and cells. For instance, radiation can kill cancer cells or can lead to gene mutations that could eventually develop into cancerous tumours [1].

Present technology manipulates photonic devices, i.e., phototransistors, photodiodes, solar cell films, and charge-coupled devices (CCDs) for radiotherapeutic and radiodiagnostic radiation detection.
and quantification [2]. This could perhaps be attributed to silicon diode-based detectors being one thousand times more sensitive than ionization chambers [3]. Conventionally, light-emitting diodes (LEDs) are luminaries [4]. However, recent studies have exploited LEDs for light detection [5], and performance comparison to photodiodes [6].

In the photovoltaic configuration, energetic photon impingement on LED sensitive areas induces photocurrents. Despite this potentiality, limited research has examined LED strips for medical radiation detection.

Similar to photodiodes, LED active areas comprise PN junctions [2]. Therefore, photon absorption by the electrons localised in the N region facilitates electron-hole pair recombinations. In this study, electron and photon beams excite charge that drifts in the form of a current. This radiation-induced current was converted to a voltage and displayed by a multimeter as a Current-Voltage (C-V) signal.

This paper, therefore, investigates electron, and photon beam-induced signals with variation in beam energy, irradiation angle, irradiated field size, and surface source distance (SSD). The signal linearity to dose (MUs) was also examined. Although radiation instantly induces a current, in this study, the measurable parameter is voltage. Therefore, the LED signal was referred to as a current-voltage signal.

2. Materials and Methods

2.1. Experimental Set-Up

A cold white LED strip (12V DC SMD 5050) with 12 chips, in the form of a surface-mount device (SMD), was investigated (Figure 1(b)). Strip luminescence ensured the operability of all the chips. Performing radiation exposures in total darkness, and chip masking – with black vinyl tape, impeded ambient light/spurious signals. A 12 cm thick solid water phantom slab, placed atop the treatment couch of a Siemens Primus model 3347 linear accelerator, barred scattered radiation from striking the LED strip. The LED strip was symmetrically positioned at the geometrical centre of the slab.

Singly exposing the strip to the electron and photon beams (using the set up in Figure 1(a)) instantly induced millivoltage range C-V signals. These signals were measured using a handheld multi-functional digital multimeter (Model JDS2012A, Jinhan Company: Hefei, Anhui, China). Non-zero offset multimeter readings were deducted from the final multimeter reading in order to calculate the net radiation-induced signal. These offset readings implied ambient signal sources, i.e., heat [9] and light [10].

Figure 1. (a) Schematic of the experimental set-up.

Figure 1. (b) Cold white LED strip: Without and with black tape masking.

Figure 1. (c) Super flab bolus atop a solid water phantom.
Compton scattering, photoelectric effect, coherent collision, and pair production are the probable energy transfer mechanisms/interactions that could occur between radiation photons and anatomical particles. Therefore, post-interaction maximum dose build-up, in tissue, could be at a depth of about 1 - 4 cm ($D_{\text{max}}$) [7] – depending on the beam type and energy. Because the structural composition of bolus mimics that of human tissue [8], 30 × 30 cm$^2$ super flab bolus (Figure 1(c)), with 1.5 cm thickness, facilitated dose build-up; hence maximum absorbed dose at the bolus-chip interface. High absorbed dose implied more charge recombinations resulting in a signal detected. The sticky nature of super flab bolus further eliminated air spaces at the bolus-chip interface; this optimised and enhanced dose absorption by the chip.

2.2. Electron Beam Irradiation
Signal dependence on electron beam energy was investigated using 6 MeV and 9 MeV electron beams. Signal angular dependence was also investigated at 0°, 25°, and -25°. However, the electron applicator being 5 cm away from the phantom slab limited the choice of angles. Electron applicator sizes of 10 ×10 cm$^2$, 15×15 cm$^2$, 20×20 cm$^2$, and 25×25 cm$^2$ were employed for field size variations. SSD was varied in steps of 5 cm from 95 to 120 cm. Prescribed dose variations were from 10 up to 100 MUs – in steps of 10. While one parameter was varied for each parameter dependence examination, constant SSD, field size, MU, and irradiation angle values were 100 cm, 20×20 cm$^2$, 10 MU, and 0°, respectively.

2.3. Photon Beam Irradiation
Photon beams of 6 and 10 MV were utilised to investigate the effect of photon beam energy on the signal. Both the 6 MV and 10 MV beams were used for only the beam energy parameter variation, whereas the 6 MV beam was used for the rest of the parameter variations in this study. Signal dependence on the irradiation angle was examined at 0°, 45°, and -45°. Field sizes of 10×10 cm$^2$, 20×20 cm$^2$, 30×30 cm$^2$, and 40×40 cm$^2$ were used for signal-field size dependence investigations. Signal-SSD dependence was examined whilst increasing the SSD from 60 to 120 cm – in 10 cm steps. Similar to electron beam irradiations in section 2.2, the MUs were from 10 up to 100 MU – in the steps of 10.

3. Results
3.1. Beam Energy Response
As observed in Figure 2, the 6 MeV, 9 MeV, and 10 MV electron and photon beams produced appreciably higher signals relative to the 6 MV photon beam. The 6 and 9 MeV beams, respectively, contributed 46.9% and 53.1% of the total electron beam signal. On the other hand, the 6 MV and 10 MV photon beam signals were, respectively, 39.4% and 60.6% of the total photon beam signal.

The photon and electron beams were 43.6 % and 56.3% of the total radiation-induced signal, respectively. From Figure 2 (a) and (b), the 10 MV signal is equivalent to the 6 MeV signal – 1.0 mV.

![Figure 2. (a) Electron Beam Signal Vs Energy Graph.](image)

![Figure 2. (b) Photon Beam Signal Vs Energy Graph.](image)
3.2. Angular Dependence
The highest signal was recorded when both the 6 MeV and 9 MeV electron beams were perpendicular to the chip sensitive area, as observed in Figure 3(a). Further, the maximum C-V signal increased with an increase in the electron beam energy, i.e., the 9 MeV beam produces a higher signal than the 6 MeV beam – as seen in Figure 3(a). The 9 MeV signal was higher than the 6 MeV signal at both -25° and 0°; However, it was less than that of 6 MeV at 25°. Incidence angle alteration from 0° to 25° and -25° resulted in 73.3% and 87% 6 MeV signal depreciation, respectively. Similarly, 91.2% and 73.5% 9 MeV signal drops were observed at 25° and -25°, respectively. Likewise, the 6 MV photon beam signal was angular dependent, and the signal values were 0.3, 0.7, and 0.3 mV at 45°, 0°, and -45°, respectively – as illustrated in Figure 3(b).

Figure 3. (a) Electron Beam Signal Vs Gantry Angle Graph. Figure 3. (b) Photon Beam Signal Vs Gantry Angle Graph.

3.3. Field Size Response
Irradiation during the 20×20 cm² field size showed the highest C-V responses; 1.0 mV and 1.13 mV for the 6 MeV and 9 MeV beams, respectively. Similarly, a peak signal was observed during the 20×20 cm² field irradiation with the 6 MV beam.

Nevertheless, lower signal values were obtained at 10×10 cm², 15×15 cm², and 25×25 cm² fields irradiated with both 6 MeV and 9 MeV beams. In the same way, low 6 MV photon beam induced signals were observed at 10×10 cm², 30×30 cm², and 40×40 cm² fields. These signals were, however, slightly higher than those obtained during the 10×10 cm², 15×15 cm², and 25×25 cm² field irradiations with both the 6 MeV and 9 MeV beams – Figure 4 (a) and 4 (b).

Figure 4. (a) Electron Beam Signal Vs Field Size Graph. Figure 4. (b) Photon Beam Signal Vs Field Size Graph.
3.4. SSD Dependence

Between SSDs of 95 and 105 cm, there was a positive correlation between electron beam C-V signal and SSD. The signal, however, reduced on further SSD increment to 120 cm. Nonetheless, the 6 MeV beam signal was higher than the 9 MeV beam signal, as shown in Figure 5(a).

The 6 MV photon beam signal similarly exhibited the same trend; there was a signal appreciation whilst increasing the SSD from 60 cm up to 90 cm. The signal, however, decreased on increasing the SSD to 120 cm, as elaborated in Figure 5(b).

Figure 5. (a) Electron Beam Signal Vs SSD Graph.

Figure 5. (b) Photon Beam Signal Vs SSD Graph.

3.5. Dose Linearity

There was a positive correlation between the 6 MeV C-V signal and the monitor units (MUs); the signal was linear with an R-square value of 0.65028. However, increasing the beam energy to 9 MeV resulted in a negative correlation between the signal and monitor units; the R-square value also increased to 0.85551, as illustrated in Figure 6(a).

The 6 MV photon beam signal showed a weak correlation to monitor unit variations - compared to the electron beams. It was independent of monitor units in the 20 – 90 MU range; hence the R-square value was 0.05985 – Figure 6(b).

Figure 6. (a) Electron Beam Signal Vs Monitor Units Graph.

Figure 6. (b) Photon Beam Signal Vs Monitor Units Graph.

4. Discussions

An increase in the electron and photon beam energy proportionately increased the C-V signal. However, electron beams stimulated more electron-hole pairs; hence had a higher signal than photon beams. Therefore, electron beams deposited more energy than photon beams, as stated by [1]. The chip response was also angular dependent, i.e., the signal fluctuated depending on the angle between the normal to the chip sensitive area and the beam’s central axis. When this angle is increased from 0°
to 90°, diode output signals could depreciate by 10% [3]. Photodiode and phototransistor detector angular dependence was also observed by Paschoal [11], and they attributed it to detector asymmetry.

The field size regulates the quantity of radiation striking the detector sensitive area. Therefore, large field sizes could imply higher signals and vice versa. In our study, the optimum field size was 20×20 cm². This field covered all the strip chips hence a higher signal generated. As radiation is propagated from its source, its intensity diminishes according to the inverse square law [12]. Therefore, the signal obtained should be inversely proportional to the square of the SSD. However, in our study, SSD increment initially yielded a signal appreciation followed by a decay. This could be attributed to the scattered radiation and interactions such as Compton that could still stimulate charge recombinations.

Because monitor units are calibrated to depict a specific absorbed dose, e.g., 100 MU = 1 Gy, the 6 MeV linear C-V response to MUs implied signal linearity to absorbed dose hence stable sensitivity. The 9 MeV and 6 MV beam nonlinearity could be as a result of charge trapping; semiconductor band gap energy is approximately less than 5 eV – which is very low compared to the MeV range beams.

Since LEDs detect radiation spectrums similar to those they emit [13], they could perhaps feasibly detect low energy radiations, i.e., radiations whose wavelength is equivalent to or longer than the wavelength of the light emitted by the LED. Additionally, amplification and signal refining could be applied to increase the low mV range signal obtained in our findings.

5. Conclusions
LEDs respond to dosimetric parameters such as angular dependence, beam energy response, field size dependence, monitor unit variation, and SSD dependence during photon and electron beam applications. Nevertheless, based on the overall evaluation of all the dosimetric parameters, LEDs could produce better results when used for the detection of low energy radiation; for instance, diagnostic radiology radiation.

Acknowledgments
This work was supported by the Short-Term Grant, Universiti Sains Malaysia, Grant no.: 304/PPSK/6315117. Authors would like to also express their gratitude towards Hospital Universiti Sains Malaysia for the technical support rendered during the execution of this study.

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