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An optocoupler-based method for dosimetry in low energy X-ray beams

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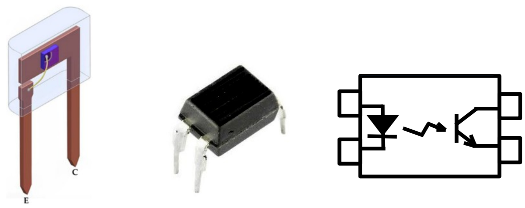
Abstract—**Semiconductor electronic devices have been used as detector in X-ray beams for decades, and the principle of operation is based on measuring the electrical current generated while such a device is under the ionizing radiation. Some devices that can be used are: photodiode, phototransistor, bipolar junction transistor, MOSFET, among others. In this work, a method based on an optocoupler device is presented for radiation dose real-time measurement in low-energy X-ray beam, which is used in medical radiology. Such an optoelectronic device has an LED and a phototransistor inside it, in general, for signal communication between two circuits that have to be electrically decoupled. Therefore, the light emitted by the LED excites the phototransistor. To develop an innovative method for radiation detection, the output signal from the phototransistor is used to feed back the current source of the LED so that the phototransistor is kept at a constant operating point by the photonic intensity of the LED, which is shielded from the ionizing radiation beam to be detected. When the optocoupler is irradiated an additional signal on the phototransistor is superimposed on the signal originated by the LED. Then, a time-integrated error signal is digitized and stored in real-time. Each step of the error signal is proportional to the accumulated radiation dose during the device exposure. This innovative method was tested under different conditions varying both peak kilovoltage (kV) and workload (mAs) in order to compare with results from other electronic devices. The experimental results showed that the optocoupler-based method for X-ray dose measurement works and consequently may become an option for X-ray dosimetry.**

Keywords — **Optocoupler, X-ray, Dosimeter.**

I. Introduction

Semiconductor devices such as photodiode, MOSFET, bipolar junction transistor (BJT), and other devices are widely used as an ionizing radiation detector [1-5]. When the *pn* junction (photodiode) is under X-rays, pairs of electron-holes are generated and a current in the device $i_d(t)$ [1] can be measured. i_d is proportional to the intensity of the radiation beam and also the energy of the incident photons. While on the subject, regarding the effective energy of X-ray photons used in medical diagnosis, it can vary between 50keV and 100keV depending on the peak potential applied to the X-ray tube, generally between 70kV and 150kV, respectively. Furthermore, the mean energy of the X-ray beam also depends on the radiation filtration, which can change the spectrum from soft to hard X-ray [6-7]. Indeed, the probability of interaction of X-ray photons depends on the material. For low atomic numbers and energies between 50keV and 100keV pratically more than 90% of the interactions corresponds to the Compton effect and less than 10% are the other effects [8].

A typical phototransistor, which can be seen as a BJT with transparent encapsulation, has the base-collector *pn* junction operating as a photodiode and the photocurrent, i_{ph}, is actually the input bias or base current (I_B=i_{ph}), which is amplified by the transistor gain, β, resulting in the output collector current, I_C=$\beta \cdot I_B$. Generally, a typical phototransistor has only two terminals: collector and emitter, and it is called as floating base device (Fig. 1a), since only the photocurrent i_{ph} is needed to make the device operate as a light photon detector.



(a) (b) (c)

Fig. 1. a) Transparent floating base phototransistor [4]; b) Typical optocoupler which has an opaque package; c) An optocoupler symbol.

If the phototransistor is used as an X-ray detector, in a typical device with transparent package, there will be two components of the noisy current, i_n: 1) photocurrent produced by the ambient light; 2) current due to the electron-hole pairs produced by the effect of room temperature. A phototransistor that has an opaque package (an ambient light filter) will obviously not have the first noisy component or at least it can be about 40dB weaker than the desired output signal. In fact, it was one of the reasons why the optocoupler was chosen to develop the dosimetry method for X-ray used in diagnosis. Following Santos [4], one can assign the current due to X-ray photons as I_{BX}, which is amplified and results in I_{CX}, i.e., the current due only to the interaction of X-rays on the phototransistor chip. It is important to note that, in the case of X-rays incident on the transistor, the thickness of its package contributes to the build-up cap effect [9], and I_{CX} would be actually a current greater than that of a transistor without a package, even if it is opaque or transparent. Therefore, I_{CX}, is produced by the device as a whole (package + chip) [4].

If a BJT is used as a detector to determine some parameter of a radiation beam, such as the photon energy, collimation techniques must be used to avoid errors due to scattered photons. However, for dose measurement in a patient, it is important to measure the total dose delivered to him, with the contribution of the scattered radiation. Therefore, the X-ray beam applied to the diagnosis must not be collimated in order to can perform dosimetry in a patient, or even in a certain position within a phantom.

Another important thing is about the effect of the directional dependence. The angle of incidence of the X-ray photon beam on the transistor changes its response, *i.e.*, I_{CX} varies with the device angle. It was observed that as the photon incidence angle increases (up to ≈50º) on a BJT there is an increase in the current I_{CX} [4]. This is due to the increase in the path of the X-ray photon in relation to the face of the chip itself, *i.e.*, the increase in the thickness of the device package (build-up cap effect) because of the angle change. For angles greater than 50º the current I_{CX} start to decrease [4]. Based on the explanations given above, a dosimetry method using an optocoupler was developed, as detailed next.

II. Material and Methods

A. Basic circuit

In general, an optocoupler has the pin diagram as shown in Fig. 1c, and terminals A and C are anode and cathode, respectively, corresponds to the LED where the input current, I_{LED}, flows from a current source. The output signal is often from a floating base phototransistor, and terminals C and E are collector and emitter, respectively. Fig. 2 shows a circuit to measure the output current, I_C. For I_C measurement a series ammeter must be connected with the voltage source (source-meter instrument). The ratio between I_C and I_{LED} is referred to the current transfer ratio (CTR) of the optocoupler, which varies nominally between 50 and 500.

B. Optocouplers

Two types of optocouplers were selected to carry out the experiments: 1) SFH620A-2, Vishay®, with CTR≈130; 2) HMHA2801A, Infineon®, with CTR≈100. The last one has mini-flat SMD (Surface Mounted Device) type package and is soldered directly to its specific printed circuit board (PCB). The first one is for AC input signal, therefore has two LEDs, one for positive bias and other for negative bias. This choice allows testing the sensitivity of the phototransistor from two LEDs, providing lower relative uncertainties. In addition, it has a DIP-type package and this will allow for some comparison with the SMD-type package optocouplers.

C. PCB and shielding of the LED

Two types of PCBs were made: one for the SMD-type device and other for the DIP-type device. In the last one a DIP socket was soldered in the PCB, which was fixed in a plastic holder (Fig. 3). This technique was required for the following reasons: 1) to be able to hold the sample to be irradiated at a given angle, θ; 2) to allow the LED to be shielded from the incident X-ray photons; 3) to exchange the optocoupler device with practicality and make statistics. As can be seen in Fig. 3, the optocoupler is placed so that the LED must be at the end of the PCB to shield only it.

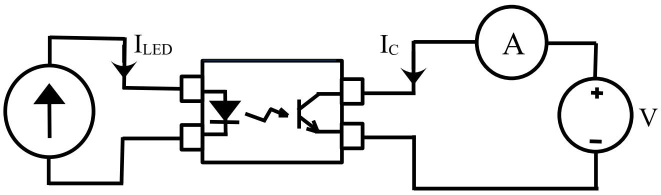


Fig. 2. Basic circuit for measuring the output current I_C of an optocoupler.

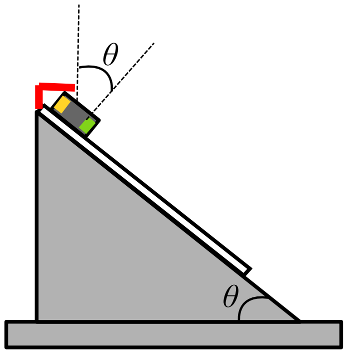


Fig. 3. Illustration of a plastic holder (gray); with PCB (white); LED shield (red); and the optocoupler with its LED (orange) and phototransistor (green).

D. Radiation generator and X-ray tube parameters

The radiation generator was a typical clinical equipment for X-ray beam applied to the diagnosis: Polymat 30/50 plus, Siemens®. The X-ray tube parameters chosen in the control panel of the clinical equipment were: 1) 1000 ms pulse time; 2) 73kV, 102kV and 125kV peak kivoltages; 3) 50 mAs, 100 mAs and 200 mAs for workloads. Each point of each graph presented in results corresponds to an average of two readings.

E. Experimental setup

Fig. 4 illustrates the experimental setup for irradiation procedures. θ is the incidence angle of the radiation beam in relation to the surface of the phototransistor chip face in the optocoupler. Therefore, the photons of light coming from the LED arrive perpendicularly at the phototransistor chip face, whereas X-ray photons reach the chip with the angle θ.

F. Instruments

The main instruments used in the experiments are: 1) 2450 source-meter, Keithley®, working as a current source, I_{LED}. 2) a 6430 source-meter, Keithley®, working as voltage source, V_{CE}, and current meter, I_C; 3) an EFF1705, 2 channel source-meter, Scients®, for real-time measurements in the clinic.

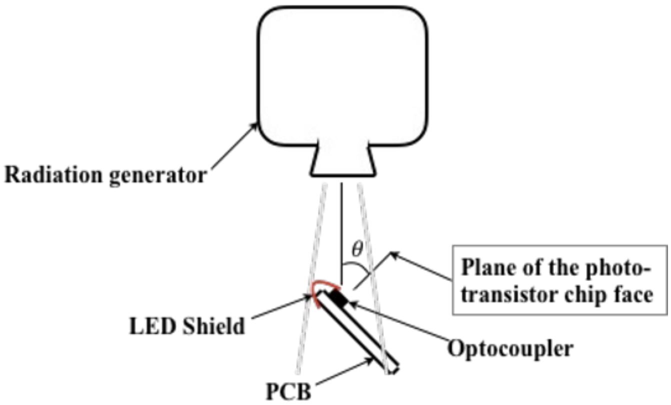


Fig. 4. Experimental setup for irradiation procedures.

G. Basic methods

First of all, an evaluation was carried out to check which angle of incidence, θ, the optocoupler should have for an optimized efficiency, *i.e.*, we are interested in obtaining the highest value of current I_{CX}. The values of angles chosen for this experiment were: 30º, 45º and 60º, following Santos [4].

Secondly, in dosimetric techniques, generally there are two important parameters to take into account: 1) the detector response as a function of the X-ray tube peak potential; and 2) the detector response as a function of dose rate, which is equivalent to say the device response as a function of X-ray tube current. For example, the X-ray photonic intensity with workload Q_I=100mAs is 50% less than the workload of Q_2=200mAs, both during a time of 1s. However, I_{CX} could not be linearly proportional. In other words, I_{CX} does not necessarily obey a linear function so that $I_{CX1}=I_{CX2}/2$. The X-ray tube peak potential chosen were: 73 kV, 102 kV, and 125 kV. These choices represent approximately 100kV±25%. The workload was set to be: 50 mAs, 100 mAs, and 200 mAs. In general, 50 mAs is lower workload normally used at the clinic, therefore this will be the worse case.

The third method is to assess how the phototransistor responds to the radiation dose step. Although knowing that the response of the detector must be analyzed during the exposure of the radiation beam, as explained above, the most important in this study is the variation of the electrical state of the phototransistor inside the optocoupler, since it is known that such a device degrades after each radiation dose step [4]. In this experiment one must deliver systematically dose steps to the device, so that be possible to analyze if there is a correlation with the variation of I_C and the accumulated radiation dose. Therefore, the characteristic curves of $I_C \times I_{LED}$ were plotted before and after irradiation to make comparisons.

H. Proposed technique

The three initial methods described above will culminate in the proposal of this work. Fig. 5 can help us clarify that. Based on the results of the phototransistor response under radiation beams, it can be seen that I_C decreases with the accumulated radiation dose [4]. It is actually a decrease in the transistor gain after each X-ray dose. Therefore, at each radiation pulse, an error signal can be registered on a microcontroller (µC), for example. The µC corrects the signal and changes the bias, I_{LED}, so that the phototransistor operating point is returned, *i.e.*, as V_{CE} is constant, the current of the LED is increased so that the I_C returns to the previous value. Practically, this is a digital feedback. Using the timer in the µC and the registered error signal, both dose rate and dose on the irradiated device can then be measured, after a calibration procedure.

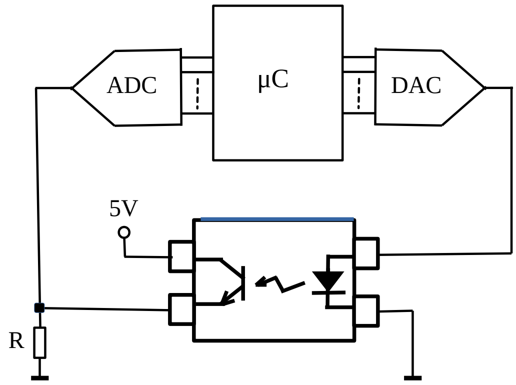


Fig. 5. Simplified circuit schematic of the optocoupler-based method for dosimetry in low energy X-ray beams.

III. Result and Discussions

First of all, it was observed that at the angle of θ=30° I_{CX} was less than 50% compared with θ=45°. At the angle θ=60° the LED shield itself covered part of the phototransistor and I_{CX} also decreased. Therefore, the chosen angle was 45°.

Fig. 6 and 7 show the real-time I_{CX} measurements for the chosen peak potentials and workloads, for the two types of optocouplers. Note that it does not depend on the type of optocoupler, i.e. thicker (DIP) or thinner (SMD) package. Also note that the behavior of the detector response is independent of the CTR, that is, it does not matter what type of optocoupler is used as the response to the X-ray beams is similar. However, a higher CTR can provide a higher output signal.

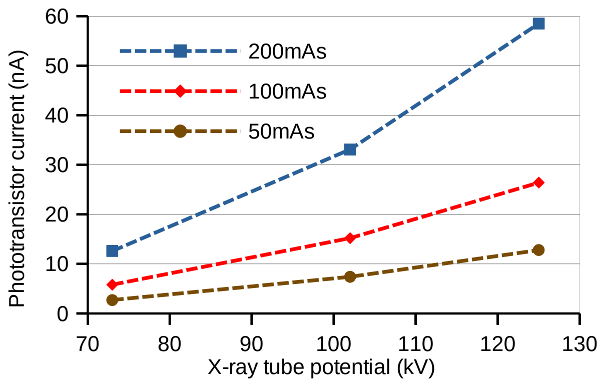


Fig. 6. I_{CX} as function of X-ray tube potential, parameterized by the workload for the HMHA2801A optocoupler (SMD type).

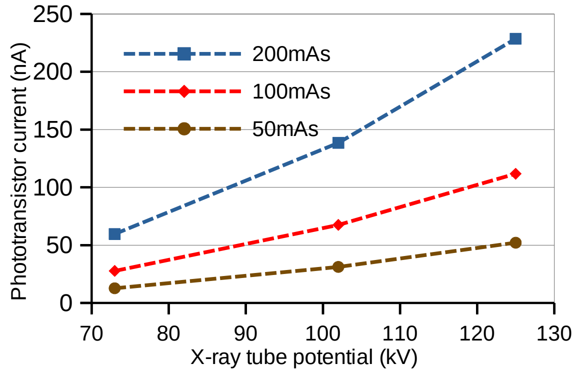


Fig. 7. I_{CX} as function of X-ray tube potential, parameterized by the workload for the SFH620A-2 optocoupler (DIP type).

Also, one can observe that the DIP type optocoupler has a higher I_C output signal than the SMD type, as this is due to the buildup cap effect since the CTR of both are similar.

The most important result of this work is shown in Fig. 8. It is I_C as a function of I_{LED}, parameterized by each 1Gy dose step. These data correspond to a SFH620A-2 optocoupler. It is worth noting that 1Gy dose is equivalent to more than one thousand (lung) radiography exams [6]. Therefore, it can be said that, in X-ray beams applied to medical diagnosis, the optocoupler lifetime could be long, i.e., it can be used at least for 3000 exposures.

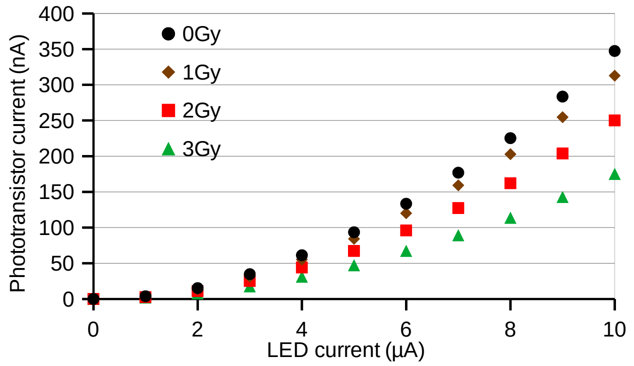


Fig. 8. I_{CX} as function of X-ray tube potential, parameterized by the workload for the SFH620A-2 optocoupler (DIP type).

IV. Conclusions

An optocoupler-based dosimetry method for low-energy X-ray beams was presented. For this purpose, the optoelectronic device must be placed at 45° angle in relation to the face of the phototransistor chip inside the optocoupler. As there is gain degradation in a transistor after irradiation, basically, the phototransistor operating point must be corrected from a digital feedback applied to the LED inside the optocoupler. Also the LED must be shielded. In this way, each X-ray pulse generates an error signal that is recorded. The results show that each dose step corresponds to the LED current step stored by the digital circuit, which is previously calibrated as a function of the dose. It can be concluded that this innovative method works and may become an option for low energy X-ray dosimetry in the near future.

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