Proceedings of the Korea Nuclear Society Autumn Meeting
Seoul, Korea, October 2001

Development of Solid-State Detector for X-ray Computed Tomography

S.W Kwak[1)], H.K Kim[1)], Y. S Kim[1)], S.C Jeon[1)], G. Cho[1)], Y. Yi [2)], S.W. Yuk[2)], J.B Park[3)], S.M Goh[4)]

[1)]Korea Advanced Institute of Science and Technology, 373-1 Kusong-dong, Yusong-gu, Taejon 305-701 Korea
[2)]Korea University, 1, 5-Ka, Anam-dong Sungbuk-ku, Seoul,136-701 Korea
[3)]LISTEM Corp., 414-1, Chongchon-2 dong, Pupyong-gu, Inchon, Korea
[4)]Chonbuk National University, 664-14 1ga Duckjin-Dong Duckjin-Gu Chonju Chonbuk 561-756 Korea

Abstract

Solid-state detector has recently been used for x-ray Computed Tomography (CT) because it has high light conversion efficiency, quantum efficiency and compact size, compared with Xenon gas detector. In order to reduce radiation dose to the patient in medical x-ray CT, it should be needed to maximize the quantum efficiency and signal-to-noise ratio (SNR) of the detector. In this study, 16-channel detector array consisting of a $CdWO_4$ scintillator, glued onto a p-i-n photodiode has been prepared as a sample CT detector. We investigated overall detector sensitivity with respect to variation of the p-layer thickness because the p-layer mainly restricts the signal-to-noise performance of the detector by increasing the leakage current and decreasing the light photon signal from the scintillator as its thickness increases. From the measurement, we obtained a reasonable dynamic signal response as a function of the incident x-ray magnitude. However, comparing with a reference detector (Analogic cooperation), about 2.5 times lower signal magnitude was achieved even with < 2 μm thickness of the p-layer, which is probably due to the quality of the layer made by the diffusion process.

I. Introduction

X-ray computed tomography is one of the most important diagnostic techniques because it gives accurate information about patient disease. For medical x-ray computed tomography (CT), so far the most popular detector type is a solid-state sensor array composed of a scintillation material and a photo-sensitive device such as a silicon photodiode [1,2]. High radiation absorption efficiency, photosensitivity, large dynamic range, fast response time, low patient dose and noise are required for XCT. In the design of detector element, the signal-to-noise ratio (SNR) should be maximized in order to reduce the patient radiation dose or to extract the maximum information from the given dose. When the photodiode is used as the detector element, its leakage current mainly contributes to the shot noise and thus governs the overall noise performance. For the purpose of the CT application, photodiode is used for sensing light from CdWO4 scintillator. Photodiode must satisfy stringent requirements such as high quantum efficiency at operating wavelength, high response time and minimum noise. In this paper, the effect of p-layer thickness on the quantum efficiency and the noise performance of the photodiode was also studied by measuring and simulating the actual photodiode. 16-channel detector array consisting of a $CdWO_4$ scintillator and a silicon photodiode with p-i-n structure has been fabricated for prototype XCT detector and basic performance of the detector was measured and compared with the commercial product. Based on the result of this paper, further designs of photodiodes will be discussed for the better performance of the CT detector

II. Materials and Methods

For the construction of a photodiode with p-i-n structure, an n-type silicon wafer with high volume resistivity (1,200 – 4,000 Ω·cm) was prepared because this should become an intrinsic layer. Then, successive doping by diffusion and typical photolithography processes realized the structure of p-n$^-$-n$^+$. Two kinds of photodiode are fabricated. One has the i-layer thickness of about 77 μm and the p-layer thickness of about 3 μm, the other the i-layer thickness of about 78.3 μm and p-layer thickness of about 1.7 μm. Silicon nitride layer was finally coated onto the p-layer for the passivation. It is noted that, in the design of our photodiode, the use of a guard ring around the photodiode active area was considered, which might provide a sink for the leakage current generated outside the active area. The fabricated photodiodes were characterized by measuring the leakage current at darkness as a function of the applied bias with a semiconductor parameter analyzer (HP 4155A). For qualitative discussion, the electrical behavior was also estimated by using a device simulation tool called ATLAS (Silvaco International). ATLAS is just a boundary-value partial-differential-equation (PDE) solver, and thus numerically solves the Poisson and carrier transport equations on the grid points of a given device structure. A $CdWO_4$ scintillating crystal was coupled onto the fabricated photodiode, which is typical structure of the solid-state CT detector. Figure 1 shows the constructed 16-channel detector array with a collimator. Each element of the crystal array has a dimensions of 1×3×20 mm^3, and its side-surfaces were polished and Teflon-taped for the enhancement of light collection efficiency and anti-cross talking among the crystal elements, respectively. The response for an x-ray beam with respect to various water depths was measured and compared with the results from a commercial CT detector (Analogic cooperation).

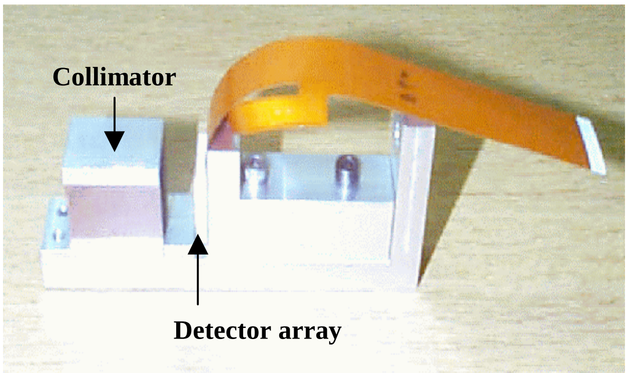

Figure 1: A photograph of 16-channel sample detector array, which consists of $CdWO_4$ scintillators and silicon photodiodes with a p-i-n structure. The detector array sits on the aluminum mount and the x-ray beam collimator is also included.

III. Results and Discussion

Dark reverse leakage current of the photodiode with an n-type guard ring as a function of the applied bias is shown in Fig. 2 in which the calculation results from the device simulation software, ATLAS are also included. The reasonable leakage current, ~nA was measured at interesting bias region (or at zero bias). The calculation result below the reverse bias of 12 V, which is the full-depletion bias voltage, agrees well with the measurements. Disagreement at high reverse bias voltage regions might be understood by the coarse grids (or nodes) defined in deep region of the i-layer for the calculation. An interesting region in the p-i-n photodiode is, in general, restricted to the p-i interface because it develops the depletion region where the light interacts and then contributes the electrical signal. Since our photodiode has somewhat thick i-layer (~77 μm), we only defined fine grids around the p-i interface region accounting for the computation time. Much better agreement is expected when defining fine grids for the calculation thorough whole region of the photodiode. It is noted that there is an abrupt increase of the leakage current around the full-depletion bias voltage. We believe that this is firstly observed through this study. We presume that the reason is due to the decrease of the bulk resistivity when two full-depletion regions at p-i and i-n interfaces meet. From the simulation results as shown in Fig. 2, it is observed that the employment of a p-type guard ring gives higher leakage current rather an n-type guard ring. This may be due to the additive leakage current from the another depletion layer made at the interface of the p-type guard ring and the n-type substrate.

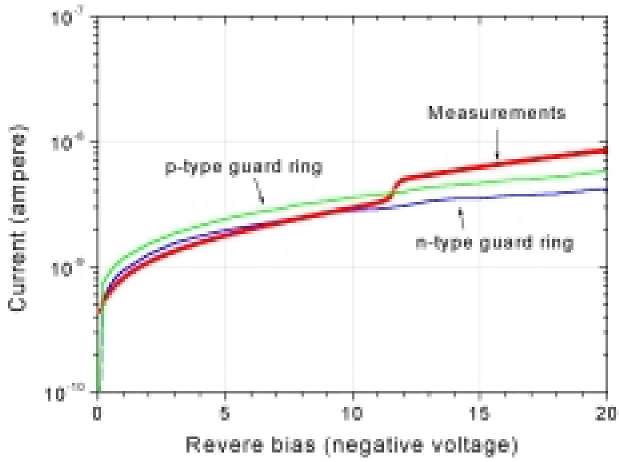

Figure 2: Dark reverse leakage current of the p-i-n photodiode with respect to the applied reverse bias voltage. In our design, an n-type guard ring was considered. The simulation results for both of the n-type and p-type guard-rings are also included for the comparison.

Due to the strong absorption of light in p-layer, quantum efficiency decreases with increasing p-layer thickness. On the other hand, barrier height of thin p-layer photodiode reduces and as a result the leakage current increases. Thus the trade-off between quantum efficiency and leakage current is considered. In this paper, two kinds of photodiodes with different p-layer thickness of about 3μm and 1.7μm respectively are fabricated and their leakage current is compared in Figure 3. Figure 3 shows that leakage current of thinner p-layer(1.7 μm) photodiode is higher that of 3μm photodiode. The reason is due to the reduction of barrier height that holes generated in p-layer by incident light must cross.

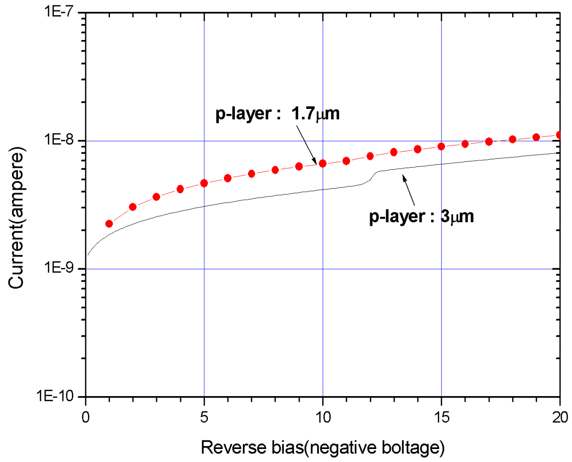

Figure 3: Dark reverse leakage current of the p-i-n photodiode as function of the applied reverse bias voltage under different p-layer thickness

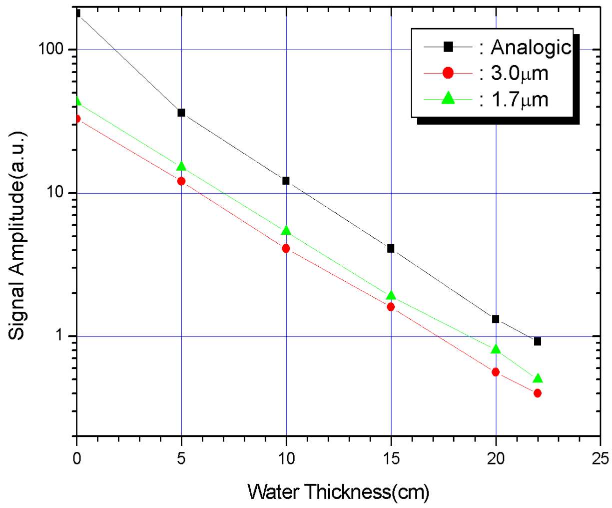

Figure 4: Signal output of the CT detector ($CdWO_4$ + p-i-n photodiode) when irradiating x-rays for various water depths. For the comparison, the commercial Analogic CT detector was also evaluated at the same condition.

Figure 4 shows the measurement of signal output under the irradiation of x-rays when the photodiode is coupled with the $CdWO_4$ scintillator for the CT detector. In the measurement, water phantom having various thicknesses was considered in order to simulate the detector response for the attenuated x-rays. According to the measurements as shown in Fig. 4, our design of the CT detector presents a typical exponential response. However, the two absolute magnitudes of signal output are poorer compared with the commercial Analogic CT detector. Detector using the photodiode with 1.7μm p-layer shows high photosensitivity in Figure 4. As already described, the photodiode with about 3 μm-thick p-layer largely attenuates the light quanta emitted from the scintillator, resulting in lower signal output. Based on the results of this paper, p-layer thickness must be thin and the depletion layer must be sufficiently thick to a large fraction of the incident light to be absorbed in order to increase the quantum efficiency.

IV. Summary and Future Study

We have fabricated a photodiode with p-i-n structure by using standard microelectronic process for the CT detector. The performance of the leakage current is reasonable, order of nA at zero bias. In case of the use of guard-ring to suppress the leakage current, it was evaluated that an n-type was more efficient rather than a p-type. From the x-ray measurements with respect to variation of the water attenuator thickness, the exponential decease was observed. However, comparing with the commercial CT detector as a reference, the absolute magnitude is somewhat lower (about 2.5 times), which might be due to the relatively thick p-layer or the layer quality during the fabrication process.

For the better performance of the CT detector, we have a plan to make another designs of photodiodes with extremely thin p-layer. To achieve this, an ion-implantation process to define the p-layer is considering rather that the previous diffusion process because the ion-implantation gives thinner p-layer and good uniformity [3]. The other design is a Schottky diode [4]: direct deposition of the transparent conduction oxide (TCO) such as an indium-tin-oxide (ITO) onto the intrinsic layer instead of the p-layer, providing higher light quanta transfer to the interaction region. We expect that these designs of the photodiode work well for the application into the CT detector.

REFERENCES

[1] PA Glasow, B Conrad, K Killig, and W Lichtenberg, "Aspects of semiconductor current mode detectors for x-ray computed tomography," IEEE Trans Nucl Sci, 28 (1981) 563-571.

[2] DP Boyd, "Status of diagnostic x-ray CT: 1979," IEEE Trans Nucl Sci, 26 (1979) 2836-2839.

[3] S Wolf and RN Tauber, *Silicon Processing for The VLSI Era, Vol. 1, Process Technology*, Lattice Press (1986).

[4] SM Sze, *Physics of Semiconductor Devices*, 2nd ed., John Wiley & Sons (1981).