X-RAY DETECTOR BASED ON BULK MICROMACHINED PHOTODIODE

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Abstract:

This paper reports the design, fabrication, assembly and testing of a x-ray detector based on a bulk micromachined photodiode (BMMPD) with a cavity filled with a scintillating crystal. The x-ray photons that reach the detector are first converted to visible light by the scintillating crystal. The visible light is then detected by the BMMPD, producing an electric current whose value is proportional to the incident x-ray intensity. The tests are done using a x-ray tube powered with a voltage of 35 kV, and a current ranging from 0 mA to 1 mA. With this setup, very promising results were obtained.

Keywords: *X-rays, BMMPD, Scintillator, Radiography, Micromachining*

I. INTRODUCTION

Several medical imaging methods, such as computed tomography, ultrasound and magnetic resonance are digital, while conventional x-ray imaging remains an analog technique [1]. X-ray imaging techniques usually have very strict exposure requirements due to the narrow brightness depth of the traditional radiographic silver films. They also offer very few possibilities of image processing. On the other hand, digital radiography systems offer the possibility of imaging with a wider range of exposure requirements and provide an image that may be processed and displayed in a variety of ways.

The main advantages of digital radiographic systems may be divided into three classes:

- Dose reduction,
- image processing and display in real time, and
- flexibility in image storage and retrieval.

The first advantage of digital radiography is the possibility of dose reduction. In conventional radiology, the dose is determined by the sensitivity of the image receptor and the film brightness depth. In digital radiology, both these constrains can be relaxed. Dose reduction can be achieved by adjusting the dose to give the required signal to noise ratio in the image. Further reductions are possible by using the x-ray spectrum that gives the lowest dose for a given signal to noise ratio and by recovering any losses in contrast using digital techniques.

The second advantage of digital radiography is the possibility of changing the characteristics of the image during the medical evaluation. The way of mapping the image into levels of brightness on a screen can be completely controlled by the user.

The third advantage of digital radiology is the possibility of image storing in a computer database and or its fast transmission to long distances.

Having in mind the advantages of digital radiography relatively to the traditional one, which uses silver films, the main purpose of the present work is to validate a simple and efficient solution for a x-ray detector using the electronic integration and micromachining technologies.

II. BACKGROUND

One of the first x-ray sensors developed was based on a silicon charge coupled device (CCD). The silicon has a low x-ray absorption coefficient, but for each 1 keV of x-ray photons absorbed, about 277 electrons are excited. This enables the construction of x-rays sensors with better sensibility than the traditional radiographic silver films. However, the small number of detected photons in CCD results in a significant quantum noise. In order to reduce the quantum noise, the radiation dose can be increased or the quantum efficiency of the sensor can be improved. The increase in the x-ray dose is obviously not desired for medical applications. The quantum efficiency of the sensor can be increased by adding a scintillating layer above the CCD. Since the x-rays are first absorbed by the scintillating layer, which has a high absorption coefficient, and then converted into visible (or near visible) light, the quantum efficiency of the detector is improved. A drawback of this approach is that the spatial resolution of the device is approximately equal to the thickness of the scintillator layer (fig. 1(a)). An increase of the thickness of

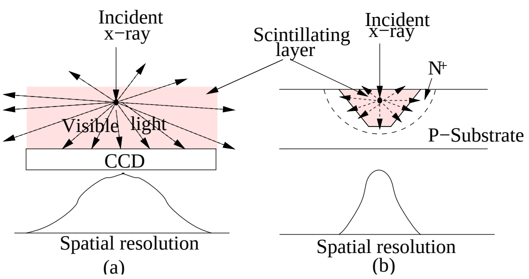

Fig. 1: Spatial resolution of: (a) CCD with scintillating layer. (b) BMMPD with scintillator.

the scintillating layer without decreasing the spatial resolution can be achieved by using BMMPDs, as it is shown in fig. 1(b). In this case, the light produced by the scintillator is confined to the corresponding BMMPD cavity.

III. DESIGN

In medical imaging diagnosis, the x-rays are produced with voltages from 25 kV to 120 kV, approximately. These voltages produce an intensity peak ranging from 10 keV to 100 keV. A standard silicon wafer (525 μm thick) only absorbs about 2.2% of the 100 keV x-rays energy. A 16.2 mm thick layer absorbs 50% of the same radiation, so in a practical application, each photodiode junction needs to be at least 20 mm thick which is completely impractical [2]. Therefore, as it was pointed out in the previous section, a x-ray scintillation layer which converts x-rays into visible light is a good approach for increasing the quantum efficiency of the silicon detectors.

In the case of image acquisition by an array of BMMPDs filled with scintillating crystals, the light produced by each scintillator is isolated from its neighbors, reducing losses and cross-talk between adjacent detectors. Moreover, introducing a reflective layer above the scintillator (in the x-rays path) confines the visible light inside the cavity of the BMMPD, increasing the efficiency.

Therefore, the device consists in a P-type silicon substrate, where a cavity is made. The inner walls of the cavity are doped to form a N^+ region and the cavity is then filled with a scintillating crystal. Finally, a film of a reflective material is placed above the scintillating crystal, as it is shown in fig. 2.

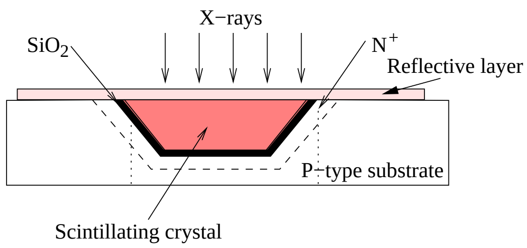

Fig. 2: Diagram of the x-ray sensor based on a BMMPD.

Scintillating crystal

This application requires a scintillating crystal with high light yield and reasonably fast decay time [3]. CsI:Tl satisfies both conditions. It has an emission wavelength of 560 nm which combined with silicon devices has one of the highest quantum efficiency of all used materials. Table 1 shows the most important properties of CsI:Tl as a scintillating material.

Fig. 3 shows the x-ray absorption by a 400 μm thick layer of CsI:Tl (400 μm is the depth of the BMMPD cavity), in the range from 10 keV to 100 keV. The calculations are based on the mass energy-transfer and mass energy-absorption coefficients of the materials [4]. In this figure can be also observed the effect of the k edges of Cesium and Iodine at 35.9846 keV and 33.1694 keV, respectively.

Table 1: Properties of CsI:Tl as a scintillating material at room temperature.

Density (g/cm^3)	4.51
Effective Atomic Number	54
Light yield $(phot/MeV)$	65900
Emission wavelength (nm)	560
Decay time (ns)	10^3

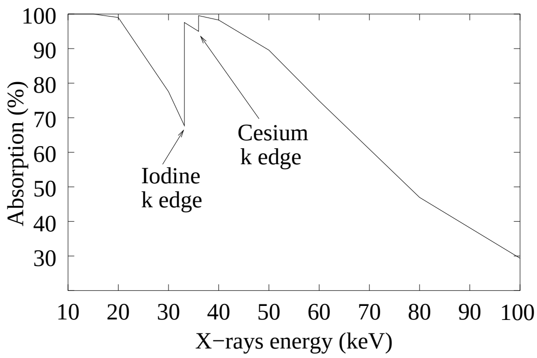

Fig. 3: Absorption of a CsI:Tl crystal $400\ \mu m$ thick.

Photodiode

For silicon, the penetration depth of light with $\lambda = 560\ nm$ is $\approx 0.75\ \mu m$ [5]. To maximize the efficiency of the diode by minimizing recombination, the junction depth needs to be as shallow as possible. Simulations have been carried out using the device simulator SUPREM to verify the calculated junction depth of $0.1\ \mu m$.

IV. FABRICATION

Photodiode

For the fabrication of the BMMPD chip a p-type silicon wafer is used. Two boron implantations onto the front ($1 \times 10^{14}\ cm^{-2}$, $30\ keV$) and back ($5 \times 10^{15}\ cm^{-2}$, $50\ keV$) respectively are implemented to assist the ohmic contact to the metal in later stages. The wafer is then covered with an oxide and nitride layer. Patterning photoresist and dry etching the nitride and oxide layers leaves the mask for etching the cavity. The cavity is etched into the wafer with KOH. The nitride and oxide films are then stripped from the wafer. After the creation of the cavity using KOH, Arsenic is implanted ($1 \times 10^{16}\ cm^{-2}$, $50\ keV$) through a $40\ nm$ thick oxide layer to form the pn-diode. This oxide is then stripped off and a new $80\ nm$ thick oxide layer is grown thermally to insulate the metal tracks for the diode connection and to act as an anti reflection coating. The oxide is patterned with buffered hydrofluoric acid to create contact windows to the silicon. Layers of titanium and aluminum/silicon are then sputtered on the front and back of the wafer and patterned on the front the metal tracks. A picture of the chip is shown in figure 4. Three extra diodes were included in the design for test purposes.

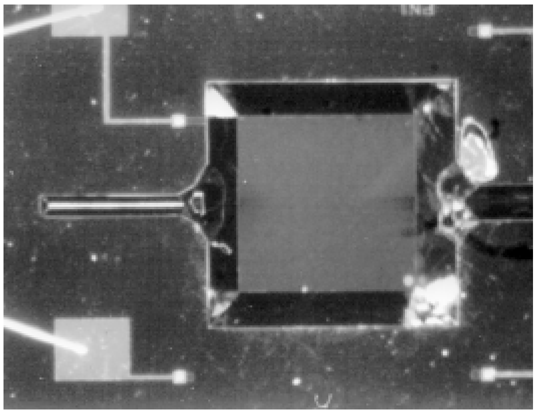

Fig. 4: Diode detector. Cavity size: 2 mm x 2 mm.

Scintillating crystal

The scintillating crystal (CsI:Tl) was placed inside the cavities by using a clamping pressure of about 10 MPa. In order to achieve a perfect fill it is necessary to remove first the air inside the cavity. So, this fabrication step must be processed inside a vacuum chamber. The CsI:Tl crystal was produced by Molecular Technology GmbH, Berlin, Germany. Fig. 5 shows a picture of the chip after this step.

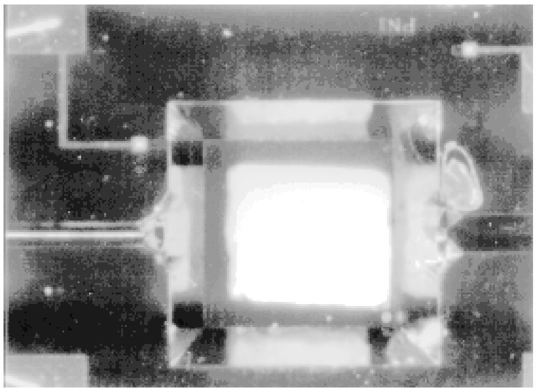

Fig. 5: Diode detector filled with CsI:Tl.

Reflective layer

As a final step, a film of reflective ink was deposited above the scintillating crystal. This film was deposited using a pressurized spray through a deposition mask.

V. TEST PROCEDURES AND RESULTS

I-V characteristic curves for the diodes have been obtained using a Hewlett-Packard 4155A four-probe analyzer. The characteristics of the three test diodes were measured to provide valuable information in the evaluation of the BMMPD characteristics and performance. First, the characteristic of the BMMPD ($2\ mm \times 2\ mm$) was compared to the test diodes characteristics. The results in fig. 6 show a softer breakdown knee for the BMMPD. However, it has low reverse leakage and can be readily used for light detection.

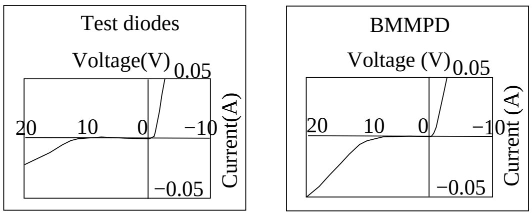

Fig. 6: Dark current I-V characteristics for BMMPD and test diode 1.

The spectral response of the BMMPD was measured using an Oriel spectral analyzer system motorized monochromator UV VIS. The output current was measured using a Keithley 487 picoamperimeter/ voltage source. No bias was applied to the photodiode. Fig. 7 shows the measured current for each wavelength.

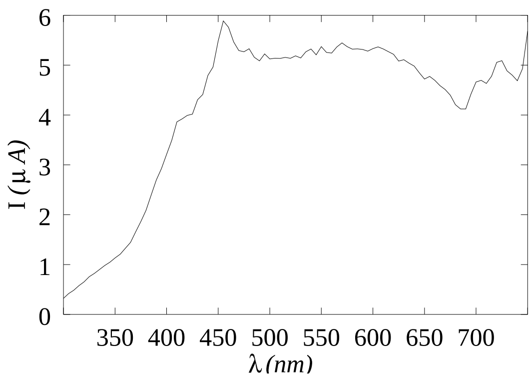

Fig. 7: Spectral response of the BMMPD.

The x-rays tests were made using an x-ray tube (Leybold) with molybdenum anode, which characteristic short wave radiation is $K_\alpha = 17.4\ keV$ and $K_\beta = 19.6\ keV$. The tube was powered with a voltage of $35\ kV$, and a current ranging from $0\ mA$ to $1\ mA$. Fig. 8 shows the measured values.

As a curiosity, the x-rays tubes for dental radiography, using traditional films, are powered with a

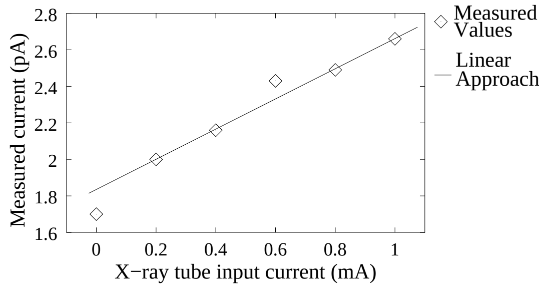

Fig. 8: Experimental results of the detector with a x-ray tube input voltage of 35 kV.

voltage near $70\ kV$ and a current near $10\ mA$, i. e. a power about 20 times higher than the one used in our experiments.

VI. CONCLUSION

This approach, BMMPDs with scintillating crystals, reveals to be suitable to make x-rays detectors. The photodiode cavity allows to increase the scintillator thickness without decreasing the spatial resolution of several detectors, when placed together. So, for x-rays imaging, it can improve the performance of a CCD with a scintillating layer on the top. As a future work, several photodiodes will be placed side by side in the same chip.

VII. ACKNOWLEDGMENTS

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