CMOS X-Ray Imager for Dental Radiography

N. F. Ramos[1], J. G. Rocha[2], S. Lanceros-Mendez[3],
R. F. Wolffenbuttel[4] and J. H. Correia[2]

[1]University of Minho, Dept. Ind. Electronics, Campus de Azurem, 4800-058 Guimaraes, Portugal.
Tel: +351 253 510190 Fax: +351 253 510189 Email:Email: nramos@dei.uminho.pt http://www.dei.uminho.pt
[2]University of Minho, Dept. Ind. Electronics, Campus de Azurem, 4800-058 Guimaraes, Portugal.
[3]University of Minho, Dept. Physics, Campus de Gualtar, 4710-057 Braga, Portugal.
[4]Delft University of Technology, Lab. Electr. Instr., Delft, The Netherlands

Summary: *This paper describes a 64 pixels X-ray imager architecture and fabrication process. The imager is composed by a 8×8 photodiodes array, fabricated in CMOS process, and an array of wells £lled with scintillating crystals. A thick-£lm of aluminum is etched in order to achieve square wells with 500 μm depth. The wells are £lled with CsI:Tl scintillating crystals and placed above the photodiodes. The scintillating crystals convert the X-ray energy into visible light, which is guided into the photodiodes by the re¤ective aluminum walls, avoiding crosstalk between adjacent detectors. Usually, the spatial resolution of the scintillating x-rays detectors is identical to the scintillators thickness. By using the light guides, the scintillator thickness can be adjusted in order to achieve optimal absorption ef£ciency, since the spatial resolution is established by the pixel size (200 μm side).*

Keywords: X-ray, Digital Radiology, Scintillator, Etching
Category: 4 (Non-magnetic physical devices)

1 Introduction

During the last sixty years, the concept of medical imaging has been associated to X-ray imaging systems based on silver £lms, which perform image acquisition and provide physical support for image storage and display. These systems usually demand very strict exposure requirements, due to the narrow brightness depth of the traditional radiographic silver £lms. They also offer very few possibilities of image processing [1].

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

- Reduction of the radiation dose,
- Less time from image acquisition to image display,
- Possibility of image manipulation using digital image processing techniques, and
- Remote storage and retrieval.

The £rst advantage of digital radiology is the possibility of dose reduction. In conventional radiology, the dose is determined by the sensitivity of the image receptor and the £lm 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.

The second advantage is very important in medical emergency situations. The digital X-ray systems can provide images in a few seconds while developing a silver £lm takes several minutes.

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

The fourth advantage of digital radiology is the possibility of image storing in a computer database and/or transmission of the images to long distances.

The X-ray imaging systems for dental medicine must £ll some particular requirements:

- It must be inexpensive, easy to use and easy to replace, since most of the dental medicine technicians do not have a good knowledge on radiology.
- Reduction of the radiation dose is very important since the energy and intensity required to cross a tooth and produce an image are signi£cantly high.
- The sensor active area can be small (about $15\ mm \times 20\ mm$), which allows the use of standard fabrication processes.

Due to the high number of dental medicine facilities, the development of X-ray imagers for dental radiography is very interesting for the market.

2 System design

In dental medicine imaging, the X-rays are produced with voltages from 50 kV to 70 kV. These voltages produce an intensity peak ranging from 40 keV to 60 keV, approximately. A standard silicon wafer

(525 μm thick) only absorbs about 3.38% [1] of the 60 keV X-rays energy, not being suitable for the making of X-rays sensors.

Therefore, a x-ray scintillation layer is necessary to convert X-rays into visible light, which is then converted to an electric signal by means of an array of photodiodes.

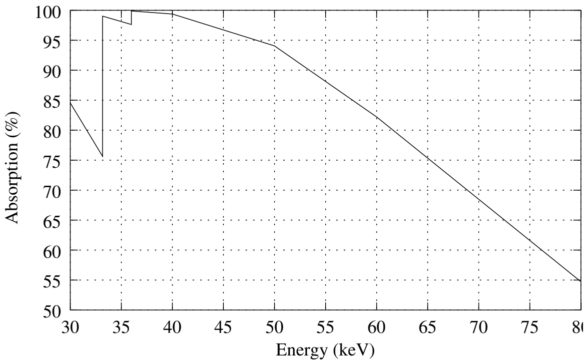


Fig. 1: Absorption of the 500 μm scintillator (CsI:Tl).

Figure 1 shows the percentage of X-rays absorbed by a 500μm thick CsI:Tl layer (simulated results). From 40 keV to 60 keV the scintillator absorbs more than 80% of the X-rays.

Each scintillator is isolated from its neighbors by aluminum, which allows multiple re¤ections and guides all produced light to the photodiode.

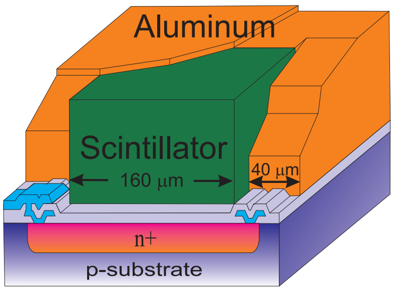


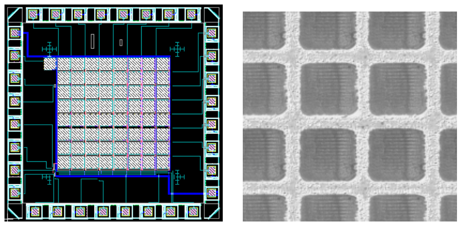
Fig. 2: Structure of each pixel.

Moreover, introducing a re¤ective layer above the scintillator (in the X-rays path) con£nes the light inside the scintillator for increasing the ef£ciency.

Therefore, the device consists of many micromachined aluminum cavities, where the scintillator is deposited. This structure is placed above the silicon die, which contains the photodetectors and readout electronics. Figure 2 shows the structure of one pixel.

3 Fabrication

The photodiodes are fabricated using a standard CMOS n-well 1.6 μm process (Fig. 3 (a)). The chosen photodiode structure was the n+ substrate due to its highest quantum ef£ciency and to its spectral response in the green region of the spectrum. This spectral response is appropriate because, the scintillating crystal used in this work yields green light.



(a) (b)

Fig. 3: (a) CMOS photodiodes array. (b) Photo of the aluminum cavities £lled with the scintillator.

The cavities opened on the thick-£lm of aluminum were made using an excimer KrF laser MINex (Lambda Physik) working in pulsed rating at 248 nm with pulse energy of 30 mJ and peak power of 2 mW, optical mirror and lenses for focusing of the laser beam, translation table with the sample and controlling system. The translation table has an absolute accuracy of 3.6 μm and repeatability of 0.9 μm [1].

The LASER was focused on the surface of the aluminum thick-£lm, using a plano-convex lens with a focal distance of 250 mm. With this setup, the diameter of the LASER cannot be smaller than 71 μm, due to the scattering of the LASER photons. However, this is acceptable for making 160 $\mu m \times 160$ μm cavities.

The cavities are £lled with the scintillator by a clamping pressure machine, in a low-temperature vacuum chamber. Figure 3(b) shows the aluminum cavities £lled with the CsI:Tl scintillator.

4 Conclusions

This approach, scintillators encapsulated in aluminum plus CMOS photodiodes, reveals to be suitable to dental radiography imagers. The aluminum encapsulation allows to increase the scintillator thickness with low cross-talk and good spatial resolution. The re¤ectivity of the aluminum inner-walls of the cavities are poorly affected by the LAT etching. CMOS technology allows to implement x-ray detectors which reveals advantages relatively to the CCD technology. As a future work the size of the cavities will be reduced, in order to increase spatial resolution.

References

[1] J. G. Rocha, N. F. Ramos, et. al., “X-rays Microdetectors Based on an Array of Scintillators: A Maskless Process Using LASER Ablation,” Proc. of IEEE Sensors Conference, 10-12 June 2002, Orlando, Florida, USA.