SELECTION OF DIMENSIONS AND GEOMETRY OF A STRUCTURED SCINTILLATOR FOR X-RAY IMAGING

Anna Sahlholm[*], **Olof Svenonius**[*], **Carina Näsgårde**[*], **Per Wiklund**[*], **and Jan Linnros**[*†]

[*]*Scint-X AB, Electrum 229, 164 40 Kista, Sweden, anna.sahlholm@scint-x.com*
[†]*Department of Microelectronics & Applied Physics, Royal Institute of Technology, Electrum 229, 164 40 Kista, Sweden, linnros@kth.se*

ABSTRACT

Scintillators are commonly used for converting x-rays into visible photons in digital x-ray imaging. We have fabricated a structured scintillator in a novel way, by filling a deep pore array in silicon with a scintillating material. Change of the pore pattern to a hexagonal geometry has significantly reduced optical interactions with the sensor such as moiré pattern, and thereby eliminated the need to match every pore with a specific pixel. The structured scintillator yields superior performance in terms of image resolution and contrast. An application for high resolution x-ray diagnostics under microgravity conditions is currently being developed in cooperation with Swedish Space Corporation and Technische Universität Dresden within the ESA GSTP study "X-ray Diagnostics for Space".

Key words: scintillator; x-ray; high resolution.

1. INTRODUCTION

Scintillators are commonly used for converting x-rays into visible photons in x-ray imaging detectors where a silicon based sensor is used. However, the scintillator layer degrades the lateral resolution since secondary photons are emitted isotropically. Thus, a trade-off usually exists between image resolution and detector sensitivity, and the thickness of the scintillator layer has to be optimized for each application.

Structured scintillators [3], on the other hand, such as CsI needles grown by a special chemical vapour deposition technique [5], may function as light guides, ideally bringing the generated secondary photons down to a single detector pixel. The light guiding performance of the needles is although not perfect, which results in 'crosstalk' between pixels, limiting their use in high-resolution imaging.

We have fabricated a structured scintillator in a different way, by filling a deep pore array in silicon with a scintillator, more specifically thallium doped caesium iodide, CsI(Tl). The pore array is fabricated using standard silicon technology, i.e. by lithography and deep reactive ion etching (DRIE), while filling of the pores is accomplished by melting CsI(Tl) powder into the silicon pore array, working as a mould. Prior to the melting process, the pore walls are covered by an oxide layer, yielding a lower refractive index than the CsI (n ∼ 1.79) to provide light guiding down to an image sensor chip in close proximity, see Figure 1. The detector principle has previously been demonstrated for $30 - 50\ \mu$m pore size [1, 2, 4, 7] as well as for high-resolution material characterization at $\sim 4\ \mu$m pore size [6]. Applications include dental imaging, materials analysis and inspection, high-resolution medical imaging, x-ray diffraction etc.
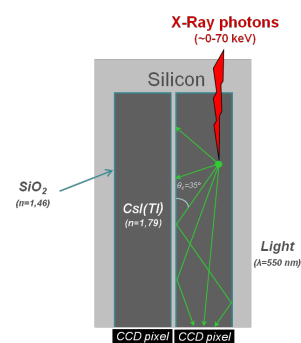

Figure 1. Schematic of secondary photon lightguiding inside a CsI(Tl) filled pore towards a detector pixel following the absorption of an x-ray quanta.

The structured scintillator yields superior performance in terms of image resolution and contrast. Changing the pattern from a square structure to a hexagonal close packing structure, significantly reduces optical interactions such as moiré pattern, and thereby eliminates the need to align the scintillator to the sensor chip. Results of measurements on the modulation transfer function (MTF) and x-ray image resolution for scintillator samples of different pitch and different patterns will be presented.

An application for high resolution x-ray diagnostics under microgravity conditions is currently being developed in cooperation with Swedish Space Corporation and

Technische Universität Dresden within the ESA GSTP study "x-ray Diagnostics for Space". The evaluations of this imaging technique show that real-time visualization of objects in the range of less than 5 μm is attainable.

2. EXPERIMENTAL

The structured scintillator plate was fabricated from standard 4-inch silicon wafers using lithography and deep reactive ion etching (DRIE), using a standard silicon ICP (inductively coupled plasma) machine to create a pore array in silicon. The pore pitch was 20 - 40 μm, wall thickness was $\sim 3\,\mu$m and etch depths were in the range of 200 μm. Figure 2 (top) shows cross sectional SEM (scanning electron microscopy) images of a representative sample. The wafers were subsequently thermally oxidized to create a 'cladding' layer of lower refractive index to provide light-guiding down to the CCD pixels. Wafers were then filled with CsI(Tl) by melting in a furnace at $\sim 620°$C with the silicon pore array acting as a mould, see Figure 2 (bottom). The excessive CsI(Tl) was finally removed by polishing and samples of ~ 1cm^2 up to ~ 9cm^2 size were produced by dicing.

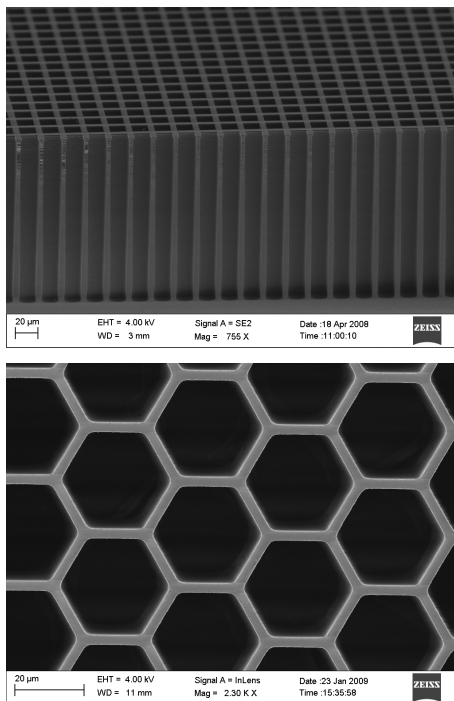*Figure 2. SEM cross sectional imaging. (top) Silicon pore array after DRIE etching with 20 μm pitch and 200 μm depth. (bottom) Hexagonal pore pattern.*

The scintillating plates were first characterized using a standard dental x-ray source (Planmeca Intra) operating at 70 kVp (~ 35keV mean energy). The standard exposure time used was 125 ms, corresponding to a dose of ~ 75mR. Imaging was performed using CCD-detectors from Hamamatsu (S8981), having either a 1.5 or 3.0 mm thick fibre optic plate (FOP), where the standard scintillator had been removed. The CCD pitch was 20 μm in a 1000 x 1500 pixels square array. All MTF curves were measured using this setup.

The scintillator for Swedish Space Corporation was then characterized at their facility, using a 90 kV 90 μA microfocus x-ray source with a spot size of $\sim 5 - 7$m. This molybdenum source benefits from a thinner silicon substrate, thus this parameter was changed compared to our previous scintillators. Imaging was performed using a Vosskühler CCD-11000 detector, having a 50 mm thick FOP where the standard scintillator had been removed. The CCD (charged coupled device) pitch was 9 μm binned to an effective pitch of 18 μm in a 4022 x 2680 pixels square array.

2.1. Results and discussion

Due to previous results, where we have seen moiré patterns, we have changed the geometry of the scintillator pores. This reduces optical interactions between the scintillator matrix and the pixel structure of the sensor, and thereby further eliminates the need to exactly align the scintillator with the sensor. This without affecting the spatial resolution in the image. The results are shown as comparisons between a commercial scintillator with CsI needles, and the same silicon sensor used together with Scint-X structured scintillators.

Figure 3 shows the imaging performance in terms of the modulation transfer function (MTF) for samples of different pitch. The graphs were calculated from images of a sharp lead edge placed at a small angle on top of the sensor. As expected, smaller pitches result in superior imaging performance, while the 40 μm pitch still holds sufficient contrast at 10 linepairs per mm for applications in dental imaging. Indeed, the MTF characteristics are far superior to CsI(Tl) needles as offered by several vendors and approach those of direct conversion silicon detectors although at a much higher detection sensitivity [8].
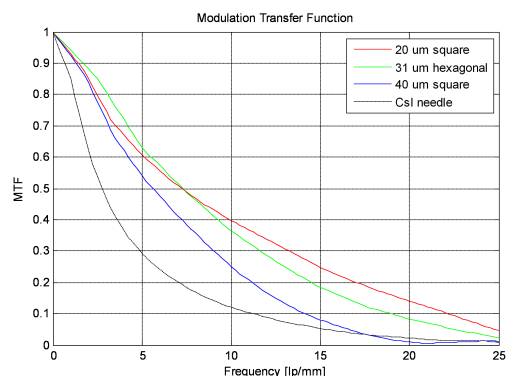

Figure 3. Modulation transfer function measured for three different pore types compared to a CsI needle matrix.

In Figure 4 we demonstrate images of a line pair grating using first a commercial detector using a CsI needle scintillator (top) and then a structured scintillator plate having a 20 μm pitch (bottom). Looking at the images one observes the superior imaging quality of the bottom image where the line pairs are significantly sharper (note that brightness and contrast have been adjusted to provide similar image appearance).
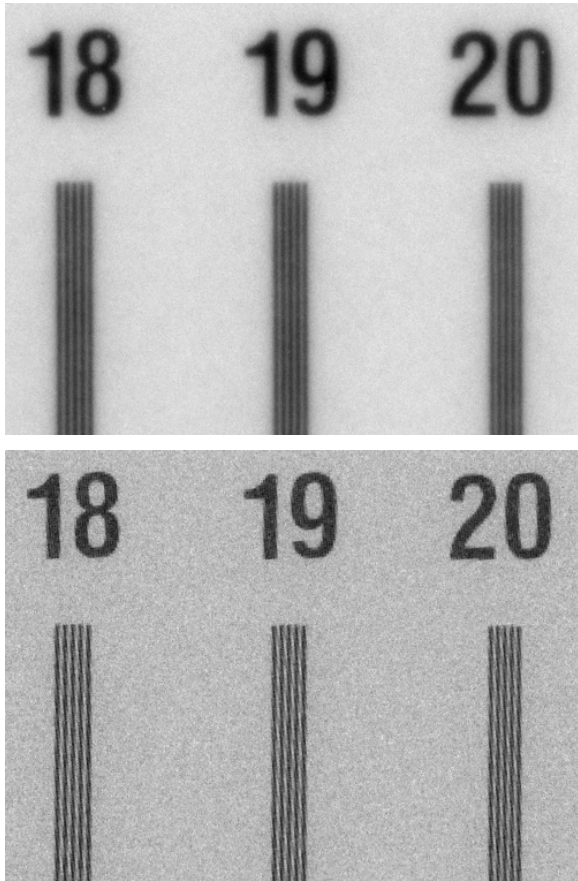

Figure 4. Comparison of image quality obtained by different detector configurations. Top: A commercial detector. Bottom: A sample with 20 μm square pitch.

3. CONCLUSIONS

We conclude that the structured scintillator yields superior performance in terms of imaging quality as evidenced by the MTF characteristics and x-ray images. The light output is, however, weaker than for non-structured scintillators. This is because light guiding only accepts a certain cone of the emitted light, resulting in a smaller fraction of detected light in each sensor pixel.

A hexagonal pattern with dimensions optimized for each sensor and x-ray source gives the possibility to use the structured scintillator in a wide range of applications. This technique is promising for all types of x-ray imaging where resolution is an important parameter.

REFERENCES

[1] Badel, X., Linnros, J., Kleimann, P., et al. 2004, IEEE Trans. Nucl. Sci., 51, 1001

[2] Badel, X., Norlin, B., Kleimann, P., et al. 2006, IEEE Trans. Nucl. Sci., 53

[3] Deckman, H. W., Dunsmuir, J. H., & Gruner, S. M. 1989, J. Vac. Sci. Technol. B, 7, 1832

[4] Kleimann, P., Linnros, J., Fröjdh, C., & Petersson, C. 2000, IEEE Trans. Nucl. Sci., 47, 1483

[5] Nagarkar, V., Gupta, T., Miller, S., et al. 1998, IEEE Trans. Nucl. Sci., 45, 492

[6] Olsen, U., Badel, X., Linnros, J., et al. 2006, Nucl. Instrum. Methods A, 576, 52, iWORID, 2006

[7] Simon, M., Engel, K.-J., Menser, B., Badel, X., & Linnros, J. 2008, Medical Physics, 35, 968

[8] Spartiotis, K., Pyyhtiä, J., Schulman, T., Puhakka, K., & Muukkonen, K. 2003, Nucl. Instrum. Methods A, 501, 594