Microdome-grooved Gd$_2$O$_2$S:Tb scintillator for flexible and high resolution digital radiography

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Abstract: A flexible microdome-grooved Gd$_2$O$_2$S:Tb scintillator is simulated, fabricated, and characterized for digital radiography applications. According to Monte Carlo simulation results, the dome-grooved structure has a high spatial resolution, which is verified by X-ray image performance of the scintillator. The proposed scintillator has lower X-ray sensitivity than a nonstructured scintillator but almost two times higher spatial resolution at high spatial frequency. Through evaluation of the X-ray performance of the fabricated scintillators, we confirm that the microdome-grooved scintillator can be applied to next-generation flexible digital radiography systems requiring high spatial resolution.

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References and links


1. Introduction

Since their discovery by W. C. Röntgen in 1895, X-rays have had a major impact on numerous fields including medicine, science, and engineering. X-ray imaging is the most important and widely-used method for noninvasive medical diagnosis in the hospital environment. Digital radiography (DR) converts X-ray images to electronic data that can be displayed on a monitor. DR is the latest advancement in X-ray imaging technology, and is now used in medical diagnostic applications including dental, chest, and C-arm X-ray systems, as well as mammography. Figure 1(a) shows a schematic view of a DR system with an indirect conversion method using a scintillator and a photodiode array. The scintillator converts the incident X-ray into visible light, and the photodiode array converts the visible light into electric charge.

As shown in Fig. 1(b), the converted visible light in the conventional scintillator scatters in all directions, thereby degrading spatial resolution. Several methods have been developed to suppress spreading of the converted visible light, including columnar-structured cesium iodide (CsI:Tl) [1], pixel-structured CsI:Tl [2,3], and pixel-structured gadolinium oxysulfide (Gd₂O₂S:Tb) [4–7] (hereafter, Gd₂O₂S:Tb and CsI:Tl are referred to as GOS and CsI, respectively). Silicon-based U-grooved structures with thin walls have been produced for pixel-structured scintillators using microelectromechanical system (MEMS) fabrication technology [2,3]. However, silicon and CsI are too brittle to be used for flexible X-ray detectors. Flexible x-ray image systems are very attractive due to structural robustness and extensive applicability. Since the first report of flexible x-ray image sensors by R.A. Street et al., who developed x-ray image sensors with flexibility based on thin film transistor (TFT) and organic semiconductor materials on a polyethylene naphthalate (PEN) substrate [8,9], researchers have focused on the realization of a flexible scintillator with high x-ray sensitivity and spatial resolution. J.C. Blakesley et al. reported modeling results for an organic X-ray imager [10] and P.E. Keivanidis et al. presented X-ray stability and the response of an organic semiconductor material in a photodiode application [11].

The flexibility of scintillators is very important for next-generation flexible DR systems. For dental applications, a flexible digital intraoral X-ray detector could reduce patient
inconvenience caused by the rigidity of conventional detectors. In contrast to CsI, GOS-based scintillators are very flexible, because GOS is generally mixed with organic materials. Polymer-based U-grooved scintillators have also been developed in order to exploit the flexibility of polymers [7]. Although U-grooved polymer structures are a good prototype for flexible scintillators, they have two crucial drawbacks: low fill factor and a difficult fabrication process. While low wall thickness helps increase the fill factor, it is very difficult to realize using polymer microreplication technologies such as injection molding, hot embossing, and casting. B. Wowk et al. reported that a pyramidal-structured scintillator produced higher light output than either a U-grooved scintillator or a nonstructured plat scintillator [4]. G. Hull et al. meanwhile verified that a conical-shape scintillator had higher light collection efficiency than a cylindrically-shape scintillator [12]. However, fabrication of array-type micro-pyramidal and conical structures is more difficult than fabrication of U-groove structures.

In this study, we propose a new flexible scintillator with a microdome-groove array (as delineated in Table 1) that offers easy microfabrication, a high fill factor, and high spatial resolution.

2. Simulation

To compare the performance of dome-grooved and nonstructured scintillators, the light collection efficiency (LCE) and relative point spread function (PSF) were calculated using DETECT2000 [12,13], a Monte Carlo-based simulation program, and two different types of scintillators were fabricated and characterized. Furthermore, a commercial scintillator, the Min-R2000 (Carestream Health Inc., USA), was characterized for relative comparison. The configurations of the dome-grooved and nonstructured scintillators and their simulation parameters are shown in Table 1. The total thickness of the scintillation material of the dome-grooved scintillator is 60 µm, which is the sum of the groove height (40 µm) and over-layer
Table 1. Configurations and design parameters of dome-grooved and nonstructured scintillators.

<table>
<thead>
<tr>
<th>Structural type of scintillator</th>
<th>Parameter descriptions for simulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Microdome-grooved</td>
<td>$T_{GS}$ (40 $\mu$m): Thickness of dome-grooved structure</td>
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<tr>
<td></td>
<td>$T_{OL}$ (20 $\mu$m): Thickness of over-layered structure</td>
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<tr>
<td></td>
<td>$W_{GS}$ (40 $\mu$m): Width of groove-structure</td>
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<tr>
<td></td>
<td>*Structural pitch of dome-grooved scintillator = $W_{GS}$</td>
</tr>
<tr>
<td>Nonstructured</td>
<td>$T_{NS}$ (60 $\mu$m): Thickness of nonstructured scintillator</td>
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thickness (20 $\mu$m). The thickness of the nonstructured scintillator is the same as that of the dome-grooved scintillator, 60 $\mu$m.

In the simulation, as shown in Figs. 2(a) and 2(b), we assumed that light is generated at 6 different positions (height = 5, 15, 25, 35, 45, and 55 $\mu$m) and the generated light spreads in all directions. Furthermore, we assumed that one million photons are generated at each fixed light generating position. As a ‘METAL’ surface condition for the simulation using

![Fig. 2. Simulation conditions and results of microdome-grooved and nonstructured scintillators.](image-url)

(a) Configuration for simulation of dome-grooved scintillator. (b) Configuration for simulation of nonstructured scintillator. (c) LCE (solid) and LCE in pixel (open) of scintillators according to light generation height. (d) Relative PSF of scintillators according to distance.
DETECT2000, a reflector with 90% reflectance and non-transmission located between the scintillator and the substrate was adopted in the simulation. The absorption and scattering mean free path of GOS were assumed to be 10 mm [14] and 0.04 mm, respectively. Figures 2(c) and (d) show the simulated LCE and PSF results, respectively. Figure 2(c) shows the LCE and the LCE in a pixel according to the light generation height. The LCE and the LCE in a pixel are defined as the rate of collected photons over the total photons generated at a fixed height in the scintillator at a large detection area of 10 × 10 mm$^2$ and the scintillator pixel area of 40 × 40 μm$^2$, respectively. Figure 2(c) shows that the nonstructured scintillator has a better LCE than the dome-grooved scintillator; conversely, the dome-grooved scintillator has a better LCE in a pixel than the nonstructured scintillator. This clearly indicates that the pixelized dome-grooved structures are not helpful for the LCE but are helpful for the LCE in a pixel. The low LCE of the dome-grooved scintillator is caused by optical trapping at the dome-shape sidewall. On the other hand, the high LCE in a pixel is due to suppression of excessive light spreading by the dome shape. Figure 2(d) shows the relative PSF according to the distance from the center of the structure where the photon was generated. Relative PSFs were calculated and displayed at two light generation heights, 5 and 55 μm, as shown in Figs. 2(a) and 2(b). For the light generation height of 5 μm, there is no significant difference between the two scintillators in terms of PSF. However, for the light generation height of 55 μm, the PSF calculated from the dome-structured scintillator rapidly decreased with increasing distance compared to the nonstructured scintillator. This means that the generated light in the dome-structured scintillator is not spread out over a long distance, while that in the nonstructured scintillator is spread widely. The high LCE in a pixel and low PSF reflect high spatial resolution. Based on the results of our simulation, we suggest that the proposed dome-grooved structure enhances spatial resolution.

3. Microfabrication and X-ray characterization setup

The fabrication process of the microdome-grooved scintillator is illustrated in Fig. 3. Diffuser lithography was performed to acquire dome-shaped AZ9260 photoresist microstructures [15–17] and Ni electroforming was conducted to fabricate a Ni micromold. A dome-grooved polymer sheet was produced by a polymer microcasting technique using UV-curing resin (Nano Photonics Chemical, Korea). After coating an Ag reflective layer, solution-type GOS precursor mixed with 75 wt% GOS particles (Phosphor Technology, UK) and 25 wt% UV resin was poured onto the dome-grooved polyethylene terephthalate (PET) film. The fabrication process was then completed with solidification of the GOS precursor by UV exposure. The dome-shaped microstructure has a thick lattice at the bottom and a thin lattice.
at the top. Therefore, during the polymer microreplication process, due to the tapered structure, the microstructures were easily separated from the Ni mold. In addition to easy fabrication, the thin lattice at the top increased the fill factor. Figures 4(a) and (b) show an inclined scanning electron microscopy (SEM) micrograph of the replicated polymer microdome grooves and a cross-sectional SEM micrograph of the fabricated scintillator, respectively. Due to an imperfectly aligned cutting line, we could not capture the highest point of the groove. Figure 4(c) shows a prototype of the fabricated dome-grooved scintillator of 30 × 40 mm$^2$ in size, and Fig. 4(d) demonstrates its flexibility. The fabrication process of the nonstructured scintillator is simpler than that of the dome-grooved scintillator. A reflective layer of Ag was directly coated on the PET film. The subsequent fabrication processes were the same as those of the dome-grooved scintillator.

A CMOS photodiode pixel array (Rad-icon Imaging Corp., USA) was employed as a readout device for optical photons emitted from the scintillators. The CMOS photodiode array has a format of 512 × 1024 pixels with a pitch of 48 µm. The Nyquist limit for a device with a pixel pitch of 48 µm is about 10 mm$^{-1}$. The scintillators were directly overlaid onto the active area of the CMOS photodiode array. To minimize the air gap that could form between the bottom of the scintillator and the top surface of the CMOS photodiode array, a thin polyurethane foam layer was applied for compression between the scintillator and the CMOS photodiode array and was held in place by a 1 mm-thick graphite cover. During measurement, a 100 µm-thick film of PET was used to avoid contact damage between the scintillator and the bare photodiode pixel array. The readout time was fixed at 1.0 s. For the X-ray source, a 60 kVp spectrum from a fixed tungsten anode and a 125 µm-thick beryllium exit window...
(Oxford Instruments X-ray Technology, USA) were used. An additional aluminum filter with a thickness of 2.5 mm was used to remove the low energy part of the spectrum.

4. X-ray characteristics

Figure 5 compares the X-ray imaging performances of the fabricated samples and a commercial GOS-based scintillator (Min-R2000). Although the fabrication method and the thickness of the Min-R2000 were different as compared to the two fabricated samples, it nevertheless provides a good reference, because both the method and thickness are now widely used in medical applications. To evaluate imaging performance, the measured values of four image quality parameters are required; sensitivity, the modulation transfer function (MTF), the noise power spectrum (NPS), and the detective quantum efficiency (DQE). Sensitivity is defined as the collected charge per unit area per unit exposure to radiation. The MTF is expressed as a function of spatial frequency and describes spatial resolution. The NPS gives the noise as a function of spatial frequency. The DQE describes the ability of an imaging system to transfer a signal relative to noise from its input to its output, and is a frequency-dependent measure of the dose efficiency of the imaging system [18–20].

The DQE can be calculated using the detector output signal as follows [21]:

\[
DQE(f) = \frac{S^2 \cdot MTF^2(f)}{NPS(f) \times q_0}
\]

Fig. 5. X-ray imaging performance of the fabricated dome-grooved (square), nonstructured (circle), and Min-R2000 (triangle) scintillators in terms of sensitivity, MTF, NPS, and DQE. (a) X-ray sensitivity according to X-ray input dose. (b) MTF. (c) NPS, and (d) DQE according to spatial frequency at a representative X-ray input dose of 308.83 µGy.

where \( S \) denotes the signal output of the detector and \( q_0 \) represents the incident photon fluence to the detector per unit area.
The MTF and DQE are generally recognized in the scientific community as primary metrics describing the performance of X-ray imaging systems [22]. In this study, sensitivity was measured by averaging the pixel response values in the obtained images as a function of the X-ray input dose. The NPS was measured as a function of the input air kerma of the detector, and the MTF was measured using a slanted-edge method to avoid aliasing [23–25]. The DQE was calculated with the measured NPS and MTF by using Eq. (1).

Figure 5(a) shows the sensitivities of the dome-grooved scintillator, the nonstructured scintillator, and the Min-R2000 scintillator with various X-ray input doses. For the X-ray feasibility test, under consideration of dental applications, an X-ray beam quality of 60 kVp and a 2.5 mm Al filter thickness were used. The image integration time in the complementary metal–oxide–semiconductor (CMOS) photodiode array increased to 1.0 s due to the low X-ray sensitivity of the scintillator. The Min-R2000 showed higher sensitivity than the other devices due to its 84 µm thickness. The calculated equivalent thickness of the dome-grooved scintillator was roughly 45 µm while the thickness of the non-structured scintillator was 60 µm. Although the two fabricated samples had the same thickness, the nonstructured scintillator had a higher response than the microdome-grooved scintillator. The relatively low sensitivity of the dome-grooved scintillator is thought to be mainly due to two reasons: (i) a relatively small quantity of scintillation material, and (ii) high photon trapping at the Ag reflector surface due to the structural morphology.

Figures 5(b)-(d) provide a comparison of the MTF, NPS, and DQE of the three different scintillators. They are plotted with respect to increasing spatial frequency and are compared at a representative X-ray input dose of 308.83 µGy. Figure 5(b) clearly shows that the dome-grooved scintillator has a much higher MTF than the other two scintillators. The spatial resolution of the nonstructured scintillator and the Min-R2000 at 10% of the MTF are almost the same (approximately 5.7 mm\(^{-1}\)) while the spatial resolution of the dome-grooved scintillator at 20% of the MTF remained up to the Nyquist frequency. As shown in Fig. 5(c), the NPS of the dome-grooved scintillator showed nearly white spectrum noise due to its low sensitivity and low quantum noise.

Although the dome-grooved scintillator had the lowest DQE at low spatial frequency, as shown in Fig. 5(d), it had a lower rate of decrease than the others. This is attributed to the relatively high MTF of the dome-grooved scintillator. The DQE at zero spatial frequency; i.e., \(DQE(0)\), indicates the theoretical maximum performance of the detector. \(DQE(0)\) can be calculated from the quantum efficiency \(\eta\) and the Swank factor \(I\) as follows [6,26]:

\[
DQE(0) = \eta \times I
\]

(2)

where \(\eta\) is a scintillator-related term (i.e., the quantum absorption efficiency (QAE) of the scintillator), and \(I\) is a system noise-related term representing degradation of the signal-to-noise ratio. In this study \(DQE(0)\) was acquired by an approximate DQE value at near zero frequency. The \(DQE(0)\) values of the dome-grooved, nonstructured, and Min-R2000 scintillators were 0.027, 0.047, and 0.073, respectively. The high \(DQE(0)\) of the Min-R2000 is mainly due to the relatively thick scintillation layer and high scintillation material density. The same scintillation material was used for the dome-grooved and nonstructured scintillators. However, the dome-grooved scintillator and the non-structured one had different volume content, approximately 72000 \(\text{um}^3\) and 96000 \(\text{um}^3\), respectively. Therefore, the dome-grooved scintillator had a much lower \(\eta\) value according to variation of the volume content ratio. Also, the almost two-fold lower \(DQE(0)\) value of the dome-grooved scintillator relative to that of the non-structured scintillator was attributed to its relatively lower volume content and its low probabilistic noise factor \(I\) stemming from its low sensitivity, as delineated in Eq. (2). Therefore, further study, including optimal structural design to reduce optical trapping at the
sidewall of the dome groove, is necessary to improve $\eta$ and $I$ by increasing the GOS density of the scintillator. Over the spatial frequency of $2 \text{ mm}^{-1}$, the DQE of the dome-grooved scintillator is higher than that of the nonstructured scintillator; this means that the dome-grooved scintillator can show a clearer image at a high spatial frequency. This high spatial resolution stems from the suppression of excessive light spreading by the microstructured dome groove.

Figures 6(a)-(c) show X-ray images of the resolution bar pattern obtained from the dome-grooved, nonstructured, and Min-R2000 scintillators, respectively. Because of the low sensitivity of the two fabricated scintillators, the display levels of X-ray images obtained from the two scintillators were adjusted to correspond with that of the Min-R2000. The X-ray image obtained from the dome-grooved scintillator is much clearer than the others, verifying that the dome-grooved structure is helpful in enhancing the spatial resolution of the scintillator.

5. Summary

In summary, a flexible microdome-grooved GOS scintillator was simulated, fabricated, and characterized for DR applications. The proposed dome-shape microstructure has a thick lattice at the bottom region and a thin lattice at the top. Due to this tapered structure, the microstructures are easily separated from the metal. The dome-grooved structure offers two attractive features that enhance performance: a high fill factor and high spatial resolution. A low lattice thickness at the top increases the fill factor and the latticed dome shape suppresses excessive light spreading, thereby enhancing spatial resolution. Although the sensitivity of the dome-grooved scintillator is lower than the nonstructured and Min-R2000 scintillators, the dome-grooved scintillator has much higher DQE at a high spatial frequency range. Further study, including efforts to increase GOS density and optimal structural design to reduce optical trapping at the sidewall of the dome groove, is required to improve the sensitivity. In addition, a flexible polymer substrate was used to fabricate the scintillator, and thus the dome-grooved scintillator could easily be bent. Through microfabrication and a performance evaluation of the microdome-grooved GOS scintillator, we confirmed that it could be applied to next-generation flexible DR systems requiring high spatial resolution.

Acknowledgments

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