Development of phosphor-filled micro-well arrays for digital x-ray imaging

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1. Introduction

For better performance of spatial-frequency-dependent detective quantum efficiency DQE(f), the modulation-transfer function (MTF) of an imaging detector should be as large as possible below the Nyquist limit and then drop rapidly, hence a band-limited MTF [1]. This ideal MTF would be achieved if a scintillator is structured to match the underlying pixel size of the readout photosensitive elements [2].

In this study, we made pixel-structured scintillators covering a size of 20 × 30 mm^2 considering intra-oral imaging applications. We filled Gd_2O_2S:Tb phosphor powder with organic binders into the square-shaped well array, which was fabricated by deep reactive ion etching (DRIE) process on a silicon wafer, by sedimentation [3]. The x-ray imaging performance of the prepared samples was investigated in terms of sensitivity, MTF, NPS (noise-power spectrum) and DQE.

2. Materials and Methods

2.1 Sample preparation

Pixel-structured scintillators having three different pixel pitches of 50, 100 and 200 μm were designed, and we denote P50, P100 and P200, respectively, hereafter. All the designs have the same wall (or septum) thickness of 10 μm. The fabrication process mainly consists of two procedures; the realization of the designed square-shaped micro-well arrays on a silicon substrate and filling phosphor materials within the well arrays.

For the realization of pixel-structured scintillators, two-dimensional (2D) array of pixel-structured wells with a depth of 100 μm was prepared by using a DRIE system (Plasmalab System 100, ICP 380, Oxford Instruments Plasma Technology, Ltd., UK) on a Si wafer. Then, Gd_2O_2S:Tb phosphors (UKL65/UF-R1, Phosphor Technology, Ltd., UK) with PVA (polyvinyl alcohol) organic binders were filled within the well array by using a sedimentation method. The coverage of samples was estimated to ~ 50 mg/cm^2.

2.2 Performance evaluation

To readout optical photons emitted from the samples, we employed a CMOS photodiode array (RadEye^{TM}, Rad-icon Imaging Corp., USA) which has a format of 512 × 1024 pixels with a pixel pitch of 48 μm. To avoid damage to the surface of the CMOS photodiode array, we inserted a thin (~100 μm), transparent film between the sample and the photodiode array.

The x-ray imaging performance of the samples coupled to the CMOS photodiode array was investigated in terms of sensitivity, MTF, NPS, and DQE. As a reference, a commercial MinR-2000^{TM} phosphor screen (Carestream Health Care, Inc., USA) was also investigated. To mimic intra-oral imaging conditions, we used a 60 kVp x-ray spectrum from a microfocus x-ray source (Ultrabright^{TM}, Oxford Instruments X-ray Technology, Inc., USA). The spectrum was tailored by a 2.5-mm-thick aluminum considering the attenuation of an x-ray beam through human teeth and the measured half-value layer (HVL) was 2.5-mm-aluminum-equivalent. The distance between the focal spot and the entrance surface of the detector was 300 mm. The dose was measured by replacing the detector with a calibrated electrometer (Piranha R&F/M 605, RTI Electronics AB, Sweden).

3. Results

Fig. 1 shows SEM (scanning electron microscope) pictures depicting comparative morphologies between the reference MinR-2000^{TM} screen [Fig. 1(a) and (c)] and the sample prepared on dummy glass [Fig. 1(b) and (d)]. Fig. 1(e) and (f) shows morphology of Gd_2O_2S:Tb granular phosphors filled in the well array with a pitch of 400 μm.

The measured characteristic curves are plotted in Fig. 2(a). The sensitivities of P50, P100 and P200 samples were respectively about 12, 25 and 41% of the sensitivity of the reference screen. The main reason is thought to be the loss of signal or optical photons in the surrounding silicon wall surfaces because silicon is a strong absorber of optical photons. Considering the fill factor of the samples, the measured sensitivities are plotted in Fig. 2(b).

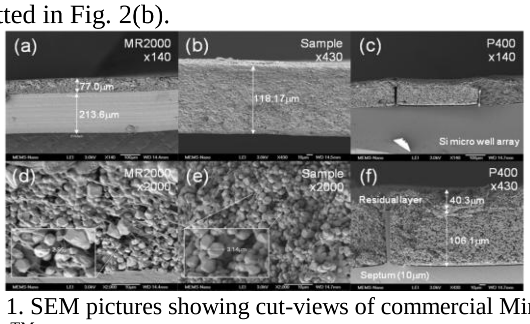

Fig. 1. SEM pictures showing cut-views of commercial MinR-2000^{TM} screen [(a) and (d)], ~118 μm sample on dummy glass [(b) and (e)] and phosphor-filled micro-well array with a pitch of 400 μm[(c) and (f)].

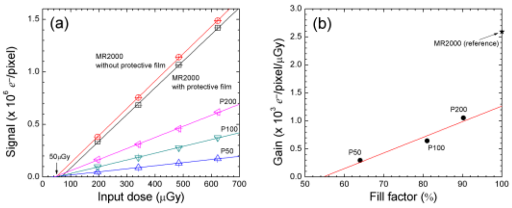

Fig. 2. (a) Characteristic curves of samples with respect to input dose. (b) Gain as a function of fill factor of the samples.

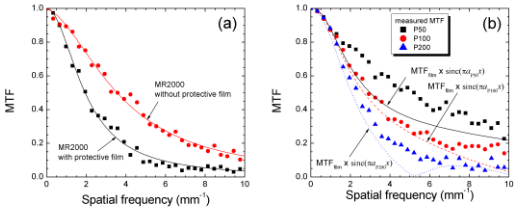

Fig. 3. Measured MTFs of the (a) commercial MinR-2000^{TM} and (b) fabricated samples.

The measured MTF results are plotted in Fig. 3. Fig. 3(a) describes the effect of the protective film inserted between the scintillator sample and the photodiode array on the MTF performance of the detector. The MTF results of the samples are shown in Fig. 3(b) and the estimated MTFs considering the degradation due to the protective film are also plotted.

NPSs of various samples for the same dose of 623 μGy at 60 kVp are shown in Fig. 4(a). The pixel-structured samples show relatively lower spectral densities over the spatial frequency compared to the reference screen. Fig. 4(b) shows the measured NPSs of the P100 sample with respect to various doses.

DQEs of various samples for the same dose of 623 μGy are shown in Fig. 5(a). The magnitude of DQE of the pixel-structured samples over the spatial frequency is much smaller than that of the reference screen due to the lower gain. However, DQEs of the samples are well preserved up to the high-frequency region. DQEs of the P100 sample with respect to various doses are shown in Fig. 5(b) and DQEs are almost independent on the dose.

With the various samples, x-ray images were taken under the intra-oral imaging condition as shown in Fig. 6. An extracted human tooth was imaged.

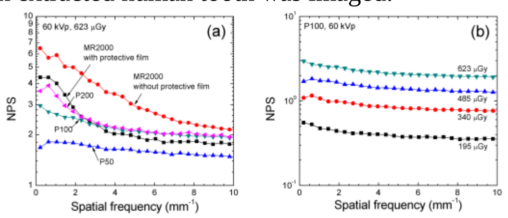

Fig. 4. Measured NPSs for (a) various samples and reference screen and (b) P100 sample with respect to various dose.

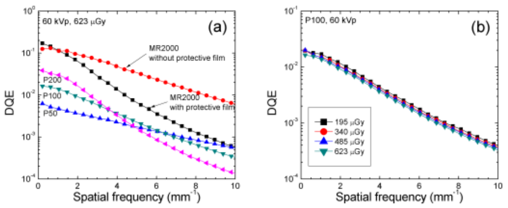

Fig. 5. Measured DQEs for (a) various samples and reference screen and (b) P100 sample with respect to various dose.

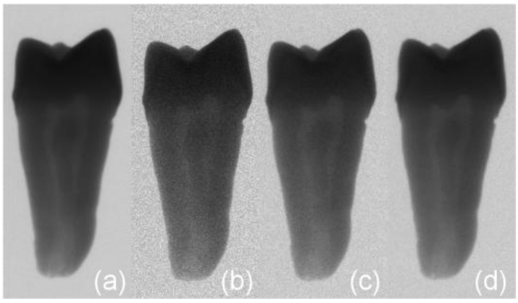

Fig. 6. X-ray images obtained from various configurations of detectors for an extracted human tooth. (a) MinR-2000^{TM}, (b) P50, (c) P100 and (d) P200 scintillators.

4. Discussion and conclusions

We employed two Monte Carlo codes, MCNPX^{TM} (Version 2.5.0., ORNL, USA) and DETECT2000^{TM} (Laval University, Quebec, Canada) for x-ray and optical photon transports, respectively. From the simulation results, the reference MinR-2000^{TM} screen showed the detection of 253 optical photons for a normalized 60 kVp spectrum x-ray photon. However, the P100 sample geometry provided only 54 optical photons which imply that the loss of gain of the pixel-structured scintillator is mainly due to the surface absorption on the silicon walls.

In this study, we have investigated the feasibility of pixel-structured scintillators for high-resolution imaging applications. The pixel-structured samples were realized by filling Gd_2O_2S:Tb phosphors into the silicon well array made by DRIE process. The MTF performance and x-ray image sharpness were promising for high-resolution imaging application. However, the overall performance was limited by the poor optical conversion gain, which is mainly due to the loss of the generated optical photons on the surrounded silicon walls during transports. If we provide a reflectance on the silicon walls and make the pixel-structured scintillator thicker, we would achieve a much improved DQE performance without sacrificing spatial resolving power.

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