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A comparative study of scintillator combining methods for flat-panel X-ray image sensors

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A B S T R A C T

An X-ray transmission imaging based on scintillation detection method is the most widely used radiation technique particularly in the medical and industrial areas. As the name suggests, scintillation detection uses a scintillator as an intermediate material to convert incoming radiation into visible-light particles. Among different types of scintillators, CsI(Tl) in a columnar configuration is the most popular type used for applications that require an energy less than 150 keV due to its capability in obtaining a high spatial resolution with a reduced light spreading effect. In this study, different methods in combining a scintillator with a light-receiving unit are investigated and their relationships are given in terms of the image quality. Three different methods of combining a scintillator with a light-receiving unit are selected to investigate their performance in X-ray imaging: upward or downward oriented needles structure of CsI(Tl), coating layer deposition around CsI(Tl), and insertion of FOP. A charge-coupled device was chosen to serve as the light-receiving unit for the proposed system. From the result, the difference of needle directions in CsI(Tl) had no significant effects in the X-ray image. In contrast, deposition of the coating material around CsI(Tl) showed 17.3% reduction in the DQE. Insertion of the FOP increased the spatial resolution by 38%, however, it decreased the light yield in the acquired image by 56%. In order to have the maximum scintillation performance in X-ray imaging, not only the reflection material but also the bonding method must be considered when combining the scintillator with the light-receiving unit. In addition, the use of FOP should be carefully decided based on the purpose of X-ray imaging, e.g., image sharpness or SNR.

1. Introduction

X-ray flat panel detectors have been widely used devices in X-ray imaging applications. Detection methods in using X-ray flat panel detectors can be divided into two categories: direct or indirect detection. X-ray detection without a scintillator can generate image information by converting X-ray photons directly into electron–hole pairs within the detector using special photoconductive materials, e.g., cadmium telluride, cadmium zinc telluride, and amorphous selenium (a-Se) [1–3]. In contrast, indirect method obtains image information from visible light photons that are converted from X-ray photons via a scintillator. Among many types of scintillators, Cesium Iodide (CsI) and Gadolinium Oxysulfide (Gd_2O_2S) are the most commonly used materials for radiography in the medical imaging and the nondestructive tests. A typical CsI plate produced by chemical vapor deposition (CVD) process is composed of needle-like amorphous structures in the vertical direction or usually referred to as columnar structure. CsI has the advantage of low optical dispersion due to its columnar structure. The disadvantage is that CsI has to be made thick for the same X-ray condition because it has a relatively low density compared to that of Gd_2O_2S [4–6]. Nevertheless, CsI with a columnar structure is the most popular choice for flat panel detector applications.

There are various options to combine CsI scintillators with light-receiving units such as a charge coupled device (CCD) or a complementary metal oxide semiconductor (CMOS) image sensor. The first option is to deposit the scintillation material via CVD onto the detector's surface directly or indirectly. For indirect deposition, the scintillation material is deposited on a different plate and then pressed against the detector surface that to be bonded. The surface in which the scintillation material was deposited initially has a good adhesion with the CsI when the surface temperature of the evaporation material reaches as high as 300 to 500 °C [7]. However, deposition of scintillation material via CVD should be carried out with a caution: the optical characteristics of a photodiode can be changed at a high temperature especially when the photodiode is made out of crystal silicon [8].

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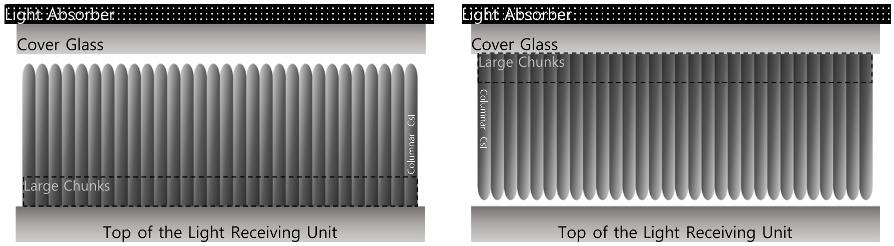

Fig. 1. Scintillation configuration A: upward (left) or downward (right) oriented needles structure. Not in scale.

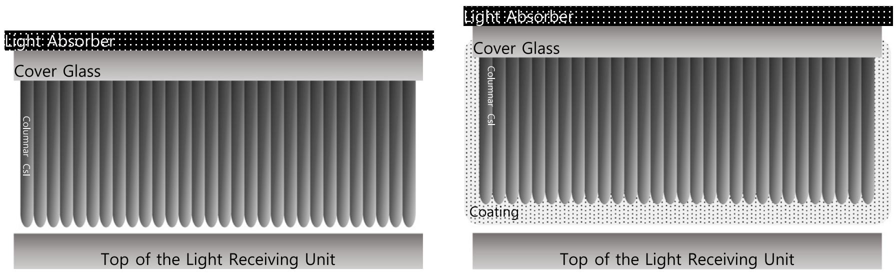

Fig. 2. Scintillation configuration B: coating material deposition. Not in scale.

The second option is to add a protective coating layer around the scintillator. The reason is because CsI is slightly hygroscopic so that it may absorb moisture present in the air and eventually leads to a degradation. To counter such effect, some products apply a protection layer at the final production level via a vacuum housing. Also, there are other products that apply the protection coating on each part of the device. In case of using the protective coating, however, it is possible for the coating material to interact with X-rays and visible photons.

The third option is to use a fiber optic plate (FOP). FOP is a plate constructed with SiO_2 optical fibers to ensure the transparency of photons with visible wavelength. When an incident X-ray photon arrives directly on a crystalline-silicon detector, some unwanted signal may arise within the device such as Fano noise [9,10]. In severe cases, the direct response from X-rays may alter or even destroy the circuit characteristics of the light-receiving unit based on the CMOS technology [11]. By inserting the FOP between the scintillator and the light-receiving unit, it is possible to prevent X-ray photons from directly reaching the light-receiving unit.

Each assembly method has great effects on the price of imaging devices as well as on the quality of the medical image. For example, if a scintillator quality is poor after the direct deposition, a sensor based on crystal silicon should be discarded, and not being reused because of the moisture problem. In the absence of a protective coating, the device housing used to isolate the sensor from moisture can become quite expensive. Furthermore, FOP is one of the most expensive component in the imaging system. By improving the assembly process without causing a loss in the image quality, manufacturers can supply medical X-ray imaging devices at lower cost. In this study, three different methods in combining the CsI scintillator with the light-receiving unit are investigated and their performance are evaluated in terms of light yield, modulation transfer function (MTF), normalized noise power spectrum (NNPS), and detection quantum efficiency (DQE).

2. Materials and methods

2.1. Scintillator structures

Three different approaches in forming CsI structures are considered; upward or downward oriented needles structure of CsI via CVD (Fig. 1), a coating layer deposition around CsI (Fig. 2), and an insertion of FOP (Fig. 3).

The direct deposition of CsI results in a columnar structure with needle-like tips facing in upward-direction whereas the indirect deposition results in needle-like tips facing in downward-direction. A main distinction between two structures is in the location of bundling of needle-like columns forming into larger chunks: at the top-surface of the detector for the upward oriented needles or at the bottom-surface of the light absorber for the downward oriented needles. These "needle clusters" are one of the reasons why the acquired images are blurred. In addition, the upward oriented needles via CVD is limited by the surface that the scintillation material was initially deposited on, leading to a poor deposition efficiency or even making it impossible to deposit the material itself. In this experiment, the top-surface of the detector is corresponded as the top-surface of the cover glass at the bottom. And the bottom-surface of the light absorber is corresponded as the bottom-surface of the cover glass at the top. Fabricated columnar CsI structures from the upward or downward oriented needles structure is shown in Fig. 1.

Furthermore, a protective coating layer can be inserted in between the scintillator and the light-receiving unit. The protection coating material was based on Parylene which is a trade name for various chemical-vapor deposited poly (p-xylylene) polymers and is known to have an excellent hydrophobicity. A 10 μm thick protection coating layer was formed uniformly around the sample. Note that the refractive index for each material at 550 nm are given by the following: p-xylylene is 1.639, CsI is 1.7495, and SiO_2 is 1.4599. CsI scintillators with and without the protection coating were fabricated, as shown in Fig. 2.

Lastly, samples with and without FOP were fabricated to investigate its effect in image quality, as shown in Fig. 3. The FOP manufactured by Hamamatsu was made up of 6 μm-diameter fibers with 3 mm in length. The FOP had a transmittance of 73% for the collimated light at 550 nm and 63% for the diffused light at 550 nm.

All the scintillation structures were manufactured in the same batch so that the thickness and the shape of the deposited scintillators were the same. The size of samples were 15×15 mm^2 on the cover glass and 20×30 mm^2 on the FOP [12]. A small amount of thallium was added to CsI in order to adjust the output photon wavelength to 550 nm, giving a higher chance of photons being captured by the CCD [13]. Fig. 4 shows different depth of views of scanning electron microscope images in the deposited scintillator samples from the same batch. According to the resulting images, the diameter of the scintillator ranged from 2 to 7 μm and the thickness of the scintillator was found to be 220