Performance evaluation of a digital intraoral imaging device based on the CMOS photosensor array coupled with an integrated X-ray conversion fiber-optic faceplate

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Available online 6 April 2007

Abstract

As a continuation of our digital X-ray imaging sensor R&D, we have developed a cost-effective, intraoral imaging device based on the complementary-metal–oxide semiconductor (CMOS) photosensor array coupled with an integrated X-ray conversion fiber-optic faceplate. It consists of a commercially available CMOS photosensor of a 35 \times 35 \mu m^2 pixel size and a 688 \times 910 pixel array dimension, and a high-efficiency columnar CsI(Tl) scintillator of a 90 \mu m thickness directly deposited on a fiber-optic faceplate of a 6 \mu m core size and an 1.46 mm thickness with 85/15 core–cladding ratio (NA \approx 1.0 in air). The fiber-optic faceplate is a highly X-ray attenuating material that minimizes X-ray absorption on the end CMOS photosensor array, thus, minimizing X-ray induced noise at the photosensor array. It uses a high light-output columnar CsI(Tl) scintillator with a peak spectral emission at 545 nm, giving better spatial resolution, but attenuates some of this light due to interfacial and optical attenuation factors. In this paper, we presented the performance analysis of the intraoral imaging device with experimental measurements and acquired X-ray images in terms of modulation transfer function (MTF), noise power spectrum (NPS), and detective quantum efficiency (DQE).

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PACS: 29.30.Kv; 29.40.--n; 29.40.Wk

Keywords: Intraoral imaging system; CMOS photosensor; Digital radiography; Fiber-optic faceplate; Image quality

1. Introduction

Digital intraoral radiographic systems, designed to replace conventional intraoral film radiography, first began to appear on the market in the early 1990s, and their technologies are only now maturing. The digital intraoral systems have gained popularity in clinical practice owing to many advantages over conventional film techniques, e.g., reduction of patient dose, time saving, availability of digital image processing, and elimination of film-processing errors. There have been many research activities focused on the development of radiation imaging sensors of better image quality. Although most of these researches to date have been based on charge-coupled device (CCD) sensors due to their high resolution, high sensitivity, and low noise [1,2], the CCD sensors are sensitive to radiation-induced damage and their production requires a specialized process that adds to fabrication cost [3]. As an alternative, the use of complementary-metal–oxide semiconductor (CMOS) sensors may be a desirable technique shift due to their inherent characteristics of lower power consumption, lower
cost, and, most importantly, higher system integration such as on-chip analog-to-digital conversion (ADC) circuitries [4,5].

In this study, as a continuation of our digital X-ray imaging sensor R&D, we have developed a cost-effective, digital intraoral imaging device based on the CMOS photosensor array coupled with an integrated X-ray conversion fiber-optic faceplate. The purpose of this work was to evaluate the imaging characteristics of the intraoral device, measuring primary characteristics of modulation transfer function (MTF), noise power spectrum (NPS), and detective quantum efficiency (DQE). Although there may be no direct relationship between these physical parameters and the diagnostic yield, the information given by these parameters will provide important information when comparisons with other imaging media are made.

2. Materials and methods

2.1. Intraoral imaging device

A schematic illustration of the fully integrated intraoral imaging device based on the CMOS photosensor array coupled with an integrated X-ray conversion fiber-optic faceplate is shown in Fig. 1. It consists of a commercially available CMOS photosensor of $35 \times 35$ $\mu$m$^2$ pixel size and $688 \times 910$ pixel array size, and a high-efficiency columnar CsI(Tl) scintillator of an about $90 \mu$m thickness directly deposited on a fiber optic faceplate of a $6 \mu$m core size and an $1.46$ mm thickness (NA = 1.0 in air). The fiber-optic faceplate is a highly X-ray attenuating material that minimizes X-ray absorption on the end CMOS sensor ($\sim$0.3% at $70$ kV$_{p}$), thus minimizing X-ray-induced noise at the sensor. It uses a high light-output columnar CsI(Tl) scintillator with a peak spectral emission at $545$ nm, giving high spatial resolution, but attenuates some of this light due to interfacial and optical attenuation factors (transmission at $545$ nm $\sim$65%). The CsI(Tl) scintillator is coated with a multilayer reflector and humidity barrier of an about $100$ $\mu$m thickness.

2.2. X-ray technique

The device was tested using a standard X-ray beam quality, RQA3, described by the IEC standard. The IEC standard defines the various RQA techniques by the added filtration and the half-value layer (HVL) of the beam. The RQA3 beam quality was set up by placing a $10$ mm of aluminum filter in the beam and then adjusting the tube voltage ($kV_{p}$) to obtain a HVL of $3.7$ mm. The $kV_{p}$ setting required to obtain the desired HVL for the used X-ray tube (tungsten target, focal spot size: $0.4$ mm, inherent filtration: $1.0$ mm Al, anode angle: $12.5^\circ$) was about $55$ kV$_{p}$. Fig. 2 shows the X-ray spectrum (photons/mm$^2$/mAs) calculated from the SRS-78 simulator at the RQA3 beam quality for which the performance analysis was done [6]. All measurements were made using a source-to-detector distance (SDD) greater than $30$ cm and a tube current of $6$ mA, varying exposure time ($0.2$, $0.32$, $0.4$, and $0.5$ s; typical exposure time intervals for our intraoral system). For each measurement, a calibrated ion chamber was positioned half way between the X-ray tube and the device in order to measure exposure to the device. The exact exposures, computed using the inverse square of relative distances, were $6.9$, $11.4$, $14.1$, and $17.1$ mR, and the corresponding photon fluences were $1.38 \times 10^6$, $2.24 \times 10^6$, $2.77 \times 10^6$, and $3.35 \times 10^6$ photons/mm$^2$, respectively.

2.3. MTF

A parameter that does carry information about the signal contrast is the MTF, which may be used as a measure of resolution. A one-dimensional (1D) MTF is calculated from the line-spread function (LSF), which is the normalized intensity distribution of the image of an infinitesimally narrow line. When the LSF is known, the MTF can be calculated as the absolute value of the Fourier transform of the LSF. In order to obtain the presampling (or aliasing-free) MTF, which represents the signal before it is discretely sampled by pixels in a digital radiographic system, we measured a finely sampled LSF by using the well-known slanted slit method [7]. Fig. 3 shows the measured MTF curves without any geometrical magnifica-

![Fig. 1. Schematic illustration of the full integrated intraoral imaging device and its photography.](image)

![Fig. 2. X-ray spectrum calculated from the SRS-78 simulator at the RQA3 beam quality for which the performance analysis was done.](image)
tion. Here the ‘ideal’ MTF based on the aperture function of the pixel is also presented for comparison. The spatial frequency at 10% MTF was estimated about 8.5 LP/mm. As shown in Fig. 3, no significant variation was found in the resolution response of the system with photon fluence.

2.4. NPS

Noise characteristics of an imaging system is quantified as the NPS, which expresses the average area occupied by individual photons per unit area (mm²). The NPS is measured from an uniformly irradiated image by using a 2D Fourier analysis method [8,9]. We first acquired an uniform image using the RQA3 beam quality at a given exposure level. Excluding the edge portion of the image, we divided the central portion into more than 10 subregions of 64×64 pixels in size. The image data for each subregion was converted into relative noise by subtracting and dividing the pixel values by its mean value. We performed the 2D Fourier transform for each subregion and scaled the computed NPS for each subregion by its mean value relative to that of a reference subregion to remove large-scale nonuniformities. Finally, we averaged the scaled NPSs for all subregions to obtain the 2D NPS. The 2D NPS was extracted by employing radial averaging of the 2D NPS. Fig. 4 shows the resulting 1D NPS curves. The NPS values at zero frequency were in the range of 4.5×10⁻⁶–1.0×10⁻⁷ mm². As shown in Fig. 4, the noise characteristics of the imaging system was slightly improved with the photon fluence. Assuming that all noise in the imaging system is due to the statistical fluctuation of X-ray photons emitted by the tube, signal and noise characteristics are combined employing the noise-equivalent quanta (NEQ), which describes the number of photons captured by the system per unit area (mm⁻²). The NEQ can be calculated as the reciprocal of the NPS, representing the noise, multiplied by the squared MTF, representing the signal.

2.5. DQE

The DQE is a measurement of the relative number of X-ray photons incident on the imaging system. All radiographic imaging systems are only able to detect a certain percentage of incident photons. The DQE of the system was deduced from the measured MTF, NPS, and exposure, X, along with estimated values for ideal SNR² per mR, q, as described in

\[
\text{DQE}(f) = \frac{G \times \text{MTF}^2(f)}{q \times X \times \text{NPS}^2(f)},
\]

where G is a gain factor that is equal to unity because of the data linearization and the units of NPS data [10,11]. The incident photon fluence ( = q X) was calculated using the measured exposure and the SRS-78 simulation program for X-ray spectrum analysis. Fig. 5 shows the measured DQE curves at the RQA3 beam quality. The DQE values at zero frequency were in the range of 0.16–0.20. The DQE values of the intraoral imaging device, which is coupled with a fiber-optic faceplate, were lower than those we expected, resulting in dose increase to the patient. This may be explained due to the light photon attenuation through the fiber-optic cladding. Thus, the fiber-optic faceplate we incorporated in our imaging device may be helpful to minimize X-ray-induced noise from the photosensor array and also to improve resolution response, but it, instead, may require the sacrifice of patient dose to some extent. Fig. 6(a) shows an example of the X-ray radiographs obtained from the intraoral imaging device with a tooth
phantom. No digital image processing was used after it was acquired. Fig. 6(b) shows the restored image just after being applied by an image restoration algorithm based on the MTF deconvolution which we are developing recently. That work will be presented in a separate paper soon.

3. Conclusion

As a continuation of our digital X-ray imaging sensor R&D, we have developed a cost-effective, intraoral dental imaging device based on the CMOS photosensor array coupled with an integrated X-ray conversion fiber-optic faceplate and evaluated its image characteristics, measuring primary characteristics of modulation transfer function (MTF), noise power spectrum (NPS), and detective quantum efficiency (DQE) at the RQA3 beam quality. According to our test results, the spatial frequency estimated at 10% MTF was about 8.5 LP/mm. There was no significant change in the resolution response of the intraoral device tested with photon fluence. The NPS and DQE values at zero frequency were in the range of $4.5 \times 10^{-6}$–$1.0 \times 10^{-5}\text{mm}^2$ and 0.16–0.20, respectively. The DQE values for the imaging device were lower than those we expected, resulting in dose increase to the patient. This may be explained due to the light photon attenuation through the fiber-optic cladding. Thus, we need to optimize our device components with the trade-off of these image parameters in mind. The image quality information given by this work will provide important information when comparisons with other different imaging media are made.

Acknowledgments

This work was supported by the Basic Atomic Energy Research Institute (BAERI) program of the Ministry of Science and Technology (MOST) under Contract no. M2-0376-03-0000.

References