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Characterization of On-site Digital Mammography Systems: Direct Versus Indirect Conversion Detectors

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We investigated the performances of two digital mammography systems. The systems use a cesium-iodide (CsI) scintillator and an amorphous selenium (*a*-Se) photoconductor for X-ray detection and are installed in the same hospital. As physical metrics, we measured the modulation-transfer function (MTF), the noise-power spectrum (NPS), and the detective quantum efficiency (DQE). In addition, we analyzed the contrast-detail performances of the two systems by using a commercial contrast-detail phantom. The CsI-based indirect conversion detector provided better MTF and DQE performances than the *a*-Se-based direct conversion detector whereas the former provided a poorer NPS performance than the latter. These results are explained by the fact that the CsI-based detector used an MTF restoration preprocessing algorithm. The *a*-Se-based detector showed better contrast-detail performance than the CsI-based detector. We believe that the high-frequency noise characteristic of a detector is more responsible for the visibility of small details than its spatial-resolution performance.

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I. INTRODUCTION

Mammography is the gold standard for the detection of breast cancer [1]. Early detection of breast cancer is directly related to the ability to detect abnormalities visible only on high-quality mammograms. Although screen-film mammography is still used, digital mammography is now popular. Based on the analysis of two sets of screening mammograms (one with film and the other with digital mammography) for almost 50,000 women in a wide range of ages, Pisano *et al.* [2] showed that the diagnostic accuracy was significantly higher for digital mammography when screening younger populations who had dense breasts.

Although computed radiography (CR) systems based on photostimulable phosphors are used for mammography, a recent study showed that the performance of CR systems fell below that of semiconductor-based digital systems, both in terms of physical measurements (such as the modulation-transfer function (MTF), noise-equivalent quanta, and lesion detectability) and cancer detection rates for screening mammography [3]. The detectors used in digital systems can be based on X-ray converters that are phosphors or photoconductors. While the phosphors (*e.g.*, cesium iodide, CsI) convert the absorbed X-ray energy into optical photons, the photoconductors (*e.g.*, amorphous selenium, *a*-Se) convert the absorbed X-ray energy directly into electric charges. Therefore, a detector based on phosphors requires photodiode pixel arrays that detect the optical photons produced in the phosphors. According to these conversion schemes, the detectors based phosphors and photoconductors are typically called “indirect” and “direct” conversion detectors, respectively. Further details on the design, devel-

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Table 1. Summary of the characteristics of the digital mammography systems considered in this study.

	System A	System B
Target material	Mo/Rh	Mo
Focal spot size	0.1/0.3 mm	0.1/0.3 mm
Inherent filtration	Be (> 0.69 mm)	Be (> 0.63 mm)
Optional filtration	Mo (0.03 mm)/Rh (0.025 mm)	Mo (0.028 mm)/Rh (0.028 mm)
Focus-to-detector distance	660 mm	660 mm
Detector design	CsI/PD-TFT array	*a*-Se/TFT array
Pixel pitch	0.1 mm	0.07 mm
Imaging area	$\sim 19 \times 23$ cm^2	$\sim 24 \times 29$ cm^2
Image format	1914×2294 pixels	3325×4096 pixels
ADC bit depth	14 bits	14 bits

opment, and operation of such detectors can be found in the literature [4,5].

Because these two designs use different processes during X-ray-to-electric signal conversion, image qualities are subject to their own unique characteristics. While the image quality of digital mammography systems has been widely investigated in the past [6–8], insufficient attention has been given to the characterization of systems based on on-site direct measurements. In this study, we evaluate the performance of two types of digital mammography detectors installed in the same hospital: system A (GE Senographe DS) and system B (Hologic Selenia) use the indirect and the direct conversion schemes, respectively. We analyzed the images obtained from the systems as they were, without further processing. For example, we accepted the manufacturers' preprocessing algorithm for the gain-offset correction. We quantitatively investigated their performances in terms of the MTF, the noise-power spectrum (NPS), and the detective quantum efficiency (DQE), including a contrast-detail analysis. In addition, we qualitatively investigated their image quality by using a breast tissue-equivalent phantom.

II. MATERIALS AND METHODS

1. System descriptions

System A uses an indirect flat-panel detector based on a CsI combined with amorphous silicon (*a*-Si) photodiode (PD)/thin-film transistor (TFT) array [9]. System B uses a direct flat-panel detector based on an *a*-Se photoconductor with an *a*-Si TFT array [10]. Table 1 summarizes the characteristics of the systems under investigation.

2. Measurements of the MTF, NPS, and DQE

The signal-transfer performance of an imaging system can be characterized by using the spatial-frequency-dependent MTF, MTF(u, v), and the noise-transfer performance can be characterized by using NPS(u, v), where u and v denote the Fourier conjugates of the spatial variables x and y, respectively, in Cartesian coordinates. For a linear and shift-invariant (LSI) imaging system, these two Fourier metrics can be combined and then expressed as a single metric as follows [5,11–15]:

$$\begin{aligned} \mathrm{DQE}(u,v) &= \frac{\overline{q}\,\overline{G}^2\,\mathrm{MTF}^2(u,v)}{\mathrm{NPS}(u,v)} = \frac{\overline{d}^2\,\mathrm{MTF}^2(u,v)}{\overline{q}\,\mathrm{NPS}(u,v)} \\ &= \frac{\mathrm{MTF}^2(u,v)}{\overline{q}\left[\mathrm{NPS}(u,v)/\overline{d}^2\right]}, \end{aligned} \quad (1)$$

where $\overline{q}$ (mm^{-2}) is the average input X-ray quantum distribution, and $\overline{G}$ (DN mm^{-2}) is the detector gain relating $\overline{q}$ to the average pixel output $\overline{d}$ in units of digital number (DN). Because the DQE includes the signal correlation in space (MTF) and the second-moment statistics in noise (NPS), the concept of the DQE is universally adopted as a primary metric for characterizing the imaging performance of X-ray imaging systems. In radiology, moreover, image quality excellence is a balance between system performance and patient radiation dose, and this balance is best described by the DQE of the system [5,16,17]. We should note that while the MTF is a multiplicative factor, the NPS is not. Therefore, the MTF should not include any aliasing whereas the NPS does include the effect of noise aliasing. Accounting for these issues, the measurement procedure of DQE has been standardized [12,18,19].