Chapter 2
Digital Radiography Detectors: A Technical Overview

Abstract Developments in digital detector technologies have been taking place and new digital technologies are available for clinical practice. This chapter is intended to give a technical state-of-the-art overview about computed radiography (CR) and digital radiography (DR) detectors. CR systems use storage-phosphor image plates with a separate image readout process and DR technology converts X-rays into electrical charges by means of a readout process using TFT arrays. Digital detectors offer several advantages when compared to analogue detectors. The knowledge about digital detector technology for use in plain radiograph examinations is thus a fundamental topic to be acquired by radiology professionals and students. In this chapter an overview of digital radiography systems (both CR and DR) currently available for clinical practice is provided.

Keywords Computed radiography • Digital radiography • Detectors • Storage-phosphor image plates • Image readout process • Electrical charges • X-ray • Readout process • TFT arrays • Analogue detector • Plain radiograph • Clinical practice

Introduction

Several digital systems are currently available for the acquisition of projection radiographs. Digital radiography systems have been replacing traditional analogue or screen–film (SF) systems over the last three decades. The transition from an SF environment to a new digital environment should be considered as a complex process. Technical factors concerning image acquisition, management of patient dose, and diagnostic image quality are some issues that could influence this process. In a transition process from SF to digital, patient radiation doses could increase 40–103% [1]. When compared to SF, digital technology could increase patient radiation doses due to the wide dynamic range they have. However, the dynamic range is useful because it contributes for a better clinical image quality when compared to traditional SF systems [2]. This is an important difference among

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analogical and digital technologies. The risk of overexposure with no adverse effect on image quality could be present. Digital imaging systems could facilitate over- or underexposure that influences a patient's dose. Overexposure could provide good-quality images, but may cause unnecessary patient dose. Although several advantages over SF systems are identified, considerable variations in image quality and effective dose can be achieved among different digital detectors [3].

According to Busch [4] the choice of the radiographic technique, the radiation dose delivered to the patient, and the diagnostic quality of radiographic image are three core aspects of the imaging process aiming the management of patient dose and image quality. This is a challenge for radiographers because clinical advantages and limitations of digital technologies for projection radiography are also dependent on the radiographer's options for a particular patient examination.

The knowledge about digital detector technology for use in plain radiograph examinations is thus a fundamental issue to be acquired by radiology professionals and students. Several literature reviews concerning digital radiology detectors have been provided by some authors [5–11]. In this chapter an overview of computed radiography (CR) and digital radiography (DR) currently available for clinical practice is provided.

Overview of Computed Radiography and Digital Radiography Detectors

Developments in digital detector technologies have been taking place and new digital technologies are available for clinical practice. Table 2.1 shows a timetable of developments in digital technologies since the early 1980s.

The first digital radiography system using the basic principle of the conversion of the X-ray energy into digital signals utilizing scanning laser stimulated luminescence (SLSL) was developed by Fuji (Tokyo, Japan) and introduced in the market in the beginning of the 1980s [12]. In the mid-1980s, the storage phosphor systems

Table 2.1 Timetable of developments in digital technologies

Year	Digital technology availability
1980	Computed radiography (CR), storage phosphors
1987	Amorphous selenium-based image plates
1990	Charge-coupled device (CCD) slot-scan direct radiography (DR)
1994	Selenium drum DR
1995	Amorphous silicon–cesium iodide (scintillator) flat-panel detector
	Selenium-based flat-panel detector
1997	Gadolinium-based (scintillator) flat-panel detector
2001	Gadolinium-based (scintillator) portable flat-panel detector
2001	Dynamic flat-panel detector fluoroscopy–digital subtraction angiography
2006	Digital tomosynthesis
2009	Wireless DR (flat-panel detector)

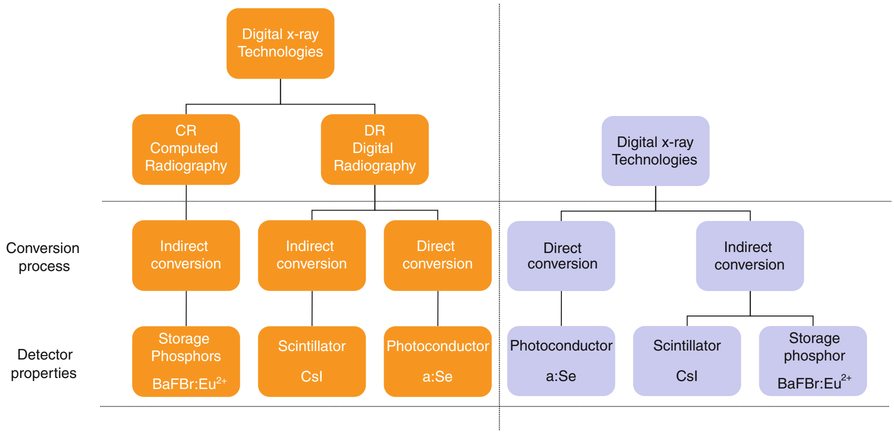

Fig. 2.1 Taxonomy of digital radiography technologies

became a new clinical application as a new imaging method for exposures at the wall stand, the Bucky table, and bedside imaging. The high technical requirements and financial costs, associated with limited image quality and difficult handling—without a reduction of examination time—delayed the transfer of storage phosphor systems into routine clinical use, which started to increase at the beginning of the 1990s [4]. Today the storage-phosphor radiography systems or CR systems play a fundamental role in the field of digital projection radiography. Other important innovation was the development of flat-panel detectors in the middle of 1995. Flat-panel detectors were initially developed to be integrated detectors in the radiology equipment, but more recently they are available as nonintegrated detectors and working as a wireless or a non-wireless technology.

Digital systems are traditionally split into two broadly defined categories [10, 11]: computed radiography and digital radiography. Although this taxonomy is commonly accepted other classifications are described [13]: direct digital radiography and indirect digital radiography technologies (including CR). In this case the detector classification is related with the conversion process of X-ray energy to electric charge. Figure 2.1 shows a schematic figure that includes a comparative diagram of the taxonomy of digital radiography technologies, the conversion process, and the detector properties.

Other taxonomic option is to give a classification according to the integration of the digital detector within the radiology equipment: in this case integrated and nonintegrated detectors terminology could be used.

Despite the taxonomy that is used the major difference among digital technology systems related with X-ray detection and readout process. Concerning CR systems they use storage-phosphor image plates with a separate image readout process, which means an indirect conversion process; DR technology converts X-rays into electrical charges by means of a direct readout process using thin-film transistor (TFT) arrays.

Table 2.2 Three components of digital detectors

Detector technology		Capture element	Coupling element	Charge readout
Computed radiography (CR)		$BaFBr:Eu^{2+}$	Photostimulated luminescence (PSL) light-guide	Photomultiplier tube; signal digitization
Direct radiography (DR)	Direct conversion	a-Se	None	TFT array
	Indirect conversion	CsI or Gd_2O_2S	Contact layer	a-Si photodiode/ TFT array

TFT thin-film transistor liquid crystal display

Table 2.2 shows the differences among detector technology concerning three components of digital detectors [14]: the capture element, the coupling element, and the charge readout element.

CR technology uses an indirect conversion process using a two-stage technique. X-rays are captured at a storage-phosphor screen (SPS) (e.g.: $BaFBr:Eu^{2+}$) and then a photodetector captures the light emitted from the SPS and converts the captured luminescence into a corresponding digital image.

DR detectors can use either a direct or an indirect process for converting X-rays into electric charges. These detectors use direct-readout by means of a TFT array despite the conversion process of the X-ray beam. Direct-conversion detectors have an X-ray photoconductor—such as amorphous selenium (a-Se)—that converts directly at only one stage X-ray photons into electric charges.

Indirect-conversion systems use a two-stage technique for conversion. They have a scintillator, such as cesium iodide (CsI) that converts X-rays into visible light at a first stage. That light is then converted—at a second stage—into an electric charge by means of an amorphous silicon photodiode array [15].

Despite the process of X-ray detection and readout digital detectors offer several advantages when compared to SF systems. These include wide dynamic range, adjustable image processing, better image quality, rapid image acquisition, and image access at remote locations [16].

Computed Radiography

Computed radiography was the first available digital technology for projection radiography. CR technology is based in SPS and its first clinical application by Fuji took place at the early 1980s.

This technology uses a photostimulable detector replacing the traditional SF cassettes. The storage-phosphor plates are exposed inside the cassettes with standard dimensions for typical plain radiography and no change of generator, X-ray tube, and Bucky wall or table mounted system is necessary. CR technology allows the radiographer to obtain plain radiography images like in a traditional SF system.

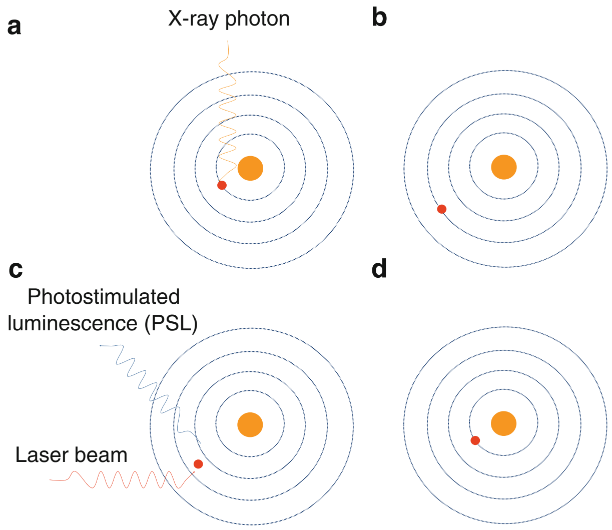

Fig. 2.2 SPS exposure and PSL; *SPS* storage-phosphor screens, *PSL* photostimulated luminescence

The difference is how the latent image is created and how this image processing is done. The basic CR imaging cycle has three steps [13]: (1) expose, (2) readout, and (3) erase.

Inside the radiography cassette an image plate (IP)—or SPS—having a detective layer of photostimulable crystals is available. The detective layer consists of a family of phosphors BaFX:Eu^{2+} where X can be any of the halogens Cl, Br, or I (or an arbitrary mixture of them) [17]. A typical SPS can store a latent image for a considerable period of time. However, according to the American Association of Physicists in Medicine [18], it will lose about 25% of the stored signal between 10 min and 8 h after an exposure resulting in the loss of energy through spontaneous phosphorescence.

The phosphor crystals are usually cast into plates into resin material in an unstructured way (unstructured scintillators) [10]. When the SPS is exposed to the X-ray the energy of the incident radiation is absorbed and excites electrons to high-energy levels (Fig. 2.2a, b). These excited electrons remain trapped at unstable energy levels of the atom. The absorbed X-ray energy is stored in crystal structure of the phosphor and a latent image is then created at these high-energy states giving a spatial distribution of these electrons at the SP detector. This trapped energy can be released if stimulated by additional light energy of the proper wavelength by the process of photostimulated luminescence (PSL) (Fig. 2.2) [18].

After the X-ray exposure and the creation of the latent image, the SPS is scanned in a separate CR reader device. The readout is a process that follows exposure of the image plate and constitutes the second step of the CR imaging cycle. A red laser beam

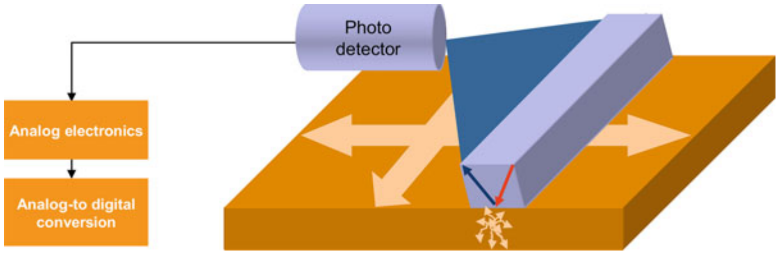

Fig. 2.3 SPS scanning process; *SPS* storage-phosphor screens

scans the photostimulable screen stimulating the emission of blue light photons under the excitation of the laser beam. When the detective layer of the IP is scanned pixel by pixel with a high-energy laser beam of a specific wavelength, stored energy is set free as emitted light having a wavelength different from that of the laser beam [10]. This triggers the process of PSL resulting in the emission of blue light in an amount proportional to the original X-ray [17] and setting free the excited electrons to their lower energy level (Fig. 2.2c, d). This light is collected by photodiodes and converted into electric charge while an analog-to-digital device converts it into a corresponding digital image. Figure 2.3 shows the SPS scanning process.

Finally the third step of the basic CR imaging cycle is the residual signal erasure. Residual latent image electrons are still trapped on higher energy levels after readout. This energy is erased after the readout process using a high-intensity white light source that flushes the traps without reintroducing electrons from the ground energy level [18].

Digital Radiography

Digital radiography flat-panel systems with integrated readout mechanisms were introduced in the market by the end of the 1990s [19]. Flat-panel systems, also known as large-area X-ray detectors, integrate an X-ray-sensitive layer and an electronic readable system based on TFT arrays. Detectors using a scintillator layer and a light-sensitive TFT photodiode are called indirect-conversion TFT detectors. Those using an X-ray-sensitive photoconductor layer and a TFT charge collector are called direct-conversion TFT detectors [19]. The reference to amorphous silicon (a-Si), which is used in TFT arrays to record the electronic signal, should not be confused with a-Se, the material used to capture X-ray energy in a direct digital detector. The structure of a DR flat-panel system is shown in Fig. 2.4.

This electronic readable system allows an active readout process, also called active matrix readout, in opposition to the storage phosphor systems where no active readout elements are integrated within the detector. The entire readout process is very fast, allowing further developments in digital real-time X-ray detectors [19].

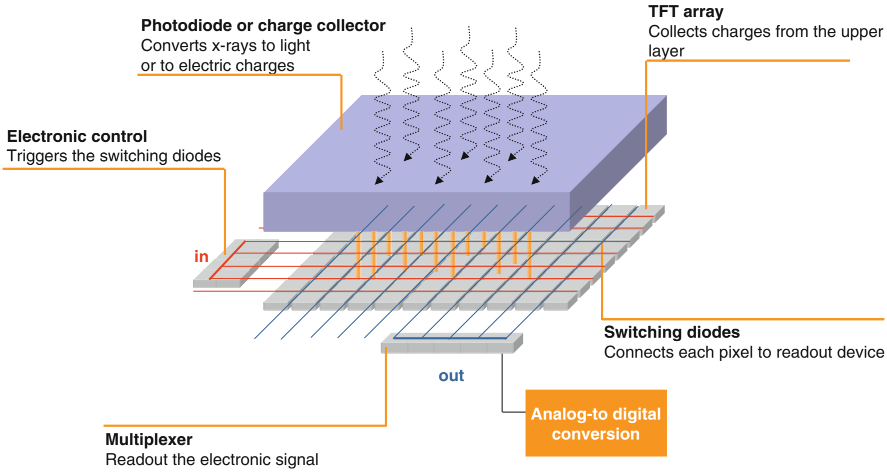

Fig. 2.4 Flat-panel structure

TFT arrays (Fig. 2.5) are typically deposited onto a glass substrate in multiple layers, with readout electronics at the lowest level, and charge collector arrays at higher levels.

Depending on the type of detector being manufactured, charge collection electrodes or light-sensing elements are deposited at the top layer of this "electronic sandwich" [20].

The advantages of this design include compact size and immediate access to digital images. The performance of DR systems greatly exceeds the performance of CR systems, which have conversion efficiencies of 20–35%, and of screen–film systems for chest radiography, which have nominal conversion efficiencies of 25% [20].

Wireless DR flat-panel systems have become commercially available by 2009. Wireless DR systems are nonintegrated detectors that could be used to obtain radiographs in a similar way to CR. With wireless DR detector it is mandatory to use a wireless LAN for communications between the DR detector unit and the workstation console. This way each performed radiograph is transferred at almost real time from the cassette DR to the workstation. The DR cassette includes a built-in battery to power supply and this allows the detector's necessary autonomy to obtain several radiographs and to transfer the obtained radiographs to the system for further viewing.

Large-Area Direct-Conversion Systems

Large-area direct-conversion systems use a-Se as the semiconductor material because of its X-ray absorption properties and extremely high intrinsic spatial resolution [19, 20].

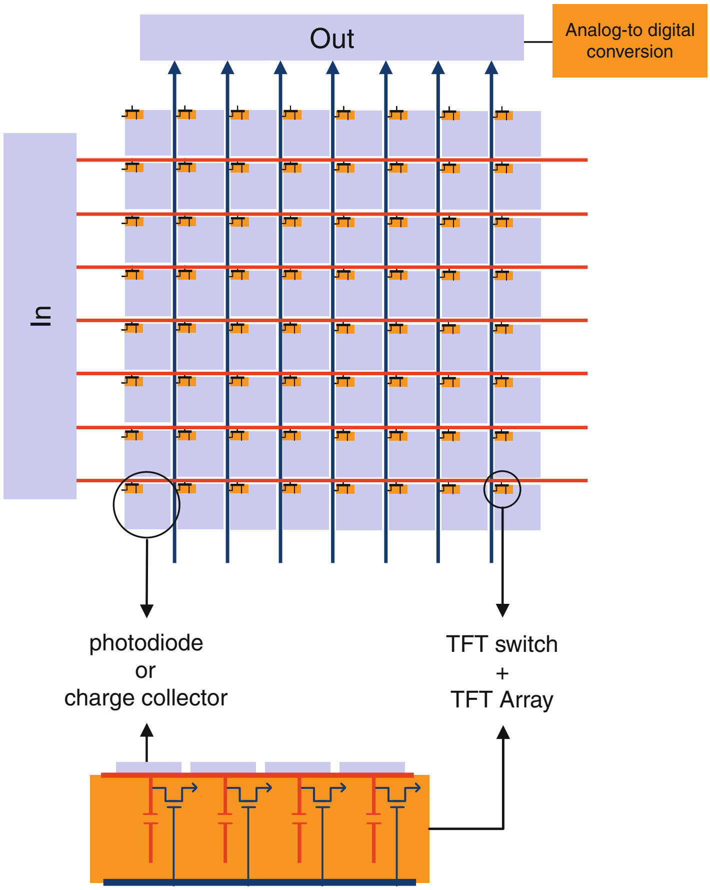

Fig. 2.5 TFT array; *TFT* thin-film transistor

Before the flat panel is exposed to X-rays an electric field is applied across the selenium layer. Then the X-ray exposure generates electrons and holes within the a-Se layer: the absorbed X-ray photons are transformed into electric charges and drawn directly to the charge-collecting electrodes due to the electric field. Those charges—proportional to the incident X-ray beam—are generated and migrate vertically to both surfaces of the selenium layer, without much lateral diffusion. At the bottom of the a-Se layer, charges are drawn to the TFT charge collector, where they are stored until readout. The charge collected at each storage capacitor is amplified and quantified to a digital code value for the corresponding pixel. During the readout, the charge of the capacitors of every row is conducted by the transistors to the amplifiers.

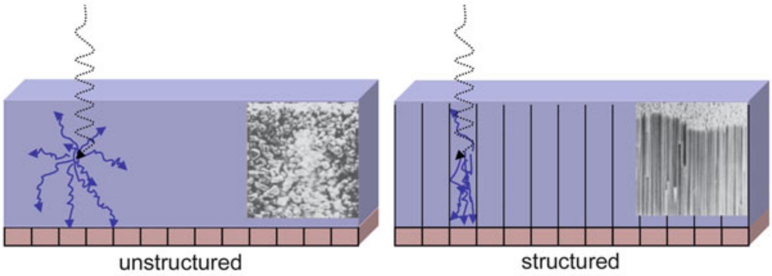

Fig. 2.6 Schematic of an unstructured (*left*) and structured scintillator (*right*)

Large-Area Indirect-Conversion Systems

Large-area indirect-conversion systems use CsI or gadolinium oxisulphide (Gd_2O_2S) as an X-ray detector. The scintillators and phosphors used in indirect-conversion detectors can be either structured or unstructured (Fig. 2.6). Unstructured scintillators scatter a large amount of light and this reduces spatial resolution [14]. Structured scintillators consist of phosphor material in a needlelike structure (the needles being perpendicular to the screen surface). This increases the number of X-ray photon interactions and reduces the lateral scattering of light photons [14].

When the scintillator layer is exposed to X-rays the beam is absorbed and converted into fluorescent light. At a second stage that light is converted into an electric charge by means of an a-Si photodiode array [15]. Indirect conversion detectors are constructed by adding an a-Si photodiode circuitry and a scintillator as the top layers of the TFT sandwich. These layers replace the X-ray semiconductor layer used in a direct-conversion device [20]. The active area of the detector is divided into an integrated array of image elements—the pixel—and each element contains a photodiode and a TFT switch available for the readout process.

Recent developments for a novel pixel-structured scintillation screen with nanocrystalline Gd_2O_3:Eu particle sizes for high-spatial-resolution X-ray imaging detectors are being made for indirect X-ray imaging sensors with high sensitivity and high spatial resolution [21, 22].

Summary

Different digital technologies are currently available for clinical practice in plain radiography. CR and DR technologies constitute a remarkable improvement based on detector technology developments. The specific properties and capabilities of a digital detector influence the choice of the radiographic technique, the radiation dose delivered to the patient, and the diagnostic quality of radiographic image. Although SF and digital technology (CR and DR) coexist at the present time in many countries the trends in the near future seem to point towards the digital technology.

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