Comparison of Different Radiographic Image Receptors

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Abstract

The aim of this study is to compare detector plates (radiographic image receptors) used in digital radiology. Detector plates are divided into computed radiography systems (phosphor plates) and digital radiography systems (CCD, CMOS, silicon photodiodes and flat panel detector). Phosphor plates use the principle of radiation-induced photostimulable luminescence, and can be read with the help of a laser reader (digitizer). Digital radiology systems are divided into indirect and direct; indirect are those that require the conversion of x-rays into photons of visible light and use scintillators for that matter, while direct x-rays are directly converted into charges with the use of photoconductors. Out of all the scintillators, cesium iodide has shown the best properties through research, while amorphous selenium proved to be the optimal photoconductor. CCD, CMOS and silicon photodiode are exclusively indirect digital detector systems, while flat panel detector can be found in both direct and indirect format. CCD slot scan detector system with full frame signal reading mode, active pixel CMOS detector, and silicon photodiode with back-illuminated PDA have shown the best results among detectors of their kind, while indirect type flat panel detectors with cesium iodide scintillator proved to be the optimal choice in modern radiography as evidenced by research with PBU-S-3, CDRAD 2.0 and TRG phantom.

Keywords: detector plates, computed radiography, digital radiography, scintillators, photoconductors

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Introduction

Digitalization of radiology marks the beginning of modern radiology, and the beginning of use of the detector plates plays key role in that process. New systems that don't require the long lasting film processing and that are more easily stored and kept occur. It all started with the development of phosphor plates for medical applications. After developing the first functional PACS (Picture Archiving and Communications System) in 1990, it was possible to transfer and store radiographic images within a health care facility without the use of films and removable disks [1]. The technological development of CCD (Charged Coupled Device) and CMOS (Complementary metal-oxide-semiconductor) detectors mark the beginning of digital radiography, and further progress lead to the creation of silicon photodiode that are a part of CT device, and a flat panel detector that has wide application [2]. The aim of this article is to compare computed radiography and digital radiography, as well as the detectors used in digital radiography.

Methods and materials

Computed radiography

"Computed radiography (digital luminescence radiography) is an indirect imaging method" that „uses phosphor plates as an image receptor and whose image is read with the use of laser beams" [3]. The phosphor plate uses the principle of radiation induced photostimulable luminescence. It usually consists of a layer of photostimulative crystals of alkaline earth halides and latanoid europium- BaFBr: $Eu2^{+}$ which are in charge of storing the signal, and the cassette in which they are located. The system is complemented with the HeNe (Helium neon) or a diode laser reader used to emit light and obtain a digital image. A laser reader that prints a radiographic image on a disk drive or PACS can be found separate or integrated into the device. The speed of the image readout process depends on the reader, but always takes less than a minute [4]. Figure 1 shows phosphor plates of various sizes.

Figure 1 Phosphor plates of various sizes

Source: http://g-sonic.com.qa/wp-content/uploads/2019/07/IPS-Plates-1.jpg

Electrons stored in the BaFBr: $Eu2^+$ layer excite $Eu2^+$ into $Eu3^+$, and create a formation of high energy electrons and electron holes called photostimulated luminescent complex- PSLC or fluorescent (F) center. The number of active F-centers obtained in the photostimulable luminescent layer is proportional to the number of absorbed X-rays. A $Eu3^+$-F-center complex is formed. „Stimulation of the $Eu3^+$-F-center complex and release of the stored electrons requires a minimum energy of ~2 eV, most easily deposited by a highly focused laser light source of a specific wavelength" [5]. The most commonly used source is a diode laser with a wavelength of 680nm. During the interaction with $Eu3^+$, the electron descends into a lower energy state with the emission of energy in the form of a photon with energy of 3eV. The energetical state of $Eu3^+$ is switched into a more stable state in the form of $Eu2^+$. Newly created photons are then optically filtered from the rest of the laser light and directed into a light guide that transmits them to a photomultiplier tube that amplifies their signal and converts them into a corresponding output voltage. With use of analog to digital converter, we

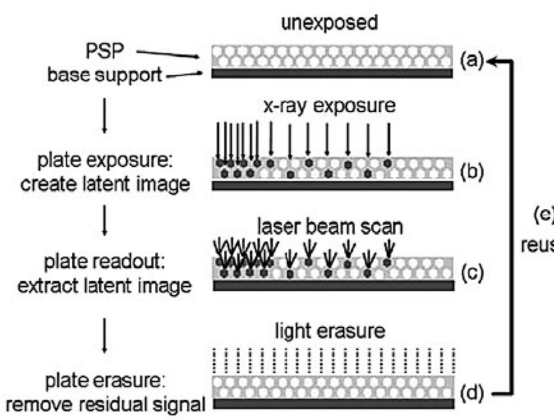

Figure 2 Schematic representation of operation of a phosphor plate

Source: https://www.aapm.org/pubs/reports/RPT_93.pdf

get a digital image. The preparation of the phosphor plate for the next exposure takes place in the reader and the entire surface of the phosphor plate is illuminated with strong light with corresponding wavelengths that remove the electrons remaining in BaFBr:$Eu2^+$ [5]. The principle of operation of the phosphor plate is schematically shown in Figure 2.

Digital radiography

Digital radiography is a method of obtaining a radiographic image with the help of detector plates. The system itself performs the process of converting X-ray energy into an electrical signal without the need for processing in a separate reader. Digital radiology systems can be divided into indirect digital radiography and direct digital radiography [3].

Scintillators

Scintillator is a part of indirect digital radiography detector that converts photons of X-rays into photons of light. It is located on top of the detector and is directed towards the radiation source. When attenuated X-rays „collide" with the scintillator, it absorbs their energy, and the electrons inside the scintillator go from a stable to an excited state. Upon returning into a stable state, the electron releases its energy in the form of light emission detected by a photodetector that generates electrical charges that can be digitized. The most commonly used scintillators are cesium iodide (CsI), gadolinium oxysulfide (GOS) and gadolinium silicate (GSO) [6].

Research [7] compares the image quality obtained using 9 digital radiographic plates with cesium iodide and gadolinium oxysulfide scintillators. A plexiglass CDRAD 2.0 phantom with 255 cylindrical fields of different depths and diameters, 20 PPMA boards between which the phantom is placed, and a CDRAD Analyzer were used. IQF (Image quality figure) (Figure 3) alongside inverse IQF (Figure 4) are then calculated.

Results from the research [7] shown in Tables 1 and 2 suggest that cesium iodide scintillators produce better

Exact observation ratio = (Exact observation / total number of squares) x100

D (i, th) – the smallest diameter that is correctly observed

C – contrast values

$$IQF = \sum_{i=1}^{15} C_i \times D(i,th)$$

Figure 3 Image quality figure (IQF)

$$IQF_{inv} = \frac{100}{\sum_{i=1}^{15} C_i \times D(i,th)}$$

Figure 4 Formula for inverse IQF

Source: https://www.researchgate.net/publication/261483259_Evaluation_and_comparison_of_image_quality_for_indirect_flat_panel_systems_with_CsI_and_GOS_scintillators

IQF values compared to gadolinium oxysulfide scintillators. Also, there is a greater difference in image quality between cesium iodide scintillators from different manufacturers above the 150 µGy dose than with the gadolinium oxysulfide scintillators.

Table 1 Mean values and standard deviations for IQFInv of CsI scintillator [7]

CsI scintillators		
Dose(µGy)	**Mean value of IQFinv**	**Standard deviation of IQFinv**
50	0.425	0.055
100	0.472	0.049
150	0.527	0.227
200	0.639	0.316

Table 2 Mean values and standard deviations for IQFInv of GOS scintillator [7]

GOS scintillators		
Dose(µGy)	**Mean value of IQFinv**	**Standard deviation of IQFinv**
50	0.275	0.022
100	0.288	0.023
150	0.304	0.042
200	0.324	0.056

Photoconductors

Photoconductor is a part of a direct digital radiography detector that converts X-ray photons directly into an electric charge. Crystals of mercury iodide (HgI2) and cadmium zinc telluride (CdZnTe) and the most commonly used stabilized amorphous selenium (a-Se) are used as photoconductor materials. The absorption of attenuated X-rays on the photoconductor creates charges, and that process is called the photoelectric effect. In order to conduct the charge towards the capacitor, the photoconductor is under constant voltage which is obtained from the main electrode at the top of the photoconductor and the pixel electrodes at the bottom of the photoconductor which is schematically shown in Figure 5 [8].

Digital detector plates

CCD detector

CCD (Charge Coupled Device) is an indirect digital detector plate. There are two types of CCD detector systems: lens coupled CCD system and slot scan CCD system. In the lens coupled system, the photons of light generated on the scintillator are reflected from a mirror that directs them onto an optical lens that focuses the light beam on the entire size of the CCD detector (Figure 6). The slot scan CCD system uses two or more connected CCD detectors that move synchronously with the X-ray tube (Figure 7) [2]. Lens coupled CCD systems provide a weaker image quality than the slot scan CCD system due to the lower amount of photon arrivals on the plate and this results in a lower signal-to-noise ratio, and lower quantum efficiency [1].

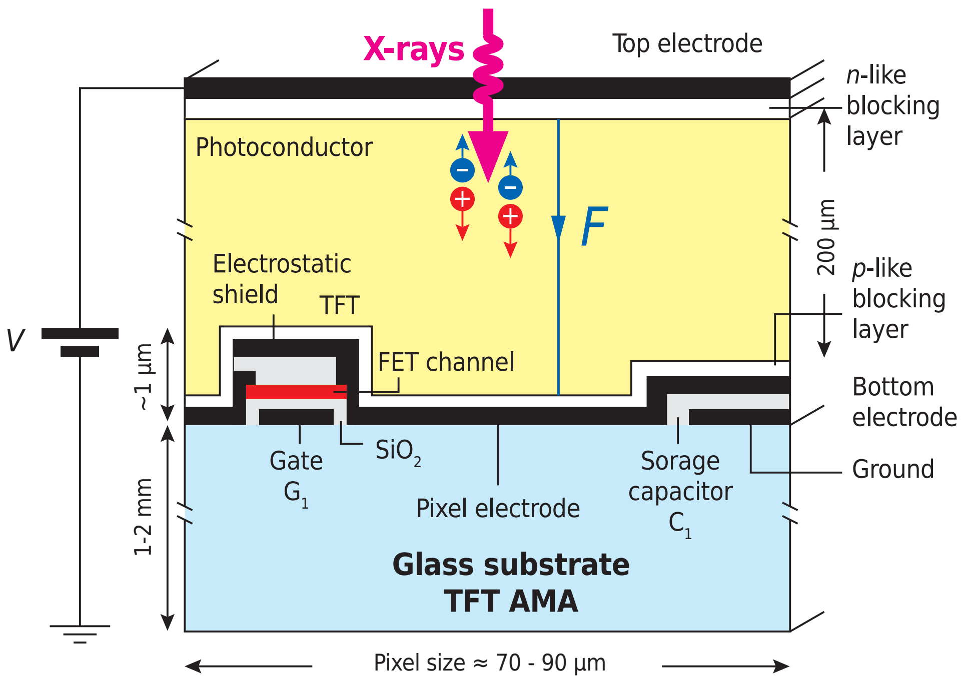

Figure 5 Schematic representation of cross section of a pixel with photoconductor

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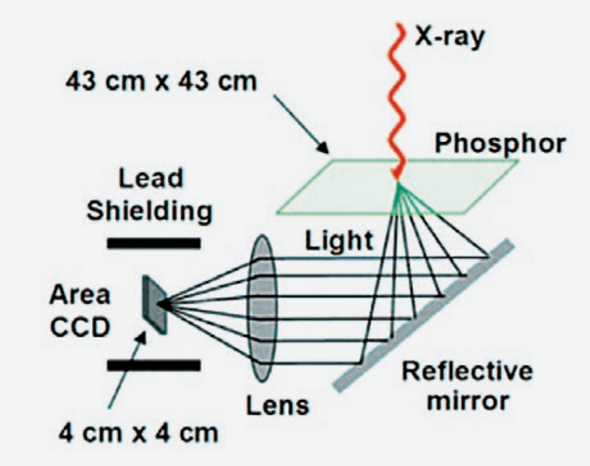

Figure 6 Lens coupled CCD system

Source: https://www.a-dimaging.com/wp-content/uploads/2014/10/CCD-Info.jpg

The basis of the CCD detector is a semiconductor plate made of silicon (Si) with an integrated circuit. The basic unit of a CCD detector is a pixel containing three photodiode electrodes responsible for the photoelectric effect (Figure 8). The charges are stored in a capacitor and their amount is proportional to the number of photons generated by the scintillation process. There are two types of charge transfer: full frame and interline mode. In full frame mode, the native image is sent to the amplifier as a whole; the charges from the pixels descend towards the bottom of the CCD where they reach the readout row that directs them into a common amplifier (Figure 9). The image fill coefficient (Fill factor) on such a detector is 100% [9]. Interline reading boards consist of pixels partially covered by metal lines that transfer charges to the readout row at a speed determined by the shutter at a certain number of frames per second (Figure 10). Such plates have a reduced fill factor and thus a reduced sensitivity [10].

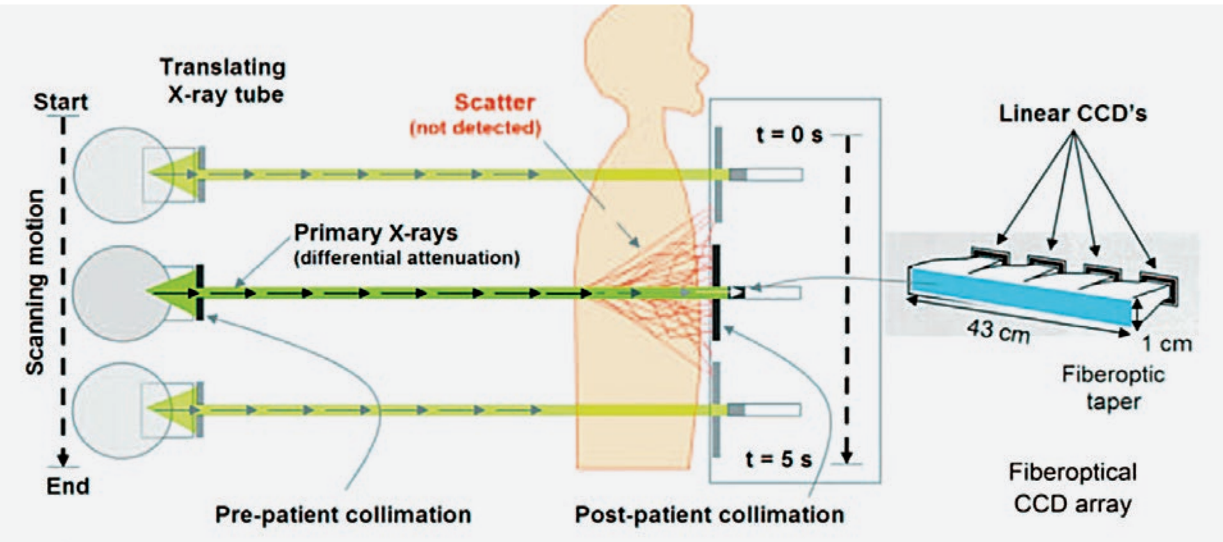

Figure 7 Slot scan CCD system

Source: https://appliedradiology.com/articles/digital-radiography-the-bottom-line-comparison-of-cr-and-dr-technology

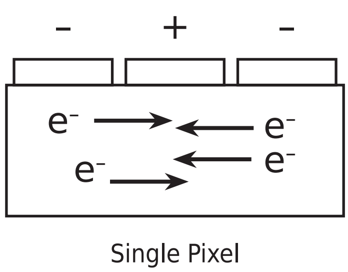

Figure 8 Pixel in CCD detector

Source: https://www.aapm.org/meetings/04SS/documents/yester2.PDF

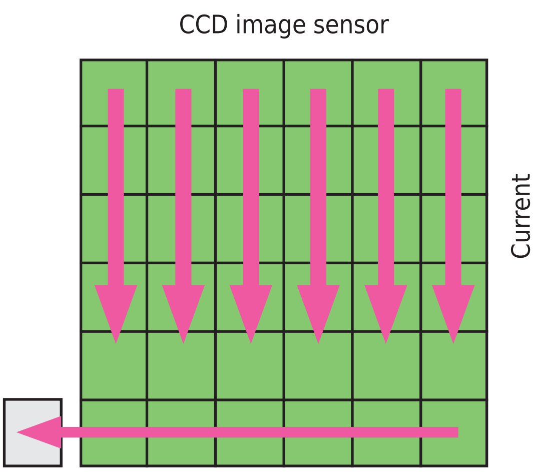

Figure 9 Full frame charge transfer type

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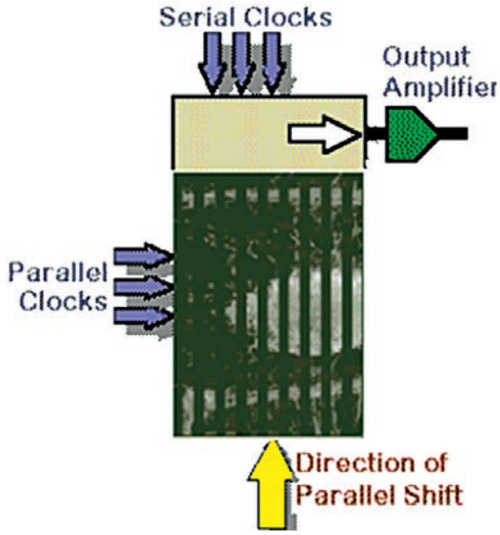

Figure 10 Interline charge transfer type

Source: https://www.fer.unizg.hr/_download/repository/MK_CCD_BW.pdf

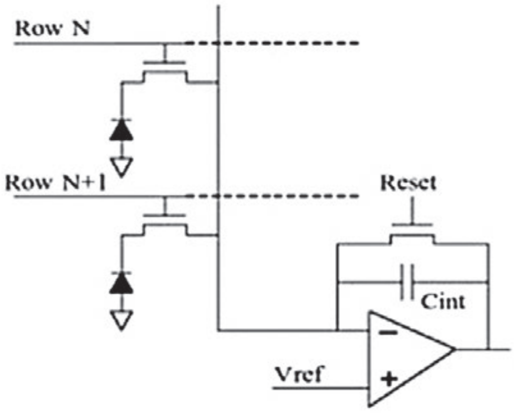

Figure 12 Schematic representation of a pixel assembly with a transistor and reset switch in a CMOS detector with passive pixels

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CMOS detector

CMOS (Complementary metal-oxide-semiconductor) is an indirect digital detector plate used in dental radiography for orthopantomogramic imaging and CBCT (Cone beam computed tomography). It consists of a scintillator, a photodiode and an integrated circuit. The CMOS detector directly converts light from the scintillator into a digital signal with its integrated analog-to-digital converter, but therefore has limited photosensitive space unlike other image receptors [11]. There are two types of CMOS detectors: passive pixel CMOS detector and active pixel CMOS detector [10].

In passive pixel CMOS detectors, each pixel is addressed using row and collumn detectors (Figure 11). After exposure to X-rays and scintillation, light reaches pixels where electric charges are generated by the photoelectric effect. The charges are stored in a capacitor and sent towards the end of each pixel column to the charge integrator amplifier (CIA) at a frequency regulated by its switch. That process results with a digital signal- output voltage that is proportional to the input charge on the amplifiers. Non-illuminated pixels are charged with dark current, which is established in all pixels after the charge from the photoelectric effect has arrived at the CIA transistor with the help of a reset switch, and this marks the beginning of a new cycle of signal generation [10]. The pixel assembly with the transistor and reset switch is schematically shown in Figure 12.

The active pixel CMOS detector uses amplifier technology in each pixel. This allows higher operating speed and sensitivity, and an increase in the overall size of the detector. The pixel in such detector consists of a photodiode and three transistors: a reset transistor, an RS (Row select) transistor for determining the row of the pixels

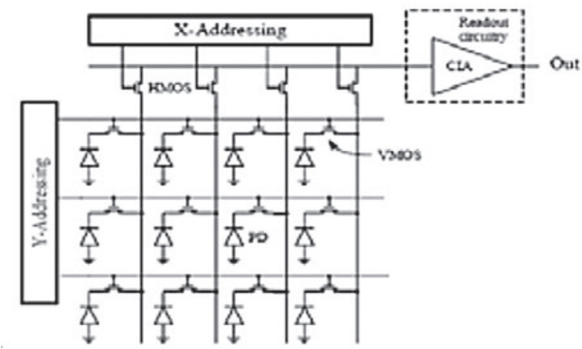

Figure 11 Schematic representation of the pixel arrangement in a CMOS detector with passive pixels

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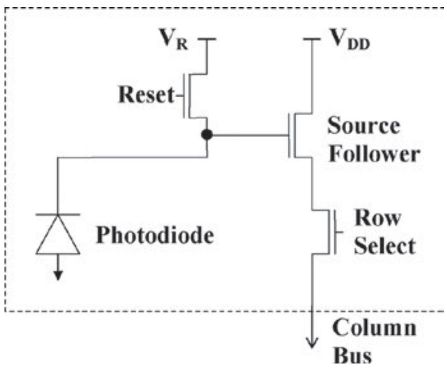

Figure 13 Schematic representation of pixels in a CMOS detector with active pixels

Source: https://www.researchgate.net/profile/Sarah_Bohndiek/publication/3432305/figure/fig1/AS:652960354795530@1532689288611/Active-pixel-sensor-3-T-pixel.png and a SF (Source follower) transistor that serves as an amplifier (Figure 13). Turning on the reset transistor brings the referent voltage to the diode, which is then read and remembered on the sample and hold assembly in correlated double sampling- CDS [10]. CDS is used to cancel FPN (Fixed pattern noise); noise generated by uniform detector exposure [12]. The charge accumulates in the photodiode, and is sent towards the amplifier during the time controlled by the reset transistor. Finally, the voltage is read from the RS transistor and interpreted so that the current of brightly lit pixels gives a lower voltage, while the current of dimly lit pixels gives a higher voltage [10].

Silicon photodiode

Silicon photodiode is an indirect digital detector plate that has replaced gas detectors in a computed tomography device. Due to its dimensions of only a few millimeters, systems composed of multiple connected photodiodes are used, which we call multi-slice systems. Thus, 64-layer and 128-layer computed tomography systems are used today. Silicon photodiode consists of a scintillator, a photovoltaic detector array (PDA) and a silicon board (substrate) with an integrated circuit that sends signals to the amplifier [13].

According to the type of photovoltaic detector array, silicon photodiodes are divided into two types: front-illuminated PDA and back-illuminated PDA. The first type has the electric circuit located on the same side (front) as the photosensitive area, while the second type has the electric circuit transferred to the rear of the detector. Back-illuminated PDAs are much easier to maintain and easier for instaling the scintillators, so they are used much more today [14].

Placing another layer of scintillator and photovoltaic detector under the already present components opens the possibility for dual energy imaging (Figure 14). Attenuated low-energy X-rays are absorbed by the upper PDA, while attenuated high-energy X-rays pass through the upper layer and reach the lower scintillator and PDA. This method of imaging contributes to the contrast of objects and to the quality of radiological diagnosis [14].

Flat panel detector

Flat panel detector is a detector plate available in direct and indirect variants of digital technology. In direct systems, amorphous selenium (a-Se) is used, while the indirect systems use cesium iodide (CsI) or gadolinium oxysulfide (GOS). Flat panel detectors are most often found in a 41x41 cm format that consists of about seven million pixels. Their main advantages are high dynamic range, high quantum efficiency (QE), high linearity, and high signal-to-noise ratio. For direct and indirect technology systems, it is common that each pixel consists of one TFT (Thin-film transistor) transistor and two other components. In direct systems these are the charge collecting electrode and the capacitor, while in indirect systems these are the photodiode and the capacitor as part of the photodiode. All pixels in the same row are connected by gate lines to the gate switches that control the operation of the flat panel detector. Likewise, all pixels in the same column are connected by data lines. This way, the position of each pixel in the matrix of the flat panel detector is determined, which is schematically shown in Figure 15 [15].

Flat panel detector is an integrating detector in which the integration time for each pixel is equal to the frame time. At equal time intervals, the TFT transistor closes the circuit under the influence of the positive voltage coming from the gate line. Then it conducts current and releases the stored electrons into the data line that leads to the bottom of the detector where the charge integration amplifier is located and it converts the charge into a proportional voltage-signal [16].

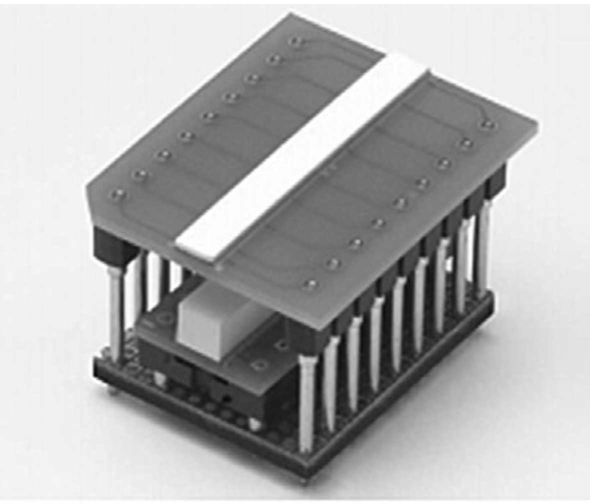

Figure 14 Hamamatsu dual energy photodiode

Source: https://www.hamamatsu.com/resources/pdf/ssd/e09_handbook_xray_detectors.pdf

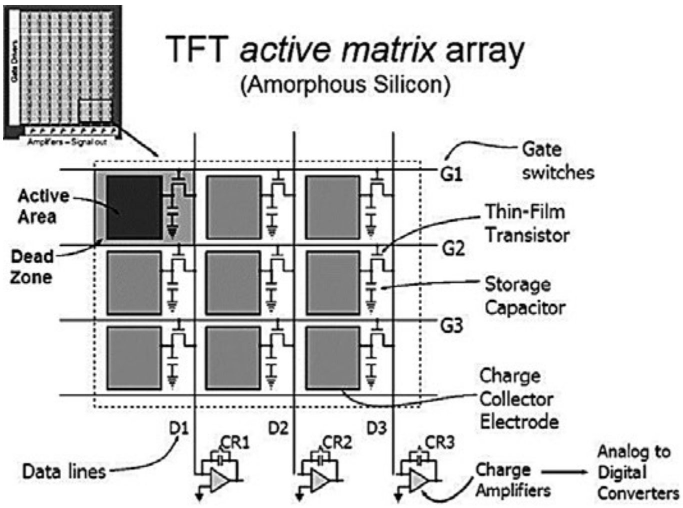

Figure 15 Schematic representation of part of the matrix of a direct flat panel detector

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Results

Comparison of computed radiography and digital radiography

Computed radiography and digital radiography are similar in terms of need of using computers and data networks to process, transmit, and store digital images. Phosphor plate systems are more affordable, and are compatible with older X-ray devices. Furthermore, portable X-ray devices most commonly use phosphor plate systems because of ease of transmission and research speed. Detective quantum efficiency (DQE) for computed radiography systems is about 30%, while for digital radiography systems it ranges from 60%-65%. Also, image quality of computed radiography systems is inferior to digital radiography systems and can only be compared with film and foil systems [17]. Digital radiography systems have an automatic exposure system, and they are characterized by higher speed than computed radiography systems, in which there is also a possibility of signal lag in the phosphor plate before re-exposure [18].

Comparison of CCD detectors, CMOS detectors and flat panel detectors

CCD detector with full frame charge transfer type proved to be the optimal CCD detector due to the 100% fill factor. CCD detectors usually have problems with image blur due to hypersaturated pixels and blooming when excess charge from the capacitor spreads to the nearby pixels (Figure 16). CMOS detector has a significantly lower ability to convert photons into charges, thus lower sensitivity and DQE than a CCD detector. It also has problems with nonuniformity (Fixed pattern noise) (Figure 17) [19]. This problem is partially solved by correlated double sampling [20]. In conclusion, a CMOS detector with active pixels is cheaper than a CCD detector due to its inferior image quality [21].

Research [22] compares the detection capabilities of lung lesions of four flat panel detectors and two CCD detectors of different properties. An anthropomorphic phantom (PBU-S-3) and simulations of lesions (288) and nodules (144) were used for the study. Results shown in Table 3 suggest that the CCD slot scan system gives the optimal results in detecting changes in the lungs, but also that it is only slightly better than flat panel detector systems. The worst results were shown by the indirect flat panel system with gadolinium oxysulfide, and the lens coupled CCD system.

Table 3 Research results according to the number of detected lesions and nodules [22]

Detector type	Detected lesions (%)	Detected nodules (%)
FPD CsI-1	120 (42)	51 (35)
FPD CsI-2	118 (41)	47 (33)
FPD GOS	110 (38)	47 (33)
FPD Se	117 (41)	48 (33)
Slot scan CCD	132 (46)	62 (43)
Lens coupled CCD	113 (39)	47 (33)

The study also suggests much lower entrance skin doses and effective doses in indirect flat panel systems with a cesium iodide scintillator compared to other systems (Table 4). The highest entrance skin and effective dose was obtained while recording with a direct flat panel system, while the doses of both CCD systems are comparable to those obtained with an indirect flat panel system with a gadolinium oxysulfide scintillator [22].

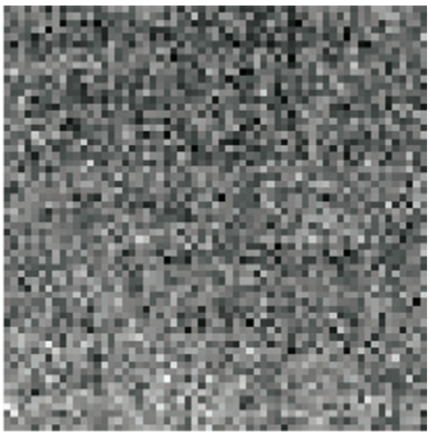

Figure 16 Nonuniformity in CCD detector

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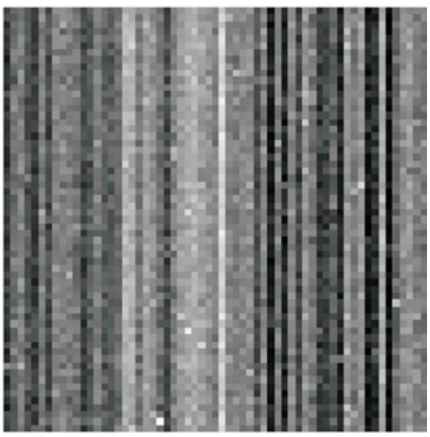

Figure 17 Nonuniformity in CMOS detector

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Table 4 Entrance skin doses and effective doses for individual detectors [22]

Detector type	Entrance skin dose (μGy)	Effective dose (mSv)
FPD CsI-1	30	0.009
FPD CsI-2	38	0.012
FPD GOS	87	0.020
FPD Se	92	0.028
Slot scan CCD	60	0.022
Lens coupled CCD	57	0.017

Table 5 IQF values while testing with the CDRAD 2.0 phantom [23]

Entrance skin dose (mGy)	IQF for FPD CsI	IQF for FPD Se
~40.7	32.0	37.8
~58.2	26.6	31.3
~75.4	25.0	24.1
~119.4	23.9	26.1

~ Average value of the measured entrance skin dose during exposure of two detectors by radiation of the same X-ray tube with the same exposition parameters

Comparison of direct and indirect flat panel detectors

Research [23] compares a direct flat detector with an amorphous selenium photoconductor and an indirect flat detector with a cesium iodide scintillator using CDRAD 2.0, and TRG phantom. The results shown in Table 5 suggest that at higher doses (75 and 119 mGy) both detectors show similar IQF values, while at lower doses the indirect flat panel detector gives an image of significantly better contrast resolution than its rival.

Better results in these tests result from higher DQE values in indirect flat panel detectors than direct flat panel detectors; the higher atomic number and density of the cesium iodide layer enables it to have a higher photon absorption capacity compared to the amorphous selenium layer. Also, such detectors produce slightly less noise than their rivals, which affects the image quality to a limited extent, and adds some value [23].

Conclusion

Modern radiology is inconceivable without the detector plates that introduced radiology into the digital age. Silicon photodiode technology has found application in computed tomography instead of the older gas detector system. The indirect type of flat panel detector with a cesium iodide scintillator has shown the best results when the image quality and protection against excessive irradiation of the patient are taken into account because it requires the lowest dose to obtain an optimal image compared to other detectors. The flat panel detector is therefore the most widely used detector in radiography, with applications ranging from pulmonary and dental radiography, all the way to diascopy and CBCT. CCD detector represents as a good alternative to flat panel detector due to its lower cost, while CMOS detector due to its speed but also limitations in image quality has found great application in dental radiography. ■

Sažetak

Cilj ovog rada je usporedba detektorskih ploča (receptora radiološke slike) koji se koriste u digitalnoj radiologiji. Detektorske ploče dijele se na sustave kompjutorizirane radiografije (fosforne ploče), te sustave digitalne radiografije (CCD, CMOS, silicijske fotodiode i flat panel detektore). Fosforne ploče koriste princip zračenjem uzrokovane fotostimulabilne luminiscencije, te se mogu očitati uz pomoć laserskog čitača (digitalizatora). Sustavi digitalne radiologije dijele se na neizravne i izravne; neizravni su oni koji zahtijevaju pretvorbu x-zraka u fotone vidljive svjetlosti i za to koriste scintilatore, dok izravni x-zračenje izravno pretvaraju u naboje i za to koriste fotokonduktore. Od svih scintilatora, cezijev jodid je kroz istraživanje pokazao najbolja svojstva, dok se amorfni selenij pokazao kao optimalni fotokonduktor. CCD, CMOS i silicijska fotodioda su isključivo neizravni digitalni detektorski sustavi, dok se flat panel detektor može naći i u izravnom i neizravnom formatu. CCD *slot scan* detektorski sistem sa *full frame* načinom očitavanja signala, CMOS detektor s aktivnim pikselima, te silicijska fotodioda sa stražnjim osvjetljenjem PDA pokazali su najbolje rezultate među detektorima svoje vrste, dok se neizravni tip flat panel detektora sa scintilatorom od cezijevog jodida pokazao kao optimalan odabir u modernoj radiografiji što je dokazano uz pomoć istraživanja sa PBU-S-3, CDRAD 2.0 i TRG fantomom.

Ključne riječi: detektorske ploče, kompjutorizirana radiografija, digitalna radiografija, scintilatori, fotokonduktori

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