PHYSICS

E. Kotter
M. Langer

Digital radiography with large-area flat-panel detectors

E. Kotter (✉) · M. Langer
Department of Diagnostic Radiology,
Freiburg University Hospital,
Hugstetterstrasse 55,
79106 Freiburg, Germany
e-mail: eko@mrs1.ukl.uni-freiburg.de
Tel.: +49-761-2703802
Fax: +49-761-2703831

Abstract Large-area flat-panel detectors with active readout mechanisms have been on the market for the past 2 years. This article describes different detector technologies. An important distinction is made between detectors with direct and those with indirect conversion of X-rays into electrical charges. Detectors with indirect conversion are built with unstructured or structured scintillators, the latter resulting in less lateral diffusion of emitted light.

Some important qualities of flat-panel detectors are discussed. The first phantom and clinical studies published report an image quality at least comparable to that of screen-film systems and a potential for dose reduction. The available studies are summarised in this article.

Keywords Digital radiography · Computed radiography · Screen-film radiography · Storage phosphor radiography

Introduction

There are many motivations for radiologists and hospital administrators to introduce digital radiography systems. One major concern of radiologists is of course obtaining better image quality. A second interesting feature is the separation of the image acquisition from the image-processing step. From an organizational point of view, the capability to store images electronically in a picture archiving and communications system (PACS) is most interesting to both radiologists and administrations. Furthermore, digital radiography systems allow acceleration of patient throughput by different means (less data typing, shorter time to image and no cassette manipulation for some systems).

Historically, the first digital radiography system was the storage phosphor radiography system (SPR), developed by Fuji (Tokyo, Japan) and introduced to the market in the beginning of the 1980s [1]. It was followed approximately 10 years later by a dedicated digital chest radiography unit, using a drum coated with a layer of amorphous selenium as detector (Thoravision, Philips, Hamburg Germany). Large-area flat-panel radiography detectors have been introduced only recently. This article reviews the different technologies of flat-panel radiography detectors (FPD) and summarises studies of clinical relevance.

Present standard in digital radiography: storage phosphor radiography

Since its introduction in the beginning of the 1980s [1] SPR has made its way into routine clinical radiography. A consensus conference held in Germany has shown that leading German radiologists estimate that SPR is suitable for most clinical applications [2]. Image quality of SPR systems has been continuously improved over the past years. The latest generation of storage phosphor plates has detective quantum efficiencies (DQE, indicating the efficiency of a detector system in detecting incident X-ray quantas; see below) comparable with that of conventional screen-film systems (SFS) [3]. The spatial resolution of SPR is of 2.5 line-pairs/mm with a pixel size of 200 µm for routine applications. Recently, SPR systems with a resolution of 5 line-pairs/mm for an image format of 35×43 cm and even a resolution of 10 lp/mm for a plate size of 24×30 cm (mammography) have been developed.

Advantages of digital radiography

Even when image quality of SPR systems was still somewhat lower than at present, these systems had a good acceptance among radiologists because of the more constant image quality they deliver. Over- or underexposed radiographs can be corrected because the image processing is separated from the image acquisition step; thus, retakes are drastically reduced. All digital radiography systems allow processing of images in a way that good local contrast can be obtained even for regions with very different densities, e.g. the lung regions and the mediastinum.

By receiving patient demographic and examination data from the radiology information system (RIS) via the Digital Image Communication in Medicine (DICOM) worklist management service, sending back data about the examination to the RIS (DICOM performed procedure step) and sending the images to the PACS, digital radiography systems greatly enhance productivity of the radiology technician [4, 5]. One of the inconveniences of SPR systems is that the radiology technician still has to manipulate the cassette with the image plate. It has been shown that productivity is further enhanced when the cassette manipulation step is eliminated with FPD systems [6]. This has also been proven in installations with a dedicated thoracic radiography system, the Thoravision system (Philips, Hamburg Germany).

Terminology

In the past, and in many published articles, the term computed radiography has been used as a synonym for storage phosphor radiography. To differentiate the newer flat-panel detector systems with direct readout from SPR, manufacturers and subsequently authors have used the term digital radiography. Since both detector systems work with computers, and both systems are digital by nature, these terms should both be used without distinction for both systems. Like screen-film systems, the term storage phosphor radiography describes the detector used to produce an image. For the new class of flat-panel detectors, sharing the property of a direct readout with an active matrix, either the terms flat-panel detector, active matrix flat panel detector or flat panel detector with direct readout should be used.

Types of flat-panel detectors

A radiographic flat-panel detector is a digital, electronically readable radiography system (see Table 1 for a classification of X-ray detectors). By definition, the detector is a slim system (flat panel) that can be integrated into existing Bucky tables. This excludes, for example,

Table 1 Classification of X-ray detectors. *TFT* thin-film transistor; *CCD* charge-coupled devices

Film-screen systems
Digital detector systems
Portable systems
Storage phosphor radiography
Conventional storage phosphor plates
Needle–crystalline screens
Stationary systems
Selenium detector with electrometer readout (Thoravision system)
Scintillators with CCD camera
Flat-panel detectors (with TFT panels)
Indirect conversion of X-rays (scintillators with photodiode and TFT readout)
With nonstructured scintillator
With structured (needle-shaped) scintillator
Direct conversion of X-rays
Photoconductor+TFT-panel readout

the Thoravision system and (most) charge-coupled devices (CCD)-based detectors. In opposition to SPR systems, the readout electronics are integrated within the detector cassette (direct readout mechanism). In order to be usable in a general diagnostic room, the detector needs to have a large dimension, i.e. at least 35×43 cm (which excludes specialised mammographic detector systems from this article).

Different types of flat-panel detector systems are available (Fig. 1).

Charge-coupled devices detector systems

Because they are not flat-panel detectors, CCD detector systems are not the main topic of this article; however, since they are the main technology competing with flat-panel detectors, a short description of their construction is included.

Charge-coupled devices were the first direct-readout detectors to be introduced in radiology approximately 30 years ago. A CCD optical detector chip is composed of up to several million independent pixels (picture elements) where incoming photons induce charge. When illuminated the array of pixels in the chip accumulates a charge pattern corresponding to the illumination pattern. This charge pattern can then be read out pixel by pixel, be digitised and transferred to computer memory.

The X-ray quanta are transformed into visible light by an X-ray scintillator like gadolinium disulphide. The visible light is then transmitted to a CCD detector very similar to those used in video cameras. The CCD transforms the visible light into a digital image.

Since CCDs are physically much smaller (2–4 cm^2, integrated on a chip) than the image area, the light emitted by the scintillator screen must be transmitted to the

Fig. 1 Direct and indirect X-ray detection. Charge-coupled devices (CCD) detectors invariably use an optical coupling between the scintillator and the CCD chip which is much smaller than the scintillator surface. In indirect conversion thin-film transistor (TFT)-based detectors X-ray photons are converted into visible light in the scintillator layer. A photodiode converts the visible light into electrical charges which the TFT array reads out row after row. Direct conversion flat-panel detectors use a layer of amorphous selenium to convert X-ray photons directly into electrical charges which are stored in capacitors to be read out by the TFT-array. (Modified from [33])
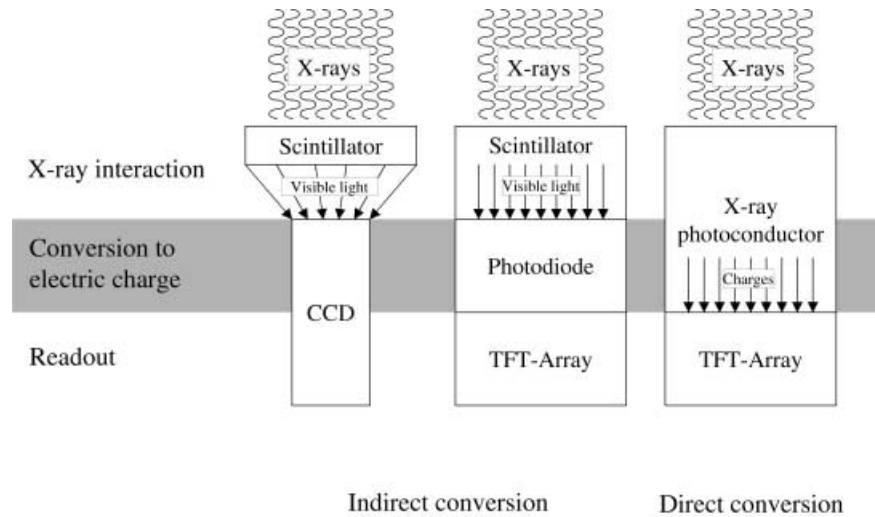
CCD by some kind of optics reducing the size of the image. This demagnifying system can consist of lenses or optical fibres, and sometimes an image intensifier is used. Whereas CCDs are very sensitive detector systems, any of these optical systems reduces the number of photons that reaches the CCD, thus increasing noise of the system and degrading image quality. Geometric distortions and light scatter are another consequence of the use of optical reduction. Finally, but less importantly with modern, cooled CCDs, thermal noise within the CCD itself can degrade the image quality.

The most important inconvenience of CCD-based X-ray detectors is the thickness of the detector system needed because of the optical system which has to be integrated into the detector. As a result, these systems are direct readout digital X-ray detectors but not flat-panel detectors.

Thin-film transistor-based, flat-panel detectors

Whereas the CCD electronics are integrated on a chip, thin-film transistor (TFT) arrays present with a surface as large as the image to be detected. Thin-film transistor-arrays are fabricated in a sandwich technique by depositing the electronics on a glass substrate in several layers. Typically, the readout electronics are at the bottom and charge collector electrodes are placed in the middle of the array. Depending on the type of detector, the top of the array is constituted by X-ray or light-sensitive elements, or a combination of both. Detectors using a scintillator and a light-sensitive photodiode are called indirect conversion TFT detectors, whereas those using only an X-ray sensitive photoconductor are called direct-conversion TFT detectors.

The charge collector electrode drains the charge generated by the photoconductor or the photodiode. The
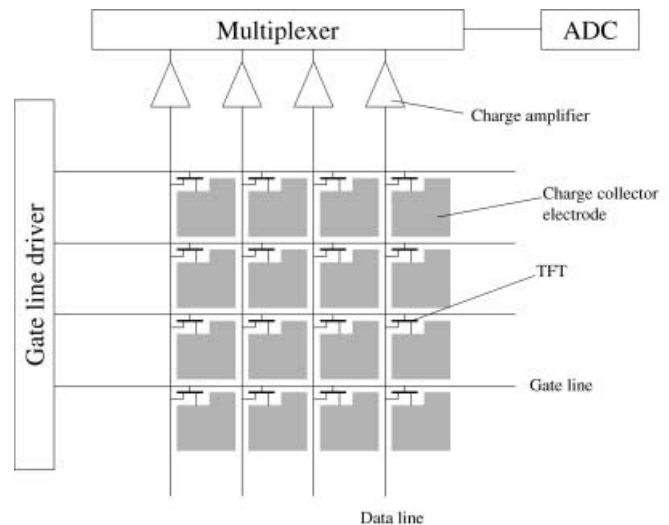
Fig. 2 Construction principle of a flat-panel detector showing electrical connection of charge collecting capacitors with TFTs. The line voltage is set high for one row after another. When the gate line voltage is high, all transistors of the row conduct the pixel charges collected in the charge collecting electrode (or capacitor) to the adjacent data lines. Charge amplifiers and multiplexers integrated in the detector then read out the signal for that row. (Modified from [32])

charge collected is proportional to the X-rays received by the detector. During the readout process, these charges are read out one row at a time. The gate lines (Fig. 2) control the readout. When the gate line voltage is set high, all transistors of the row conduct the pixel charges collected since the last readout cycle to the adjacent data lines. Charge amplifiers and multiplexer integrated in the detector then read out the signal for that row; thus, every data line transports the information for one pixel during the readout of one row, and there is one data line for every line of the detector. This readout pro-
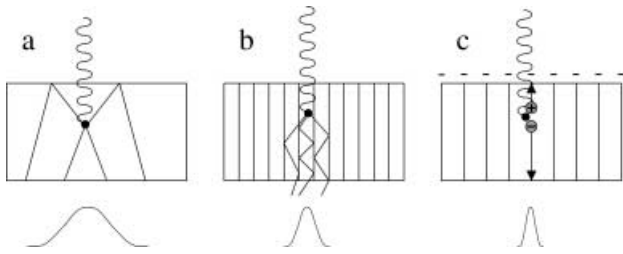
Fig. 3a–c Light diffusion is a major factor in determining resolution. **a** Screen-film system, storage phosphor radiography and nonstructured indirect conversion systems: lateral diffusion of light photons is important, or a loss of sharpness results. **b** Indirect detector with structured scintillator layer: lateral diffusion is limited by needle or column-like structure of the scintillator. **c** With direct conversion of X-rays into electrical charges, virtually no lateral diffusion is observed, since charges move perpendicularly to the surface of the selenium layer. (Modified from [32])

cess is also called active matrix readout, in opposition to the storage phosphor systems and the Thoravision detector, where no active readout elements are integrated within the detector. The entire readout process is very fast, allowing to construct in the future real-time X-ray detectors for fluoroscopic and angiographic applications [7].

Indirect conversion detectors with TFT panel

Indirect conversion detectors consist from top to bottom of a scintillator layer, converting X-ray photons into visible light, an amorphous silicon photodiode layer, transforming visible light into electric charges and, finally, the TFT-readout circuitry transforming the charge into digital values. One potential inconvenience of transforming X-rays into visible light is that lateral diffusion of the light reduces sharpness and spatial resolution of the image (Fig. 3). To overcome this problem, some indirect conversion detectors use structured scintillators, consisting of cesium iodide crystals that are grown perpendicularly to the detector surface. The single crystals have a diameter of approximately 5–10 μm and significantly reduce lateral diffusion of the scintillator light, much like fiber optics. The reduction of light diffusion in turn allows to use thicker scintillator layers, thus increasing the detective quantum efficiency of the detector system.

Direct conversion detectors with TFT panel

Direct conversion detectors consist of an X-ray photoconductor layer (typically amorphous selenium), transforming X-ray photons into electric charges, grown directly on top of the TFT charge collector and readout layer. Amorphous selenium is used because of its excellent X-ray detection properties and a very high spatial

resolution [8, 9, 10, 11]. As in the Thoravision system, an electric field is applied across the selenium layer before exposition. X-ray exposition generates electrons and holes within the selenium layer. Due to the electric field, these charges migrate nearly perpendicularly to both surfaces of the selenium layer, without much lateral diffusion. At the bottom of the layer, charges are drawn to the charge-collection electrodes, where they are stored until readout. During the readout, the charge of the capacitors of every row is conducted by the transistors to the amplifiers.

Important properties of flat-panel detectors

Detector size and weight

To be compliant with all standard examinations in radiology, the detector needs to have a dimension of at least 43×35 cm. Larger views (i.e. full spine or full leg) can be created by electronically pasting more than one single image [12, 13]. If the detector is to be retrofitted in existing radiography installations, i.e. Bucky tables, its thickness is limited by the mechanical constraints of the table. All FPD existing presently are too heavy (and too expensive) to be used to take radiographs in the intensive care unit (ICU); thus, FPD cannot replace SFS and SPR in the ICU. It is, however, imaginable to have lighter FPD with an integrated storage medium, storing radiographs taken in the ICU, the raw images being transmitted to an image processing workstation when a certain number of radiographs have been taken.

Pixel size and spatial resolution

Since the readout mechanism is integrated in the detector system, the spatial resolution of the detector is determined a priori by its construction (and not a posteriori by the readout process as is the case with SPR). Spatial resolution is thus determined by the size of the elementary units of the detector, i.e. the size of the capacitor electrode and the TFT, and the distance or pitch between these elements. Typically, spatial resolution is between 2.5 and 3.6 lp/mm (corresponding to a size of the detector units of 139–200 μm and matrix sizes in the range of 2000×2000 up to 3000×3000 elements; see Table 2), but special applications, such as mammography or even rheumatological examinations, need higher resolutions with detector element sizes equal or less than 100 μm.

The size and distance of the detector elements indicates only the maximum spatial resolution of the system. The effective spatial resolution of the detector is usually somewhat lower because of the light scattering (systems working with scintillators) and/or lateral diffusion of charges observed with most systems. With direct detec-

Table 2 Characteristics of some flat-panel detector systems currently available

Canon CXDI-11

Scintillator 200-μm thick (terbium-doped gadolinium dioxide sulphide)
Photodiode (semiconductor-type photoelectric converter made from hydrogenated amorphous silicon a-Si:H)
TFT (made from hydrogenated amorphous silicon)
Each pixel consists of an a-SI TFT and a metal insulator semiconductorphotoelectric converter.
Indirect conversion of X-rays
2688×2688 pixels
43×43 cm
160-μm pixel size
4096 grey-scale image (14-bit resolution)

Trixell (Philips, Siemens, Thomson)

Structured scintillator 550-μm thickness (thallium doped cesium-iodide, CsI:Tl)
Photodiode (amorphous silicon)
Switching diode
Indirect conversion of X-rays
3000×3000 pixels
43×43 cm
143-μm pixel size
4096 grey-scale image (14-bit resolution)

General Electric Medical Systems (Milwaukee, Wis.)

Scintillator (cesium-iodide)
Photodiode (amorphous silicon)+TFT
Indirect conversion of X-rays
2048×2048 pixels
41×41 cm
200-μm pixel size

DirectRay (Hologic, Kodak, Rochester, N.Y.)

Amorphous selenium 500-μm (photoelectric layer)
Capacitor+amorphous silicon TFT
Direct conversion of X-rays
2560×3072 pixels
35×43 cm
139-μm pixel size

tor systems, the effective spatial resolution is very near to the maximum spatial resolution.

Fill factor

The *geometric* fill factor of flat-panel detectors with TFT readout mechanism is the fraction of the pixel area occupied by the pixel storage capacitor. The smaller the pixel is, the lower the geometric fill factor will be since there is a minimum size for the pixel transistor; thus, a higher resolution means a lower geometric fill factor and a lowered DQE.

The *collection* fill factor corresponds to the fraction of the pixel area of the detector from which light or charge generated by the X-ray quantas will be collected by the pixel (i.e. optically sensitive area of the photodiode, or charge collection area).

Efficiency of the detector

The efficiency of a detector system in detecting incident X-ray quantas is described by the DQE. This is a widely accepted descriptor of the performance of X-ray detector systems. The DQE is dependent on many factors, among them the detector itself, the quality of the radiation (kilovoltage), the dose itself and the spatial frequency to be detected. The DQE is always lower than unity, since the detection of the X-ray photons is always incomplete. For radiography applications with a high dose delivered to the detector, both direct and indirect flat-panel detectors have higher DQEs than film-screen systems or computed radiography systems. For applications operating with lower dose delivered to the detector, such as fluoroscopy, there is a substantial decrease in DQE [14]. There is also a decrease in DQE observed for higher spatial frequencies.

Added noise

The main sources of added noise in well-designed flat-panel detector systems are the thermal noise of the TFT itself, the preamplifier noise and the line noise. Antonuk et al. [14] describe various approaches to reduce the noise of TFT-based flat-panel detector systems, such as

Table 3 Phantom studies with flat-panel detectors (FPD). *ROC* receiver operating characteristics; *AUC* area under curve of the ROC experiment; *SPR* storage phosphor radiography system; *SFS* screen-film systems

Reference	Type of study	Detector type	Results
[16]	Simulated bone erosions in a hand phantom (holes of different sizes drilled in polymethyl acrylate panels which were superimposed on a hand phantom, i.e. low-contrast detection task). ROC study with four observers, 7200 observations	Prototype of the Trixell/Siemens detector	Compared with a FSS (Lanex regular screen and T-MAT Plus DG film; Kodak), better diagnostic performance of the FPD with same dose ($p<0.05$). No significant difference for FPD images obtained with reduced dose (30% and 50% dose reduction)
[17]	CDRAD contrast detail phantom (four alternative forced choice experiments). Test signals: holes of different diameters and depths up to 2.0 mm (i.e. high and low contrast with different diameters)	GE Revolution XQ/I	Compared to both Insight regular and Insight HC the FPD shows better detection of test signals. Test signals with low contrast are more frequently detected with the FPD even when the dose is reduced by 20%
[18]	CDRAD 2.0 phantom. Comparison of FPD (at 400, 600 and 800 equivalent speed), SPR (AC-3 with ST-V plates, at approximately equivalent 200 speed) and a Kodak 160 speed SFS. Digital images evaluated at a Sectra Workstation with different monitors. Four observers, each reading three images for each setting. Comparison of image quality figure. Determination of skin entrance doses	Trixell/Philips	FPD has equal image quality at less than half the dose when compared with SPR, and at approximately at one-fifth the dose when compared with SFS
[19]	Comparison of images obtained with FPD and a 400-speed SFS (Insight VHC, Kodak). Anthropomorphic chest phantom with simulated lung structure and superimposed nodular, micronodular, linear and reticular patterns. 480 observation fields for each modality. Four observers. ROC analysis	Reduced size (15×15 cm) prototype of Trixell/Siemens detector	For dose equivalent images, FPD performs better than SFR for linear structures and micronodular opacities, whereas no significant difference is detected for nodules and reticular patterns. No significant difference between full-dose SFS images and half-dose FPD images was found
[20]	Detection of foreign bodies (glass with and without lead, bone fragments, aluminium, iron, copper, gravel, graphite) of different sizes superimposed to fresh porcine meat. Comparison of FPD (no spatial frequency processing) at different simulated speed classes to a 400-speed SFS (Lanex Regular/T-MAT Plus DG film, Kodak). 400 observation fields per modality. Four observers. One ROC curve for all foreign bodies, no separate evaluation	Reduced size (15×15 cm) prototype of Trixell/Siemens detector	At a simulated speed of 400, the FPD system performs significantly better than the SFS. At a simulated speed of 800 and 1200, no significant difference between FPD and the 400-speed SFS was detected. At a simulated speed of 1600, the FPD system was significantly inferior to the 400-speed SFS
[21]	Detection of cortical bone defects and fractures. 232 tubular deer-bones with mechanically induced fractures on 110 of them and cortical bone defects on 112 of them. Comparative images obtained with identical exposure with the FPD and 400-speed SFS (Lanex Regular/T-MAT Plus DG film, Kodak) and reduced exposure for the FPD. Four observers	Reduced size (15×15 cm) prototype of Trixell/Siemens detector	No significant difference could be found for the detection of cortical defects and fractures, even with dose reduced images obtained with the FPD detector (at one-half, two-thirds and one-fifth of the dose). Very high AUC; thus small differences may not have been detected

the introduction of multiple columns of nonresponsive pixels in order to reduce external line noise.

Size of the digital image file

Image size with FPD is dependent on grey-scale resolution and size of the pixel matrix. Size of a single image is usually between 8 and 18 Mbyte. Since this is a very huge amount of data, some manufacturers crop the resulting image to the region of interest (i.e. segmentation of the image and elimination of useless image borders) and thus transmit only a reduced amount of data to the storage system. This cropping of the image is much simplified with systems where the X-ray tube with its collimation and the FPD are integrated within one system, because the image processing software in these integrated systems can be made aware of the collimation borders.

Table 4 Clinical studies with FPD. *PA* posteroanterior; *ROC* receiver operating characteristics; *SFS* screen-film system; *FPD* flat-panel detector; *SPR* storage phosphor radiography

Reference	Type of study	Detector type	Results
[28]	Comparison of SFS (Ultravision L, Sterling, 400-speed class) and FPD (simulated 400-speed class) PA and lateral chest images obtained from 80 patients with known chest abnormalities (CT was gold standard). ROC analysis	Canon CXDI-11	FPD performed significantly better ($p<0.5$) for mediastinal abnormabilites. FPD performed slightly but not significantly better for pleural, pulmonary and spinal abnormabilities, pulmonary nodules, mediastinal masses, interstitial disease, pleural effusions and foreign bodies. FPD performed slightly but not significantly worse for chest wall abnormabilities and calcifications
[29]	Comparative and dose-adjusted (i.e. simulated 400-speed with the digital system) PA and lateral chest radiographs obtained within 24 h from 115 patients. SFS=Ultravision L, UV 400. Three observers rated the overall appearance and compared the depiction of different anatomical regions	Canon CXDI-11	FPD is preferred for 9 of 11 criteria (trachea, carina, main bronchi, costopleural border, azygoesophageal recess, paraspinal stripe, mediastinum, soft tissue, bones) for PA images (no significant difference for two other criteria: peripheral pulmonary vasculature and hilum). FPD is preferred in the lateral view for retrosternal and retrocardiac pulmonary structures, whereas SFS is preferred for depiction of trachea, costodiaphragmatic recess and hilum; no difference found for posterior cardiac border, fissures, thoracic spine and sternum
[30]	Comparative FPD and SPR (ST-V plates) radiographs obtained with the same settings. 64 consecutive patients. Comparison of the quality of depiction of 11 anatomical regions using a high-resolution grey-scale CRT. FPD images reduced by 20% in size to fit monitors. Four observers (two chest radiologists). No unsharp masking used	DirectRay	Preference for the FPD system for six regions (lung, hilum, minor fissure, rib, heart border, overall appearance). Preference for the SPR system in two regions (proximal airway and thoracic spine). No difference found for three regions (subdiaphragmatic and retrocardiac lung, azygoesophageal recess). Chest radiologists have an even more marked preference for FPD images
[22]	FPD vs 400 speed SFS (Lanex Regular/T-MAT Plus DG film, Kodak). 120 comparative images from different osseous regions. FPD images additionally with reduced dose (50% reduction for all images and 75% reduction for 40 images). Six observers. Observer preference study. No spatial frequency processing of FPD images	Reduced size (15×15 cm) prototype of Trixell/Siemens detector	No significant preference for criteria overexposure and under exposure. Significant preference of the FPD detector for soft tissue presentation at full and half dose. Significant preference for the SFS for contrast resolution when compared with FPD at half dose and 25% dose. Significant preference for the SFS for spatial resolution when compared to the FPD system at 25% dose
[24]	Comparison of skeletal images obtained at full dose with SFS and FPD and at one-half and one-fourth dose with FPD. Images of 30 consecutive patients. Eight observers, subjective image quality criteria	Full size (43×43 cm) commercial Trixell/Siemens detector	Full-dose FPD images performed better than SFSc images for all criteria except overexposure. FPD images at 50% dose showed a small but not significant inferiority to SFS images. FPD at 25% dose were significantly inferior to full-dose SFS regarding bone ortex and trabecula, contrast and overall impression
[23]	Comparison of PA thoracic images of 15 consecutive patients obtained with a SFS at full dose and with the FPD system at full and 50% dose. Eight observers, six subjective image quality criteria	Full size (43×43 cm) commercial Trixell/Siemens detector	With full-dose images, a statistically significant difference was found for five of six quality criteria. With half-dose FPD images, a statistically significant difference was found for only one criterion

Dead pixels

All flat-panel detector systems have numerous non-working or dead pixels. These pixels do not contribute to the formation of the image. The user usually is not aware of the presence of dead pixels, since the information that should be contributed to the image is interpolated during image processing; however, the higher the number of dead pixels is, the lower the overall or regional image quality will be. There is a difference between single, iso-lated dead pixels, dead pixels with one or more neighbours and complete dead rows of pixels. Floyd et al. [15] found 0.1% of dead pixels for a flat-panel detector.

Studies conducted with flat-panel detectors

The results of published phantom studies with clinical relevance and clinical studies with FPD are summarised in Tables 3 and 4. Several studies show a better image

quality at isodose and a comparable quality at a lower dose when FPD are compared with screen-film systems in phantom studies [16, 17, 18, 19, 20, 21] and in clinical studies [22, 23, 24]. This is due to the higher DQE of the FPD systems [14, 25, 26, 27]. Dose reduction of approximately 50% seems to be possible with FPD for both skeletal and thoracic radiographs [22, 23, 24]. Observer preference studies show an equivalence or preference for FPD chest images when compared with SFS [28, 29] and SPR [30]. Most studies concern the Trixell detector in a system integration by Siemens (Erlangen, Germany) and where conducted by the working group of the University of Regensburg (Strotzer et al.). Results have to be confirmed by other working groups. Only few results are available with other FPD systems. No comparison has been made between the competing FPD systems. May et al. [6] have shown that a substantial reduction in examination time can be obtained with FPD systems as compared with SFS (total examination time reduced by 50% when using FPD without RIS connection and by 68% when using FPD connected to RIS). No comparison was made in this study with the examination times of SPR.

Conclusion

Flat-panel detectors seem to be a very promising technology for high-quality digital radiography. They offer most advantages of storage phosphor radiography and seem to have advantages in handling and image quality.

Only few clinical studies have been published thus far. Image quality of FPD seems to be equal to or better than that of SFS and standard SPR. The results obtained must be confirmed by other working groups. There seems to be a potential for dose reduction without compromising image quality. Image quality may be further enhanced by reduction of system noise. While there are important differences in the construction of the FPD systems on the market, no comparison between the competing FPD has been published.

The FPD offers a large ergonomic advantage over cassette-based SFS and SPR. There is no need for cassette handling, since the readout mechanism is integrated in the detector. Time to obtain a first readout showing if the positioning was correct is much shorter (only a few seconds) than with SFS and with SPR. Rooms equipped with FDP thus allow a very high patient throughput, higher than that of SFS rooms, comparable to that of rooms equipped with Thoravision, and probably somewhat higher than that of rooms equipped with SPR-plate-changing machines.

One drawback of FPD is that usually the complete installation of an FPD in a general-purpose room necessitates the acquisition of two detector units, one to be installed in a Bucky table, the other in an upright cassette holder. Some manufacturers propose integrated universal systems which are built like a C-arm. This kind of installation allows use of one detector unit for all kinds of radiographs. At least in Germany, the acceptance of these C-arm-like systems seems to be low.

The market price of FPD is high at the time of writing. One of the main obstacles on the way to competitive market prices will be the fact that for every place where radiographs are to be taken, an active and expensive detector system is needed. This contrasts with SPR where only (much less expensive) imaging plates are needed, whereas the active plate reader is needed only once. With SPR, no downtimes of X-ray rooms should be expected, since defective imaging plates will be replaced with other plates in stock, and there are usually at least two readers in one institution. With FPD, if one detector is defective, a downtime for the X-ray room must be expected, since usually there will not be a spare detector unit and installation of the detector is complicated. Technological advances, such as the introduction of needle-crystalline SPR imaging plates [31], will further enhance the quality of SPR radiography.

While CCD detector systems seem to be only a transitional technology, the battle for market share will be that between SPR and flat-panel detectors. The FPDs are just at the beginning of their development, whereas SPR has already reached a certain degree of maturity. The idea of having one type of detector perfectly fitting every application (high-quality radiography, transportable detectors for ICU and detectors for fluoroscopic applications) is probably, at least for the next few years, too optimistic.

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