In conventional radiography, the large differences in absorption between the lungs and the mediastinum prevent optimal visualization of both regions in one image.

The first system for digital projection radiography was introduced by Fuji in the early 1980’s under the name ‘Computed Radiography’. However, for more than a decade the high initial costs prevented widespread practical application. In addition, unlike CT and MR, digital radiography did not provide additional diagnostic information over and above that of conventional radiography.

It is only in the past few years that digital radiography systems have become commercially available that can be expected to take better advantage of the opportunities of the digital technique, and to be significantly superior to analogue systems.

Disadvantages of screen-film radiography

In conventional screen-film radiography, the film has a three-fold function as the medium for image acquisition, presentation and storage. This unavoidably leads to organizational problems. In digital radiography, these functions are separated into different, independent steps. This has the advantage that each can be individually optimized.

The most important disadvantage of conventional screen-film systems is their limited dynamic range. In chest radiography, in particular, this means that the large differences in absorption between the lungs and the mediastinum (signal range) preclude optimal visualization of all regions in one image. The dynamic range and contrast are in inverse proportion to each other. A wide dynamic range provides good coverage of the lungs and mediastinum, but leads to reduced contrast. On the other hand, films with a high contrast do not provide adequate visualization of the mediastinal structures. The tolerance with respect to varying exposure levels is rather low. This is particularly evident when it is necessary to work without automatic exposure control.

Advantages of digital radiography

Digital detectors offer a much wider dynamic range than screen-film combinations. As a result, digital systems have a higher tolerance with respect to exposure variations, and allow for a better display of the whole signal range from minimum to maximum X-ray absorption. Digital systems also offer the possibility of image processing, which has a decisive influence on image quality and, as a result, on the diagnostic performance of digital radiographs [14]. This can improve the detectability of pathological structures, but can also reduce it if the parameters are incorrectly selected.

Detector systems in chest radiography

The detector system determines the diagnostic information actually acquired. In addition to conventional screen-film combinations, conventional radiography also offers asymmetrical screen-film combinations and the AMBER technique. In digital detectors there is a choice between storage phosphor screens and selenium.

Physical parameters

The physical parameters used to characterize detectors are the detective quantum efficiency (DQE), dynamic range and modulation transfer function (MTF). In addition, digital systems are also defined by the image matrix and pixel size, which can affect the spatial resolution, image noise and quantity of data per image.

Detective quantum efficiency (DQE)

The DQE describes the efficiency of a detector, i.e. the percentage of quanta for a given dose that actually contribute to the image. It is a function of dose and spatial frequency and is, by definition, affected by the various noise...
components of the system.

Dynamic range
The dynamic range of a detector (or the latitude of a conventional film) is the range from minimum to maximum radiation intensity that can be displayed as differences in signal intensity (or as density differences in conventional film).

Modulation transfer function (MTF)
The MTF describes how the contrast of image components is transmitted as a function of their size or, strictly speaking, their spatial frequency. The visual limit of spatial resolution is related to the lines of a high-contrast lead grid that can still be completely distinguished. It is expressed in line pairs per millimetre (lp/mm).

Contrast resolution
The concept of contrast resolution is related to the minimum or ‘threshold’ contrast. It is the smallest detectable contrast for a given detail size that can be shown by the imaging system with different intensity (density) over the whole dynamic range. The threshold contrast is a measure for imaging of low-contrast structures, and is largely determined by the DQE of the detector.

Typical values of the above-mentioned physical parameters for conventional screen-film systems and for digital detectors are shown in Table 1.

Screen-film systems
Conventional systems
The conventional screen-film system has a moderately good detective quantum efficiency (20–30 % at 60 keV) and a similarly good MTF for frequencies above 3 lp/mm. The strength of screen-film combinations lies in their high nominal spatial resolution (>3 lp/mm) and (depending on the film type) the high contrast resolution at optimum exposure.

In applications with a high component of scattered radiation, the spatial resolution is not so much determined by the threshold frequency as, and to a much greater extent, by the contrast and noise. The ‘real’ spatial resolution visible in the film is therefore, in part, significantly less than the maximum value indicated by the film manufacturer. Due to the limited dynamic range (about 1:30, depending on film type) the exposure must be as exact as possible. As a consequence, in exposures without automatic exposure control (e.g. lung exposures at the bedside), some incorrect exposures are unavoidable. The gradation curve of the film is sigmoid, and defines the relationship between dynamic range and contrast. Films with a relatively flat curve (L-film, C-film, asymmetric screen-film combinations) have a wide dynamic range but, at least per region, they have a limited contrast that can limit their diagnostic value in the lung parenchyma.

Asymmetric combinations
The newer asymmetric screen-film combinations are aimed at simultaneous optimization of both the contrast and the dynamic range, in order to meet the specific requirements of chest radiography. They consist of different front and back screens, and accordingly, different front and back emulsion layers on the film. A special anticross-over layer prevents interaction between the front and back screen-film combinations. The front screen-film combination makes the primary contribution to visualization of the lung regions, while the back screen-film combination serves for visualizing the mediastinum and retro-diaphragmatic region.

The radiation exposure required for asymmetric screen-film combinations is about 40 % less than that of a conventional screen-film combination with a sensitivity (speed) rating of 250. The spatial resolution depends on density [5] while the dynamic range, in the sense of the useful exposure range, is no greater than that of standard screen-film combinations.

Computed radiography (CR)
Storage phosphor screens are the most widely used detectors for digital radiography. Their DQE is less than that of a screen-film combination, while the MTF is moderate and depends on the type of phosphor screen [3, 9].

| Table 1. The physical characteristics of various detector systems. |
|---|---|---|---|---|---|
| | DQE (60 kV) | Dynamic range | Matrix size | Pixel size | Limit of spatial resolution (Nyquist frequency) |
| **Conv. film** | 20–30 % | 1:30 | - | - | 5 lp/mm |
| **Storage screens** | 30 % | 1:40000 | >1:30000 | 2k/4k | 0.2–0.1 mm |
| **Selenium** | 60 % | 2.7 lp/mm | 0.2 mm | 2.5–5 lp/mm |
As in all digital systems, the resolution of the storage phosphor screen is determined by the pixel size. In most systems the format needed for chest radiography (35 x 35 cm or 35 x 43 cm) has a matrix of approximately 2000 x 2000 pixels, with a pixel size of 0.2 mm. The resolution of 2.5 lp/mm is below that of conventional radiography. Newer systems offer a 4000 pixel matrix (pixel size 0.1 mm), resulting in a resolution comparable to that of conventional radiography, even with the larger cassette formats required for chest radiography.

A significant advantage of the storage phosphor screens is their wide dynamic range (> 1:10 000), with a linear relationship between radiation exposure and pixel value over a wide exposure range (0.01–100 mR) [18]. This largely eliminates the chance of exposures that cannot be assessed due to excessively high or low film density. However, in the case of underexposure, the quality and diagnostic value of the images is clearly reduced by the increase in image noise.

There are several generations of storage phosphor screens, which differ with respect to MTF and DQE. In particular, the composition of the newer ST-V screens has been modified with respect to that of the older ST-III N screens, in order to achieve a faster read-out and a higher detective quantum efficiency. Optimum read-out requires an increase in laser sensitivity and a change in the wavelength of the stimulating laser light (from 633 to 680 nm). These conditions are only met by the latest read-out systems (AC III, FCR 9000 etc.).

**Selenium radiography**

The use of selenium as a detector material has long been known from xeroradiography. In digital radiography, however, there is at present only one system that uses a selenium detector. This is a dedicated chest unit (Philips Thoravision). The detective quantum efficiency of the selenium system is superior to that of any other detector system (Fig. 1). Due to the higher quantum absorption and low system noise, the image noise is also significantly lower than that of other detectors [10]. As in all digital systems, the resolution is determined by the pixel size (0.2 mm; 2166 x 2448 matrix) and is nominally below that of film. The dynamic range is extremely wide (1:10 000), with a linear relationship between dose and signal over a wide exposure range [13].

The direct conversion of the absorbed quanta into an electrical charge, and from there into a digital signal, reduces the system noise by eliminating additional sources of noise during the read-out process [4]. An electrical charge applied to the selenium layer ensures that the majority of the liberated electrons are transported through the layer in a straight line perpendicular to the field gradient, greatly reducing diffusion. This occurs independently of whether the electrons arise at the surface of the layer, or deep within it. In contrast, the diffusion of emitted light in storage phosphor screens or conventional intensifying screens becomes greater when an X-ray quantum is absorbed deep within the luminescent layer, rather than at the surface. In the selenium detector, the greatly reduced diffusion results in a reduction in structure noise. In addition, the selenium...
layer offers the possibility of using thicker layers, allowing the detective quantum efficiency to be increased without affecting the image resolution.

In order to reduce scattered radiation, the Thoravision has an air gap of 15 cm between the surface of the detector drum and the front of the housing. The air gap is permanent, but an additional stationary grid can be fitted as an option. The grid has 60 lamellae/cm and a ratio of 12:1.

**Digital image processing**

The possibility of digital image processing is an important advantage of all digital systems. In chest examinations, the aim is to improve the conspicuity of pathological structures. This is achieved by increasing the MTF for fine image details by increasing the local contrast, by edge enhancement to show the outlines of the objects, and by compression of the dynamic range. The latter yields a more ‘transparent’ display of the mediastinum by reducing the density differences between the mediastinum and the lungs.

The applications of digital image processing can be divided into global contrast manipulation by changing the gradation, and local contrast manipulation by frequency modulation. Newer systems also have the option of digital noise reduction.

**Signal normalization**

In digital radiography, there is an almost linear relationship between dose and signal over a wide range of exposures (wide dynamic range). Depending on the radiation exposure and the patient’s anatomy (large or small differences in absorption), only a small part of the total dynamic range is used. For this reason, only the relevant signal range is converted into digital values. This ‘signal normalization’ forms the basis for optimization of the optical density, independently of the radiation exposure. The starting point is a histogram analysis of the exposed area. A prerequisite for correct signal normalization is correct definition of the exposed area by detection of the image collimation borders. This is much more demanding in skeletal than in chest radiography. In the selenium system this is not necessary, as the collimator settings are detected directly (dedicated chest radiography system).

In the Thoravision system, the dominants in the lung fields are also established from the signal distribution in a histogram analysis. These are arranged according to a preselected look-up table, so that the lung fields are allocated a defined mean optical density, e.g. 1.6.

**Unsharp mask filtering**

Unsharp mask filtering is the simplest and most widely used frequency filtering procedure. Unlike gradation adaptation, which results in a global change in contrast, unsharp mask filtering provides a local change in image contrast, depending on the selected filter parameters and the size and the sharpness of the outlines of the structures in the image. The character of the image can be changed considerably. Image details can be emphasized or suppressed as required.

In this procedure, a low-pass unsharp mask image \((B_m)\) is derived from the original image \((A)\). A high-pass filtered ‘edge’ image \((A-B_m)\) is then obtained by subtraction of the unsharp mask from the original image. The final filtered image \(A^*\) is created by weighted addition of the edge image to the original image according to the formula \(A^* = A + f (A-B_m)\), where \(f\) is the weighting factor.

The unsharp mask \(B_m\) is created by replacing every pixel with the average value of the surrounding pixels. The larger the area to which this procedure is applied, the more unsharp the mask image will be. Different weighting factors used for averaging in this area \((m \times m\) pixel) are available. The value of \(m\) for the area is referred to as the kernel size.

The weighting factor \(f\) determines the degree of filtration. With a low weighting factor, the appearance of the filtered image will be close to that of the original image, while with a high weighting factor the appearance will approach that of the edge image. Image noise increases with an increased weighting factor. Also, the smaller the kernel, the greater the noise will be. Nonlinear weighting factors allow different degrees of filtration to be applied to different regions of the image. For example, in a chest...
The diagnostic performance of the different imaging systems was assessed by comparing the results of phantom studies and clinical studies.

Examination it appears useful to reduce the degree of filtration in the regions with high absorption, the image noise is more evident there.

Modified forms of unsharp mask filtering are applied in the algorithms used with the latest generation of storage phosphor screen systems (FCR 9000, 9501, AC III). These are referred to as DRC (dynamic range compression) in Fuji systems, and as DRR (dynamic range reduction) in Philips systems. Both systems are aimed at reducing the dynamic range.

Image processing in selenium radiography
Selenium radiographs are first processed using unsharp mask filtering with a very small kernel (3 pixels = 0.6 mm). This increases the MTF for high frequencies, resulting in improved image sharpness. The filtering is performed with a nonlinear weighting factor, to avoid excessive noise in regions with high absorption, such as the mediastinum.

Further image processing is based on a variant of unsharp mask filtering: filtering with a large kernel (3 cm per side). Each image is divided into two sub-images, i.e. the unsharp mask image \( B_m \) and the difference image \( A-B_m \). The unsharp mask image only contains very low spatial frequencies or, in other words, information on the distribution of dense and transparent image components, which is important for image gradation. The difference image contains all higher spatial frequencies, with information on anatomical details and the local image contrast. Addition of both images gives the original image \( A = B_m + (A-B_m) \).

The degree of filtering is determined by applying different nonlinear weighting factors \( \gamma \) or \( \beta \) to the gradation image and the difference image. These weighting factors correspond to the characteristic curves for gradation \( \gamma \) and contrast \( \beta \) known from conventional film. A \( \gamma \) corresponding to the gradation curve of a wide-latitude C-film and a \( \beta \) corresponding to the contrast curve of a high-contrast G-film have proved to be of advantage in chest examinations.

Thus, the image processing in digital radiography allows the contrast and density of the image to be optimized independently of each other, which is not possible with conventional screen-film systems.

Diagnostic performance
An important consideration for the acceptance of new imaging techniques is the way in which the physical parameters mentioned above are converted into image quality and diagnostic performance. A requirement for the deployment of digital systems in Germany, as laid down in the guidelines of the Bundesärztekammer (Federal Medical Council), is that these systems should have a performance at least as good as that of conventional systems at the same or at a lower dose. The reference standard is a conventional screen-film combination with a speed rating of 400, in accordance with the guidelines for chest examinations.

Several types of study can be used for determining the image quality and diagnostic performance. In the authors' opinion the results of these studies complement each other, as they investigate different aspects.

Phantom studies
These studies, which test the detectability of simulated pulmonary lesions against an anthropomorphic background, are ideally suited to comparing different imaging systems:

- identical exposures can be compared with each other, without the need to subject a human being to repeated doses of radiation
- extremely subtle lesions can be selected, providing a statistically valid demonstration of even the smallest differences in the diagnostic performance of the imaging systems.

Clinical studies
In clinical studies, radiation protection considerations severely restrict the comparison of different techniques. Consequently, definition of a reference standard frequently requires a disproportionate expenditure of time and effort, while the selection of the lesions determines the results of the study. On the other hand, clinical studies are able to show the influence of variations in scattered radiation due to differences in patient thickness, but are also able to include the influence of anatomical noise due to
superprojection of a multitude of pulmonary structures. Direct comparison of the image quality on the basis of subjective criteria is inferior to studies based on the detection of lesions.

The results of studies dealing with the diagnostic performance of the selenium detector in comparison with storage phosphor screen systems and conventional screen-film combinations are summarized below.

**Phantom studies**

The potential reduction in the radiation dose to the patient, due to the lower radiation exposure required for digital radiographs, was one of the main arguments for the application of digital detectors, particularly in the years immediately following their introduction. In practice, however, depending on the detective quantum efficiency of the available detectors, it appeared that reduction of dose increased the image noise to such an extent that, at least with the older types of storage phosphor screens, a reduction in dose could not be achieved without loss of image quality. Only storage phosphor screens of the newest generations (e.g. ST-V) and selenium detectors have a significantly higher detective quantum efficiency.

The results of a contrast/detail study show that there are considerable differences in dose requirement between the various digital detectors now on the market. These must be taken into account in the clinical routine.

Selenium requires the lowest dose for a given degree of detail detection. Type ST-V storage phosphor screens require a 1.6 x higher dose, while Type ST-III N storage phosphor screens need as much as 2.7 x higher dose [16]. As the absorption increases (mediastinum v. lungs; corpulent v. thin patients) the difference in dose requirement reduces, but is not eliminated. However, it should be emphasized that the dose-saving advantage of ST-V screens over the older ST-III N screens only applies when the screens are used with new read-out equipment (FCR 9000, AC III etc.). When used with older read-out equipment the ST-V screens show only a minimum advantage. An additional contrast/detail study also confirmed the superior contrast resolution of the selenium detector compared with conventional film (Tmat-G Film/Lanex medium screens, speed 250 [12].

The results of these abstract contrast/detail studies could be confirmed in phantom studies. These studies used an anthropomorphic background, and tested the detection of simulated nodules, pleural lines, spot shadows and micronodular opacities [15].

Selenium is superior to other techniques in the detection of low-contrast micronodular opacities and lines.

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**Fig. 2.** Comparison of the detectability of typical lesions in a thorax phantom with different imaging systems. The values shown represent the areas under Receiver Operating Characteristics (ROC) curves. The closer the value approaches 1, the greater the detectability of the lesion and the degree of certainty in the diagnosis.
Sections from images of a chest phantom containing a simulated round lesion, a line, and fine pulmonary opacities. The lesions were very subtle, in order to detect small differences in quality between the imaging systems. There is a clear improvement in contrast resolution in the digital techniques c, d compared with the conventional radiographs a, b.

Selenium radiographs show a significant superiority in the detection of low-contrast micronodular opacities when compared with conventional films as well as type ST-III (speed 250) and type ST-V (speed 400) storage phosphor screens. This advantage was observed with radiation exposures corresponding to storage phosphor screens of speed 200 and also speed 400. It could be attributed to the superior detective quantum efficiency of the selenium detector, and the resulting reduction in image noise. This was a particular advantage in the detection of low-contrast structures (Fig. 3 a-d). A similar phantom study, which also simulated interstitial changes with low-contrast micronodular...
opacities, confirmed that the diagnostic performance of selenium radiographs is at least equal to that of conventional film exposures [7].

The good detectability of linear structures shows that a pixel size of 0.2 mm provides adequate spatial resolution in selenium radiography. A study of our own showed that selenium provided similar diagnostic performance when compared with storage phosphor screens with an increased (4k) matrix and the same exposure.

In another phantom study, selenium was found to be superior to storage phosphor screens and conventional films in the detection of lines, even with a relative dose reduction by a factor 2 or 4 [11]. With respect to the detection of pulmonary nodules, no significant difference could be observed between the various detector systems. With respect to the detection of nodules in the highly absorbent region of the mediastinum, the best results were achieved with the asymmetric screen-film combination, storage phosphor screens, and the selenium detector with the antiscatter grid.

Three possible causes must be considered for the limited detectability of retrocardiac and retrodiaphragmatic nodules in selenium radiographs acquired without the antiscatter grid:

- the increased scattered radiation component in regions with high absorption

- the possibly higher sensitivity of selenium to scattered radiation, due to its low k-edge absorption

- the possibility that image data processing for the mediastinal region is not yet optimal [2, 4, 15].

Even when the dose was reduced to the equivalent of a 600 speed film (without grid) we still found selenium to be superior to conventional film (speed 400) for detecting lines and spot shadows. However, there was a significant reduction in the detection of nodules and micronodular opacities.

To summarize, it can be stated that selenium radiography with an exposure equivalent to that of 400 speed film has a diagnostic performance equal to that of conventional radiographs made with the same exposure, and is superior for certain types of lesion. Visualization of nodules in regions with high absorption appears to be capable of improvement by optimized image processing.

In our phantom study, we were unable to demonstrate any advantages of a higher image matrix (4k rather than 2k) in storage phosphor radiographs. In particular, increasing the image matrix did not significantly improve the detectability of linear structures. Linear structures correspond to the type of lesions where the detectability is most strongly determined by the spatial resolution.

A clinical study confirmed the superiority of selenium in the detection of interstitial changes.

Fig. 4. Comparison of clinical chest exposures with Thoravision (left) and conventional radiography (right).
Clinical studies

The number of clinical studies on this subject published to date is clearly smaller than the number of phantom studies. However, the results of the contrast/detail studies and the anthropomorphic phantom studies could be confirmed in these clinical studies. Even the comparison of the quality of the visualization of anatomical structures in selenium radiographs and conventional film exposures on the basis of subjective criteria indicated that the quality of the selenium detector is better or at least equal to that of conventional film [6]. The chest wall and lung fields (upper and lower fields including the retrocardiac region) were assessed as better visualized in selenium images, and visualization of the mediastinal structures was assessed as equal in both techniques.

In accordance with the results of the phantom studies, the superiority of the selenium detector, particularly in the detection of changes in the lung parenchyma, could be confirmed in a clinical study. This was based on the examination results in 100 patients with emphysema, bronchiectasis and interstitial parenchymal changes, confirmed by CT. Selenium radiographs were found to be superior to storage phosphor screens (ST-III N). Conventional films were not included in the study [1].

Another recently published clinical study of 104 patients with a wide range of mediastinal and pulmonary pathologies (with the standard defined by CT) described a statistically identical performance for selenium radiography and conventional film in the detection of pulmonary, interstitial, pleural and mediastinal lesions [8].

Conclusion

The medical guidelines for the deployment of digital systems require that the performance should be at least equal to that of conventional standard radiography. The results of both phantom studies and clinical studies confirm that the image quality of the selenium detector is adequate to meet these requirements.

Due to its high detective quantum efficiency and low image noise, the selenium detector provides...
excellent low-contrast resolution. In the region of the lung fields this is superior to both conventional film and storage phosphor screens. The superiority is particularly marked in the detection of low-contrast interstitial structures.

In high absorption areas such as the mediastinum, it appears that some improvement in performance could still be achieved by optimization of the image processing. However, this should not be at the expense of the image quality in the lung fields.

The results of both phantom studies and clinical studies confirm that the image quality of the selenium detector complies with the medical guidelines.

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