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Storage phosphor radiography

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Summary. Storage phosphor radiography is a digital technique that uses photo-stimulable phosphor screens to substitute for conventional screen-film combinations. While the technique is more than 15 years old, it is only recently that technological and economic aspects of these systems have become favourable enough to envisage a more widespread clinical application.

Key words: Storage phosphor plates – Digital radiography

Why go digital?

The most widely used detector and display medium for projection radiography remains radiographic film. The technique is a proven and reliable standard that has been optimised during the last 100 years. There are, however, a number of diagnostic as well as economic and ergonomic drawbacks of this technique.

Diagnostic reasons

Conventional radiographs suffer from
– large amounts of scattered radiation (e.g. in the mediastinum up to 90%)
– a wide variation of attenuation (e.g. between lungs and mediastinum- or bones and soft tissues) that often exceeds the dynamic range of the screen-film system and
– the need for a high film contrast to visualise low-contrast objects
– in studies with a lack of automatic exposure control, over- and underexposure are a frequent problem.

In general, image contrast and dynamic range are inversely related in conventional screen-film radiography: a high contrast comes with a narrow dynamic range while a wide dynamic range leads to a low contrast, at least in some portions of the image. Chest radiography is a good example of the efforts to overcome these difficulties. Wide-latitude films, asymmetric screen-film combinations, mediastinal port filters, scanned equalisation techniques (SER, Amber) and digital radiography were employed, but all approaches come with specific advantages and disadvantages [1, 2]. Wide-latitude films and asymmetric screen-film combinations increase the dynamic range of the detector, but are still vulnerable to exposure errors and fail to compensate for the inverse relation between contrast and latitude. Mediastinal filters and equalisation techniques on the other hand, reduce the range of absorption differences by increasing the exposure to the mediastinum, thus improving both, the visualisation of mediastinal structures and the contrast within lungs or mediastinum. However, these systems cannot be employed for bedside studies. Moreover, edge artifacts at mediastinal or diaphragmatic borders may occur.

Economic reasons

Radiographic film, chemical film processing and laser imagers are major contributing factors to the running costs of a medical imaging department. Reduction of film size or complete elimination of films as display medium will lead to marked savings [3].

Ergonomic reasons

Conventional radiographs can get lost (up to 20% of radiographs cannot be found in time). They can be viewed only at one place and the chemical processing of images takes time (no instant image). Retakes not only increase patient exposure but also decrease the number of examinations that can be performed by one technologist [4].

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Technique of storage phosphor radiography

Basic principle

Storage phosphor radiography was introduced in the early 1980s [5]. The system is cassette-based and is therefore compatible with existing X-ray equipment. Instead of a conventional screen/film combination, it uses a photostimulable phosphor screen as the image receptor.

The photostimulable phosphor screen is chemically similar to conventional intensifying screens and exhibits similar physical properties. Like in intensifying screens, incident radiation excites electrons. Most of these electrons release the absorbed energy immediately by emitting visible light, which is used for exposing the film in a conventional cassette. In storage phosphor screens, however, a substantial amount of the absorbed energy is captured after exposure by electrons that are trapped in a metastable energy level. The absorbed X-ray relief is stored as a 'latent image' until it is released by exposure to light of a longer wavelength than the characteristic emission of the phosphor (read-out process). When released from their metastable energy level, electrons emit light (photostimulated luminescence) that can be collected by a photomultiplier tube.

For the read-out process, a fine laser beam with a spot size of 50–200 mm is used. The photostimulated luminescence is proportional to the absorbed X-ray intensity. The output of the photomultiplier is logarithmically amplified and subsequently digitised by an analogue-digital converter with 8 to 14 bit resolution.

Spontaneous decay

If an exposed storage phosphor screen is not read out, spontaneous decay of the trapped energy occurs with degradation of the latent image. However, it takes more than 6 h to be able to detect visible differences as compared to an image that was read out immediately.

Dynamic range

Storage phosphor screens have a linear response (emitted luminescence) to incident radiation over an extremely wide dynamic range (> 1:40,000). However, for medical imaging, only a fraction of this dynamic range is actually used (i. e. 1:40–1:1,000) depending on the attenuation differences in the examined object ('object range'). There are various options of how to focus only on this smaller object range ('signal normalisation') As a result of the wide dynamic range, storage phosphor systems should ideally be no longer vulnerable to over- or underexposure (Fig. 1.).

Signal normalisation

One of the main advantages of digital radiography is the automatic density control that produces an optimised density in each image independent of exposure dose. The underlying process is called signal normalisation. The prerequisite for a full compensation of over- and underexposures as well as for all further image processing is the detection of the correct object range.

There are three basic ways to cope with the discrepancy between the extremely wide dynamic range, the much narrower object range and the variations in exposures [6]:

1. A preread scan with low resolution is performed. From this scan the object range is determined immediately and the main read-out process is adjusted in a way to only incorporate a fraction of the whole dynamic range, using 8 to 10 bit for digitisation.
2. The whole dynamic range is digitised, preferentially using a 10-to-14-bit resolution. These raw data are then further analysed and the actual object range is detected and displayed.
3. The technician specifies the exposure dose for a particular radiograph (this is derived from the system speed used for phototiming or estimated when non-phototimed radiographs are taken). The system then digitises only the fraction of the dynamic range that includes a typical object range with a safety margin for over- or underexposures by a factor 2. The raw data are then further analysed and the actual object range is detected and displayed.

Method 1 has been employed in the first installations but is no longer in use because failure of the detection of the object range will lead to an irretrievable loss of information. Method 2 is now mainly used by equipment manufacturer Fuji, while method 3 is applied by Agfa systems. Crucial to all three techniques is the software to detect the correct object range.

Data sampling. The first step is to sample the image data in a coarse fashion (small matrix size), either by using a preread process (method 1) or by including only every 8–20th pixel from the raw data set (methods 2 and 3). This is done in order to reduce the extensive computational effort.

Collimation detection. The next step is to determine the areas in which there is diagnostically relevant information. The correct detection of image collimation is a precondition for proper functioning of the normalisation process. This may not be problematic in chest radiology. However, it is very important in skeletal radiology because there is a multitude of variations in collimation settings, especially if more than one image is exposed per cassette. Therefore, advanced algorithms for automatic pattern recognition are required.

Software has been continuously improved but on-site adaptation of parameters is frequently required in order to take into account site-specific differences in the exposure of radiographs. In general, the more complex the

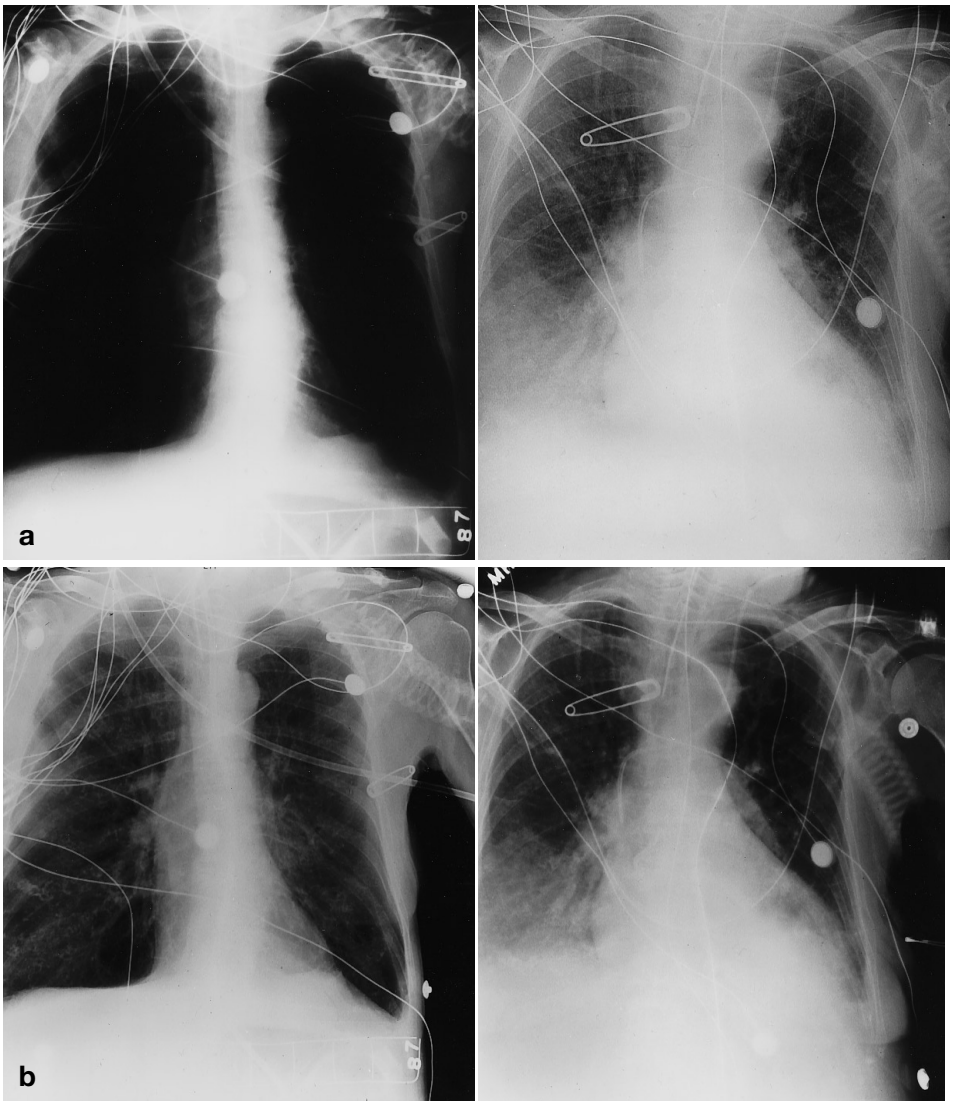


Fig. 1. Conventional **(a)** and digital **(b)** follow up chest radiographs at the bedside. The first images were taken while the patient suffered from severe cardiac failure, the second image was obtained after therapy and improvement of clinical symptoms. The simultaneously obtained digital radiographs show more consistent and optimised image density compared to the digital radiographs

collimation patterns, the more often the systems will fail.

Failure of the determination of collimation results in too dark or too light images similar to exposure errors in conventional radiography that can not always be retrieved by post processing.

Detection of the object range. After the areas with diagnostic information have been detected, a histogram of the digital signal in these areas is performed. Based on these histograms and information entered by the radiographer (type of radiograph, e.g, lateral chest), the object range is determined. These raw data are subjected to further data processing including adjustments of gradation curves and improvement of local image contrast (see below).

Spatial resolution

The spatial resolution of a digital radiography system is determined by the detector properties (the intrinsic modulation transfer function, MTF) and by the pixel size of the digital matrix [7]. The MTF depends on the thickness of the phosphor layer. Like in conventional screen/film systems, a thicker layer (e.g. ST screens, Fuji) improves quantum detection but decreases spatial resolution while a thin layer (e.g. HR screens, Fuji) requires up to 4 times more dose for a similar signal-to-noise ratio but results in an improved MTF and thus, higher spatial resolution.

Matrix size. For most applications, a matrix size of some 2000×2000 pixels ($2\ K \times 2\ K$) is employed. Depending on the cassette size, a pixel size of 0.1 mm (18×24 cm cassettes) to 0.2 mm (35×43 cm cassettes) results. Recently, a 4K- × –4 K matrix for 35- × –43-cm images was introduced that reduced the pixel size to 0.1 mm even for large phosphor screens (currently available only in a dedicated chest unit) [8]. The maximum spatial fre-

Table 1. Matrix size

Cassette size	Matrix pixel size	Pixel/mm	Max. spatial resolution (Nyquist-frequency)	
35 × 43 cm	1760 × 2010	0.2 mm	5	2.5 Lp/mm
35 × 35 cm	1760 × 1760	0.2 mm	5	2.5 Lp/mm
24 × 30 cm	1330 × 1670	0.15 mm	6. 7	3.3 Lp/mm
18 × 24 cm	1770 × 2370	0.1 mm	10	5.0 Lp/mm

quency that can be displayed in a digital system without aliasing artifacts is 2.5 lp/mm for a pixel size of 0.2 mm and 5 lp/mm for a pixel size of 0.1 mm.

The intrinsic MTF for most storage phosphor systems (Agfa screens, Fuji ST-1 to ST-V screens) resembles that of a 400-speed conventional system [9, 10]. When the matrix size is considered too, no major differences are found for spatial frequencies below 1.8 lp/mm, even if a 0.2-mm pixel size was employed. For higher spatial frequencies, however, there is an advantage for a 0.1-mm pixel size (similar performance as the conventional 400-speed system).

Image sharpness. In clinical practice, however, the subjective impression of image sharpness is determined mostly by those spatial frequencies, for which the modulation (MTF) is still high (above 20 %). Higher spatial frequencies may also contribute to the image but their influence on image perception is minimal [11].

As a result, both digital systems (even with 2 K matrix and 0.2-mm pixel size) and 400-speed conventional screen/film systems, lead to a similar impression of image sharpness. Digital image processing, however, may lead to an advantage of digital over conventional systems: processing can change the shape of the MTF in such a way that higher spatial frequencies contribute more to the image than in conventional radiography, thus leading to an increase in image sharpness (see below).

Quantum detection

Image noise on a radiograph may have a variety of causes, the most important of which is quantum noise. The higher the number of detected quanta at a given radiation exposure, the better the signal-to-noise ratio will be and, consequently also, the low-contrast resolution [9].

The detective quantum efficiency (DQE) can serve as a measure for the dose requirements of a detector system: the higher the DQE, the less dose is needed to obtain a desired signal-to-noise ratio or contrast resolution.

The DQE of storage phosphor systems used to be lower than that of optimally exposed conventional screen/film systems. New phosphor screens (e. g. ST-V, Fuji) now approach a DQE that comes close to conventional systems. Older screens (e. g. ST-III, Fuji) require more dose for similar signal-to-noise ratios.

Dose requirements

Different generations of storage phosphor screens are available that differ in quantum efficiency and noise characteristics. According to the manufacturer, recent ST-V screens (Fuji) have a 13 to 18 % lower noise level than older ST-III screens. It is important to know that the new screens develop their increased performance only in the newest hardware systems (AC III or FCR 9000 and related systems) that use lasers with adjusted sensitivity and altered wavelength (680 nm).

This increased signal-to-noise ratio may be used to reduce exposure doses or to take diagnostic advantage of the increased image quality. In a recent contrast-detail study that compared the detection of small, low-contrast objects overlied by scatterbodies comparable to the chest of slim and heavy individuals, we were able to calculate the factor by which the exposure dose could be decreased for ST-V screens while still having identical detection performance as with ST-III screens. Those factors varied from 1.7 for slim to 1.4 for heavy individuals.

Transferring these results to a clinical setting means that new types of storage phosphor screens require substantially less dose than older ones. Similar developments are under way with other manufacturers (Agfa, Kodak) and users are encouraged to inquire appropriate information about the quantum efficiency of the storage phosphor plates they use. This is especially important since digital radiography should provide at least equivalent diagnostic performance as conventional radiography with comparable exposure dose to the patients. A multitude of clinical and phantom studies have proven that ST-III plates need an exposure equivalent to at least a 250-speed system in order to achieve comparable performance to conventional state-of-the-art screen/film combinations [12–15] (Fig. 2.).

Documentation of patient exposure

In conventional radiography, the optical density of the radiographic film gives direct feedback whether the radiograph was over- or underexposed. In digital radiography, the automatic signal normalisation leads to consistent results with respect to density and contrast, independent of exposure. Underexposed images may suffer from increased image noise while overexposed images will have an improved signal-to-noise ratio. As a result, there may be a trend towards overexposing images. Thus, for reasons of radiation protection, there has to be some feedback of the actual dose the patient was exposed to [16].

On Fuji equipment, the so-called ‘sensitivity’ or ‘S-value’ is printed out on each digital hardcopy. It is a measure of the centre of the detected object range. Consequently, it can be used to monitor patient exposure. Originally, it was supposed to reflect the system speed, at which the exposure was taken. In fact, S-value and system speed are in the same range of numbers, but not identical. For practical purposes, it is enough to determine a ‘target value’ for each type of examination per-

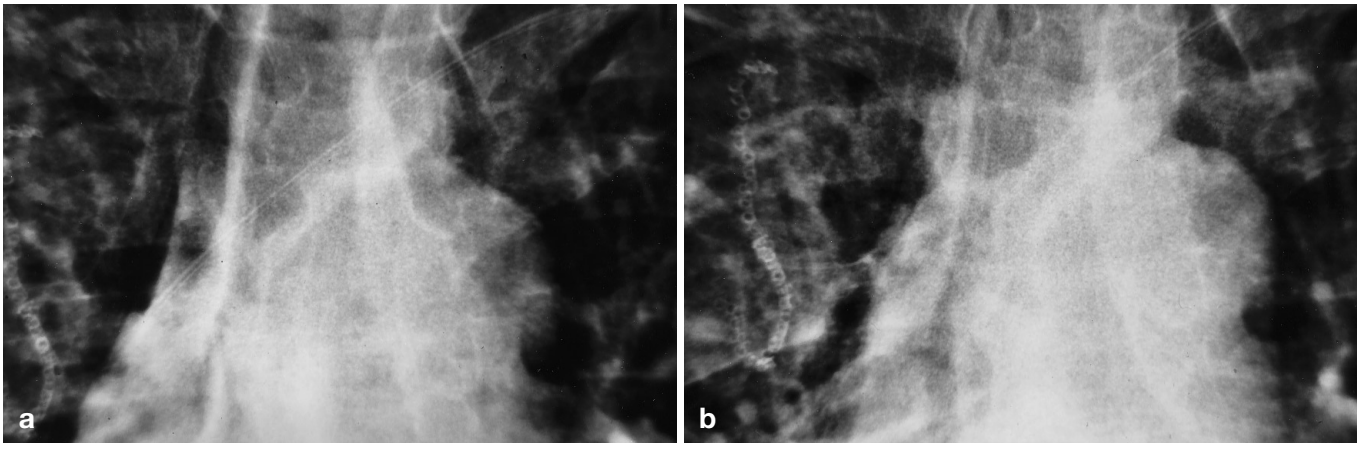


Fig. 2. AP chest radiograph **a** obtained with standard dose (250 speed) and **b** obtained with reduced dose catheter visualisation is markedly decreased with increased image noise

formed. The S-value should then remain constant over time and be as close as possible to this target value.

Agfa displays the deviation of the actual exposure (object range) relative to the expected object range at the chosen system speed. Thus, over- and underexposures can be easily traced.

Clinical experience

Techniques for image evaluation

The introduction of any new imaging technology requires the evaluation of its performance in relation to that of existing technologies. Physical imaging performance can be assessed by use of widely accepted physical measures such as detective quantum efficiency and MTF. However, for the users of such systems it is more important to know their performance relative to that of screen/film systems.

Image quality may be assessed on a subjective scale by visual assessment. Statistical methods are more elaborate, such as receiver operating characteristics (ROC) that test lesion detection and reader confidence [17]. A ROC analysis is independent of the individual decision threshold of each reader in determining the presence of a lesion [18]. It is also independent of lesion prevalence in the study group. While the results are most reliable, ROC has practical disadvantages: the process of data collection is time-consuming and there are high requirements in terms of standard of truth, number of readers and selection of pathology.

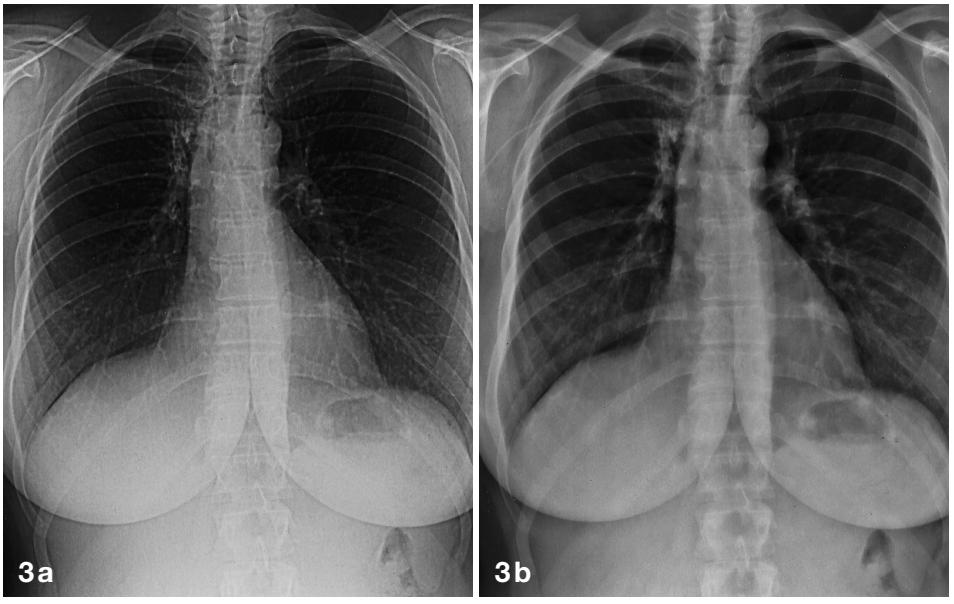
There is a multitude of studies published in the literature comparing the performance of storage phosphor radiography and conventional screen-film radiography. Clinical studies and studies with antropomorphic phantoms make different, yet valuable contributions to the comparison of the diagnostic performance of both technologies. Clinical studies are able to include a broad range of lesion types in terms of pathology and lesion conspicuity. Therefore they most often lead to similar results with either technique. Phantom studies, on the other hand, are more suitable to detect small differences in image quality by including mainly lesions of low conspicuity.

Chest radiology

Most studies were carried out in chest radiography. This may be due to the fact that the gold standard in clinical studies can be precisely defined by CT. Also for phantom studies, it is easier to simulate chest lesions rather than skeletal lesions.

It is agreed upon that the detection of pulmonary nodules, which is one of the most important clinical imaging tasks in chest radiography, is equivalent to analogue and digital techniques [12, 14, 19]. However, nodules that superproject over the mediastinum or retrocardiac area, are seen superiorly on digital radiographs [1, 20] (Fig. 3.). The reasons are the wide dynamic range of the storage phosphor plates and digital image processing that improves the conspicuity of these lesions by dynamic range compression and local contrast enhancement [13] (Fig. 4.).

In earlier studies, it used to be doubted whether storage phosphor radiography, with its limited matrix size and spatial resolution, could adequately visualise fine linear structures, such as pneumothorax lines, fine septal lines or interstitial disease [21]. Studies have shown that unprocessed digital images perform worse than conventional radiographs [22]. However, it could also be demonstrated that adequate image processing is able to improve lesion conspicuity, leading to an equivalent performance of digital systems and state-of-the-art conventional radiography [12, 15, 23, 24]. Adequate visualisation of lines requires a processing that is suited for MTF restoration at high spatial frequencies. With appropriate processing, the lower spatial resolution (2 K matrix size) could be fully compensated. Although recent improvements in detector technology allow for a 4 K matrix, we could not find major advantages with respect to diagnostic performance in a direct comparison of 2 K versus 4 K radiographs in a phantom study with optimised digital processing.



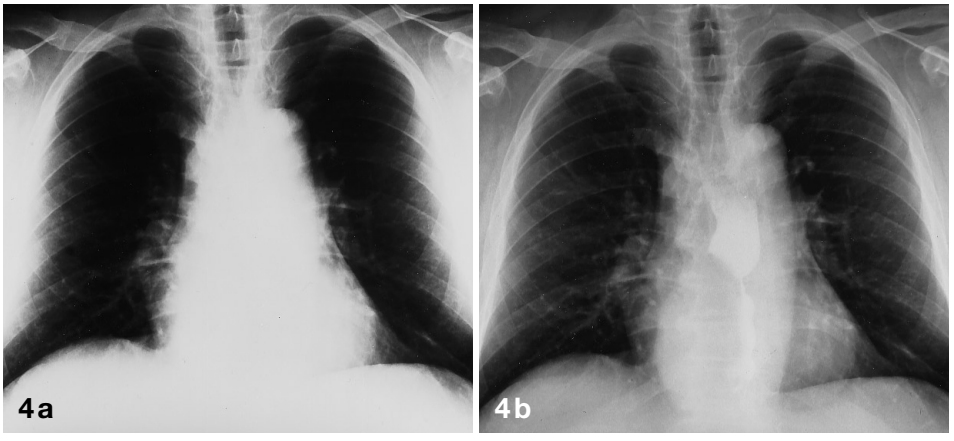


Fig. 3. Effect of dynamic range compression in a pa chest radiograph that shows small pulmonary nodules in the retrocardiac region

Fig. 4. Conventional and digital pa chest radiograph in a patient with oesophageal cancer and situation after endoscopic tube placement: Note improved visualisation of mediastinal anatomy in the storage phosphor radiograph due to image processing

Micronodular opacities, as seen in reticulonodular interstitial disease, miliary disease or subtle atypical pneumonia represent another critical type of structure. These lesions depend on both contrast and spatial resolution. Visualisation of micronodular opacities is directly related to image noise and thus highly influenced by the quantum efficiency of the detector. While several clinical studies [8, 15] suggested an equivalent performance, e. g. for detection of interstitial disease, there are controversial results in phantom studies [1, 12, 13, 25, 26]. Lesion detection depends mainly on image processing. In systems that allow only for unsharp mask filtering with a single kernel size, such as older Fuji systems, a medium size kernel (RN = 4–5) should be avoided, although it is still implemented as the standard algorithm in these systems. Filtering with medium size kernels (5 mm) non-selectively enhances small pulmonary vessels as well as pathologic lesions. The result is a very irritating image that obscures pathology [15, 25]. Radiologists tend to overcall interstitial lung disease but may miss low contrast patchy opacifications. A large kernel of 25 to 30 mm (RN = 0) was found to be advantageous for various types of lung lesions and should be used as the standard on this equipment (Fig. 5.). MUSICA processing (Agfa) does not suffer from this problem as it takes several frequency bands into consideration [21].

Skeletal radiology

Skeletal radiology takes advantage of the wide detector latitude. Soft tissues and bones can be adequately visualised after adequate image processing (dynamic range reduction) [28] (Fig. 6.).

Although a wide signal range such as in lateral views of the thoraco-lumbar spine or in ap views of the shoulder can be displayed, image processing of these areas is more problematic (Fig. 7.). Best results are obtained with advanced processing algorithms [27, 29]. Unsharp masking tends to produce images of lower contrast that are still superior to conventional radiographs of these regions, though.

Consistent image quality independent of exposure is an important advantage since many radiographs are taken without phototiming. Automatic detection of the collimated areas is a more critical task in skeletal than in chest radiography. With older systems, we found up to 20 % of images (depending on the anatomic region)

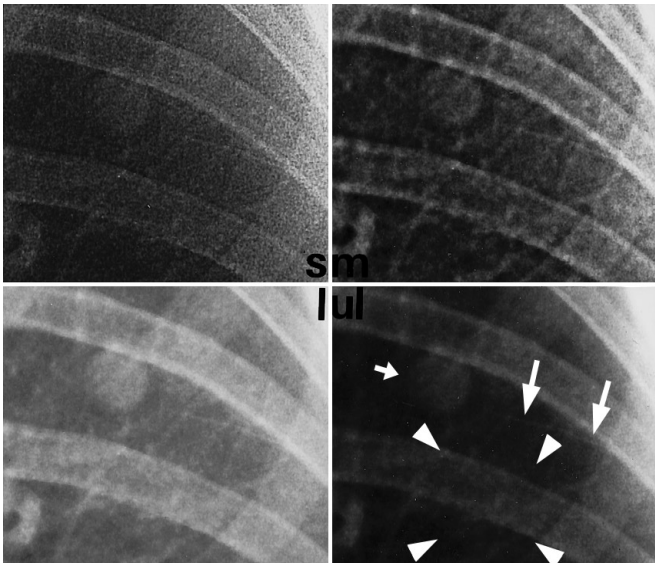


Fig. 5. Effect of unsharp mask filtering on visualisation of simulated pulmonary lesions (nodule = *short arrow*; line = *long arrows*; micronodular opacities = *between arrowheads*): *s* small kernel size (1.4 mm), *m* medium sized kernel (5 mm), *l* large kernel (2.5 cm) and *ul* ultralarge kernel (7 cm). Note the increased image noise in s) with decreased detectability of the low contrast structures, *m* is a rather irritating image due to non-selective enhancement of vascular structures, *l* shows the effects of dynamic range compression and local contrast enhancement at best advantage, while *ul* suffers from unsharpness and too low effects of MTF restoration

too light or too dark due to suboptimum detection of the object range. Newer systems are very reliable with respect to optimising image contrast and density even in situations in which two exposures are made on one cassette.

Anatomical areas with high X-ray attenuation such as lateral views of the spine or axial views of the hip are most vulnerable to image noise. Especially when image processing is performed to compensate for the wide object range, the image noise may exceed the levels that radiologists are used to in conventional radiography. Improved detector technology (higher DQE) and advanced processing (digital noise suppression) will improve image quality also in those areas.

Critical imaging tasks are subperiostal resorption in hyperparathyroidism (Fig. 8.), erosive lesions in rheumatic disease, non-displaced fine fracture lines and ill-defined low-contrast osteolytic or osteodense lesions. Detection of these structures requires optimised processing, especially MTF restoration with enhancement of high spatial frequencies and local contrast enhancement. Given proper processing, conventional and digital systems will yield equivalent performance [30, 31, 32].

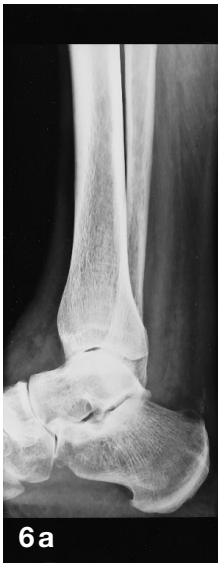
Bedside radiology

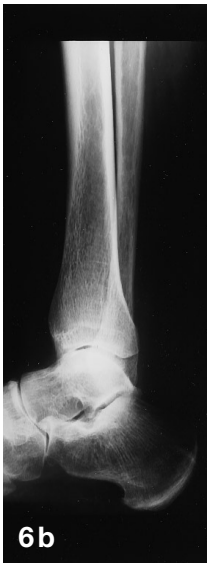
The wide dynamic range in combination with the signal normalisation process has lead to a wide acceptance of digital radiography in an environment where no phototiming is available and exposure conditions are rapidly changing as seen in intensive care or emergency units (Fig. 1.). Repeat exposures due to exposure errors could almost be eliminated. A more constant and optimised image density is highly appreciated by the clinicians, monitor devices can be seen more easily with adequate processing [33, 34]. Dose requirements for digital systems are in the 250 to 400 speed range dependent on the type of detector used. Low exposure doses tend to obscure the detection of fine intravenous catheters [35].

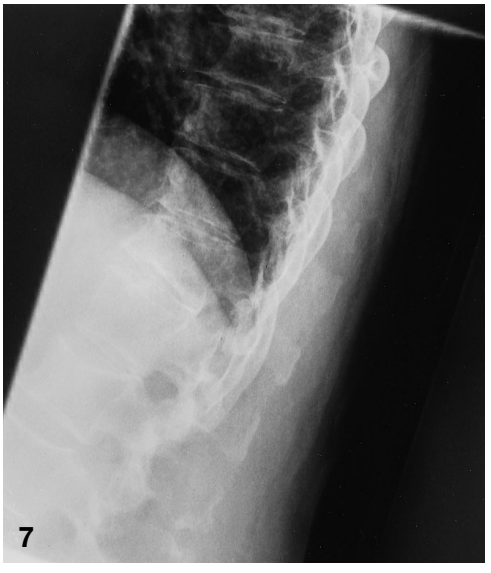
Fig. 6. Digital **(a)** and conventional **(b)** lateral view of the ankle illustrating the superiority of digital versus conventional technique to simultaneously show soft tissues and bone structures with optimised density and detail detectability

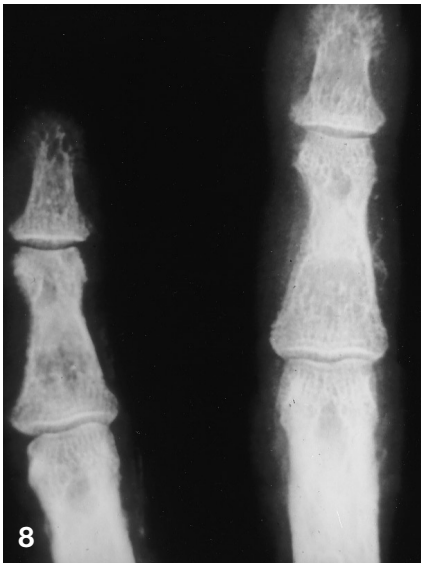
Fig. 7. Lateral digital view of the thoracolumbar spine demonstrating the advantage of the wide dynamic range in areas of wide attenuation differences

Fig. 8. Zoomed digital view of fingers with subperiostal resorptions typically seen in patients with hyperparathyroidism. Given appropriate processing, digital images adequately demonstrate these types of subtle pathology









Summary

At present, digital radiography using storage phosphor plates as detector will yield at least an equivalent diagnostic performance as conventional radiography. Diagnostic advantages are present for the mediastinum, for bedside imaging and generally in all situations in which no phototiming is available. The number of retakes due to incorrect exposures can be reduced. Since all of these are of moderate importance in daily practice, organisational aspects, such as digital storage and communication and cost reduction by means of film savings, will become the driving force for the installation of digital radiographic systems.

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