Journal of Thoracic Imaging
18:148–164 © 2003 Lippincott Williams & Wilkins, Inc., Philadelphia

Principles of Image Processing in Digital Chest Radiography

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Summary: Image processing has a major impact on image quality and diagnostic performance of digital chest radiographs. Goals of processing are to reduce the dynamic range of the image data to capture the full range of attenuation differences between lungs and mediastinum, to improve the modulation transfer function to optimize spatial resolution, to enhance structural contrast, and to suppress image noise. Image processing comprises look-up table operations and spatial filtering. Look-up table operations allow for automated signal normalization and arbitrary choice of image gradation. The most simple and still widely applied spatial filtering algorithms are based on unsharp masking. Various modifications were introduced for dynamic range reduction and MTF restoration. More elaborate and more effective are multiscale frequency processing algorithms. They are based on the subdivision of an image in multiple frequency bands according to its structural composition. This allows for a wide range of image manipulations including a size-independent enhancement of low-contrast structures. Principles of the various algorithms will be explained and their impact on image appearance will be illustrated by clinical examples. Optimum and sub-optimum parameter settings are discussed and pitfalls will be explained.

INTRODUCTION

Digital chest radiography has now come of age. Almost 2 decades have passed since the introduction of the first digital storage phosphor systems (Computed Radiography, CR) in the early 1980s. Digital image processing has always been an integral part of digital radiography,[1] but most users are hardly aware of the processing techniques integrated in their systems. The term "postprocessing" is commonly associated with the processing option available for the user and is often distinguished from the default processing that all digital radiographs are subjected to. In reality, this distinction is arbitrary since available processing options are generally identical. In an ideal environment, the default processing should be chosen so that no additional "postprocessing" is necessary.

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This article will explain principles of the various image processing algorithms and will illustrate their impact on clinical chest radiographs. Adequate parameter settings are suggested, sub-optimum parameter settings are discussed and pitfalls will be explained.

WHY IMAGE PROCESSING?

Digital radiography systems are characterized by a very wide dynamic range and linear response to the incident radiation. They can therefore capture the wide attenuation differences between lungs and mediastinum and are much less vulnerable to changes in exposure dose than conventional screen-film radiographs.[2] If no further processing were employed and the images captured by the detector systems were directly transformed into gray levels on a viewing monitor or laser film, the resulting image would be characterized by a good transparency of the mediastinum but otherwise would appear extremely "gray" because of a lack of contrast. At the same time, image sharpness may not be as good as in screen-film systems because of limitations due to pixel size and less favorable detector characteristics.

For this reason, all digital radiographs are subjected to image processing. Digital image processing is not only used to take full advantage of the positive characteristics of digital radiography systems and to overcome the limitations of conventional chest radiography but also to ameliorate the limitations of digital detectors. Image processing has a major impact on perceived image quality and may even influence the diagnostic performance of digital chest radiographs. General goals of processing are:

- to display the full range of attenuation differences in the chest,
- to optimize spatial resolution of digital chest radiographs,
- to enhance structural contrast in the lungs and mediastinum, and
- to suppress image noise.

Overcoming Limitations of Digital Detectors

Conventional film-screen systems (100 speed) have a high spatial resolution of well above 5 cycles/mm that is markedly superior to digital systems (Fig. 1). Even 400 speed systems still have a spatial resolution that is in the range of 4–5 cycles/mm. Resolution of digital systems is mainly limited by the pixel size. In the chest only 4K systems with a pixel size of 0.1mm have a maximum spatial resolution that is 5 cycles/mm and thus equivalent to conventional systems.

In clinical practice, however, the visual impression of "sharpness" of contour definition is less determined by the maximum spatial resolution than by the behavior of the MTF (see below) at lower spatial frequencies. This behavior can be substantially improved by appropriate digital processing.

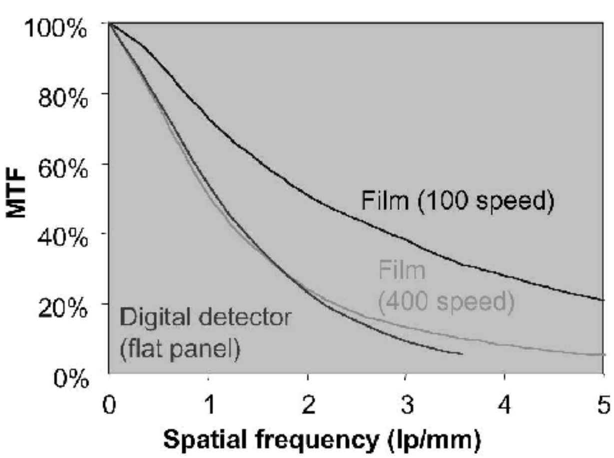


FIGURE 1. The MTF of an unprocessed digital image (flat panel detector) is similar to a 400 speed conventional radiograph but substantially worse than a 100 speed radiograph.

Overcoming Limitations of Conventional Radiography

Conventional radiography is susceptible to exposure errors and suffers from the inverse relation of contrast and latitude (Fig. 2). Digital detectors have a very wide dynamic range that makes them less vulnerable to exposure errors and allows for including the full range of object attenuation in the digital image. Image processing can then be used to take advantage of these detector characteristics.

Exposure errors are particularly common for bedside chest radiographs, which explains why digital radiography systems are very popular for this application.[2] Digital systems automatically correct for differences in exposure and virtually guarantee constant contrast and density of the digital images.

Screen-film radiography can only vary image contrast and latitude by changing the system's gradation curve via an appropriate screen-film combination. A high-contrast film (eg, a G-film) will suffer from a reduced latitude and therefore a "white" mediastinum. A wide latitude film (L-film or C-film) will be able to display the mediastinum in a transparent fashion but have a lower image contrast. Even asymmetric screen-film systems cannot overcome the inverse relation between latitude and contrast (Fig. 2). By combining 2 screens of different sensitivity and gradation characteristic, the resulting gradation curve of the whole system will have more contrast in the high absorption areas of the mediastinum at the expense of a reduced detail contrast in the lung.[3] Thus, asymmetric screen-film systems allow a redistribution of the detail contrast but no general contrast enhancement in all chest areas. Proper digital image processing can decouple the effects of gradation (overall gross density variations) from the effect of detail contrast (local density variations) and therefore can be used to optimize both.

THE BASIC DESCRIPTORS: GRADATION AND MTF

In screen-film radiography, the *gradation curve* describes how the incident radiation is transformed into gray levels (optical densities, OD) on the x-ray film (Fig. 2c). In digital processing, a gradation curve describes how digital input values are transformed into digital output values (gradational adjustment). In digital radiography systems, an additional gradation curve is needed to describe how the digital output values are transformed into gray levels on a viewing monitor or laser film. The effect of a gradation curve on a structure of interest only

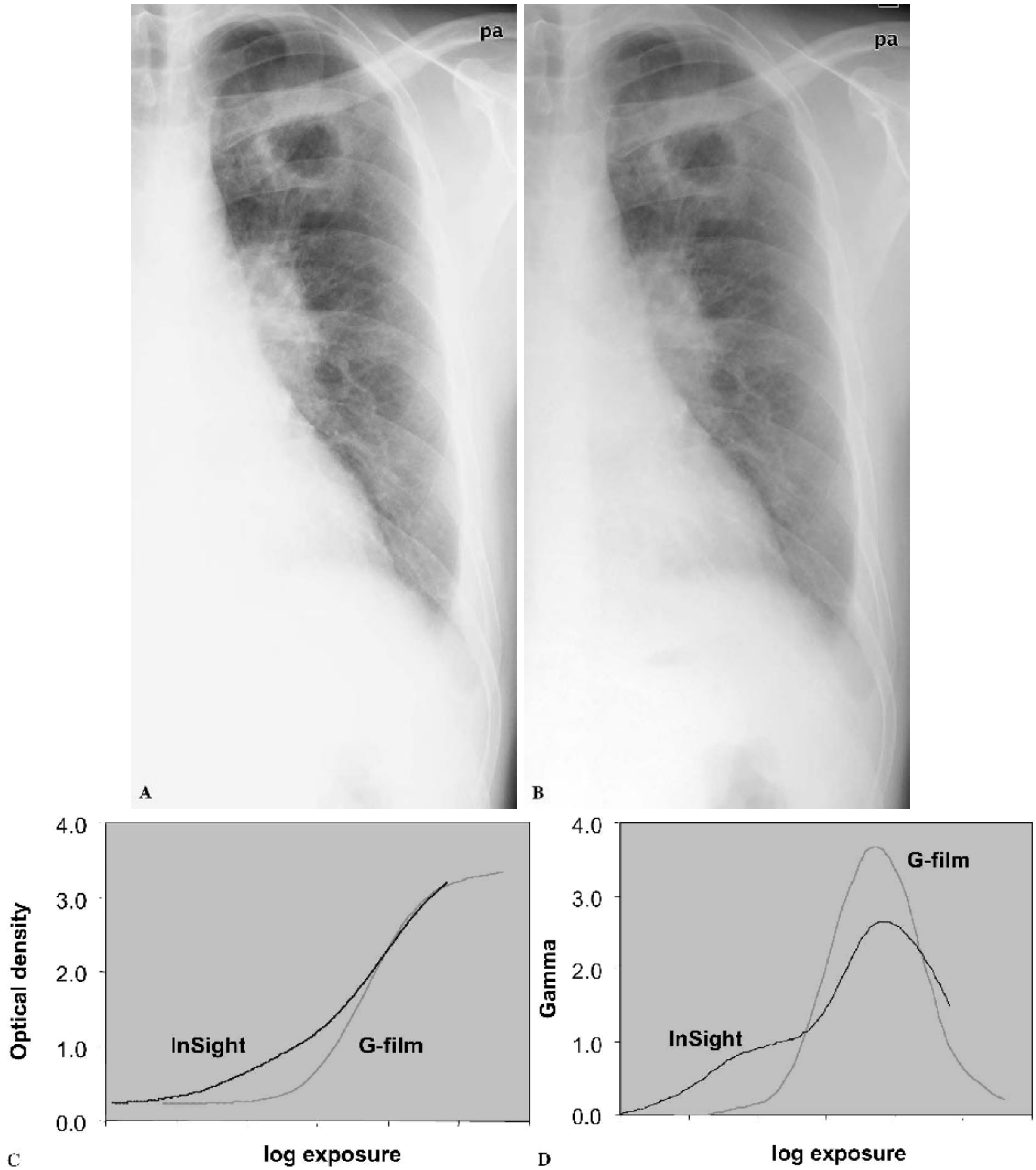


FIGURE 2. A G-type film has a high contrast in the lungs but displays the mediastinum suboptimally (**A**). An asymmetric film (InSight) yields a better contrast and higher density in the mediastinum but at the expense of less contrast in the lungs (**B**). The relation of optical density of the film to the detector dose is described by the film gradation (**C**). Its first derivative (slope) determines the local contrast or gamma (**D**).

depends on its pixel values but not on the size, contrast, or spatial frequency characteristics of the structure.

The *modulation transfer function* (MTF), on the other hand, describes how an imaging system transmits the various spatial frequencies of an image (see Figure 1).[4] High spatial frequencies correspond to small detail sizes and sharp contour definition while low spatial frequencies correspond to large detail sizes (such as the variations between lungs and mediastinum). The MTF is an essential property of any imaging system. In fact, every step during image capture and processing can be characterized by its own MTF. In general, every analog imaging step has a MTF that eventually decreases down to 0 at high spatial frequencies. The longer the MTF curve stays close to 1 (100% of the original value) and the higher the spatial frequency is at which it becomes 0, the sharper will the resulting image look. In analog (screen-film) systems, the maximum resolution is commonly described as the spatial frequency at which the MTF decreases to 4%. The maximum resolution depends on the thickness of the intensifying screen and, thus, the system speed.

For digital detector systems, the intrinsic analog MTF of the detector may vary substantially depending on detector material and technology. Digitization with pixels of finite size cuts off this MTF at the Nyquist frequency ν_N, which can be calculated from the pixel size p as follows: $\nu_N = 1/(2 \times p)$. Aliasing occurs when the MTF before digitization (presampling MTF) is non-vanishing beyond the Nyquist limit. The MTF of the total imaging system is obtained by multiplying the MTFs of each step involved. Thus, the MTF of the digitization process ultimately limits the final output.

Image processing can arbitrarily alter the input MTF. The spatial frequency composition of an image remains constant if the MTF is a horizontal line at 1 (100%), which means that all spatial frequencies of the input are kept identical on the output image. Gradational adjustment is one of the processing techniques that do just that. Most advanced processing techniques decompose the input image into 2 or more frequency ranges (frequency bands) and subject these frequency bands to various measures described below. These processing techniques change the spatial frequency content of the image and have therefore an MTF that deviates from 1.

GRADATIONAL CURVES

Since the dynamic range and therefore the potential range of signal values of digital detector systems is very large, only a portion of this range will actually contain diagnostically relevant information. The relevant range is detected by automated signal normalization, which is nothing else than a suitable linear gradation curve. Various more complex algorithms may then be used for further processing (see below). The resulting image is then usually subjected to a final gradational adjustment to obtain a more "conventional look."

Automated Signal Normalization

Automated signal normalization detects the signal range with diagnostically relevant information excluding the regions outside the collimated area and direct, unattenuated radiation. Automated signal normalization is done in 2 steps. First, the collimated region is detected to only process relevant information. Secondly, the pixel values found within this area are subjected to a histogram analysis. A linear gradation curve (linear look-up table, LUT) is then used to exclude input values where the histogram is (close to) zero. This technique automatically finds the correct signal range and can automatically compensate for exposure errors.[1] It will yield constant density and contrast independent of the exposure dose (Fig. 3). However, underexposure will lead to an increased quantum mottle on the images.

The procedure has a welcome side effect, which is *latitude optimization*. There can be a substantial variation of absorption differences in slim and obese individuals. In obese individuals this often causes relatively underexposed (low density) areas on conventional radiographs. Signal normalization takes these differences into account and ensures a more constant image quality independent of patient size (Fig. 4).

Gradational Adjustment

Gradational adjustment is the final step in the chain of image processing and comes after all other processing tasks are finished. Gradational adjustment is usually performed by a look-up table (LUT) stored in the system. This LUT stores the output pixel values for all input values (eg, from 0 to 4095 for a 12 bit system). The linear characteristics of a digital system are transformed back by gradational adjustment into a more convenient shape of the gradation curve (compare Fig. 2). Each system has a prestored set of look-up tables that can be (more or less easily) varied by the user. A gradation curve that compensates the density-dependent sensitivity of the human

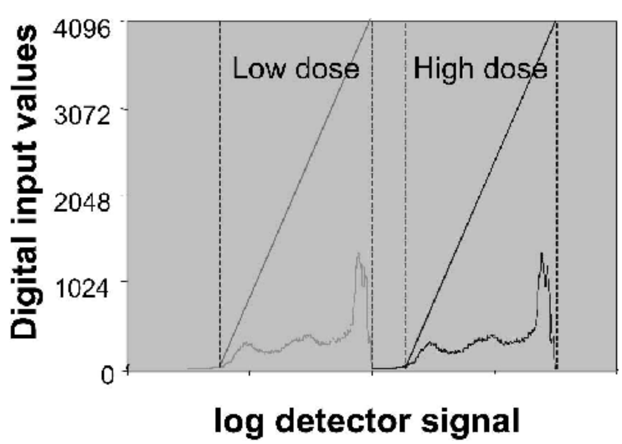
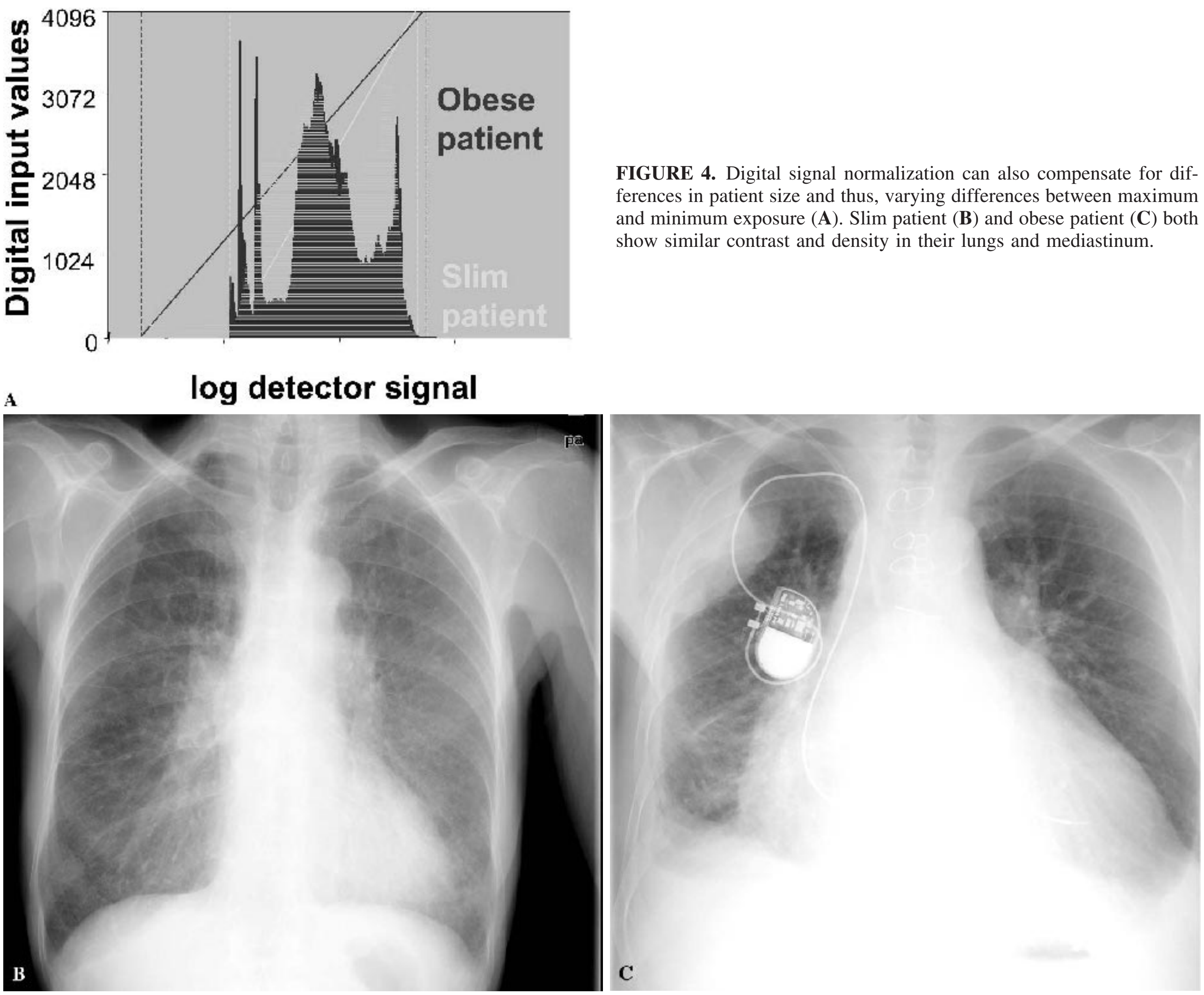


FIGURE 3. Digital detectors can transform a very wide range of input signals faithfully into digital data. Automated signal normalization detects the diagnostically relevant signal and assigns it to the digital data range (0 to max. bit), independent of the exposure *dose*.



A

B

C

FIGURE 4. Digital signal normalization can also compensate for differences in patient size and thus, varying differences between maximum and minimum exposure (**A**). Slim patient (**B**) and obese patient (**C**) both show similar contrast and density in their lungs and mediastinum.

eye has been suggested as a theoretical optimum (*Kanamori curve*).[5] However, most radiologists prefer a more sigmoid shape to match the appearance of a conventional chest radiograph. No final consensus is available yet.

SPATIAL FREQUENCY FILTERING

Spatial frequency filtering includes a wide range of image processing techniques that all change the spatial frequency composition of an image.[1,6–11] The 2 most common techniques in digital radiography are unsharp masking and multiscale filtering.

Unsharp Mask Filtering

Unsharp mask filtering or shorter, unsharp masking, is the generic name for a simple technique that was developed for edge enhancement and sharpening of images. It is implemented on the majority of digital computed radiography and direct radiography systems.[1]

In a first step, the image is blurred, using *low-pass filtering* (Fig. 5). This is usually done by locally averaging the pixel values. The wider the region (*kernel size*) used for this averaging process, the more blurred the low-pass image and the lower the spatial frequencies that remain in that image. Structures that are smaller than the kernel size are (almost) completely suppressed and no longer visible on this image.[12]

Subtracting the low-pass image from the original yields the *high-pass information* and contains those details that are suppressed on the low-pass image. Since sharp contours on the original image always contain high spatial frequencies, these contours are retained in the high-pass image. Otherwise, the size of a structure de-

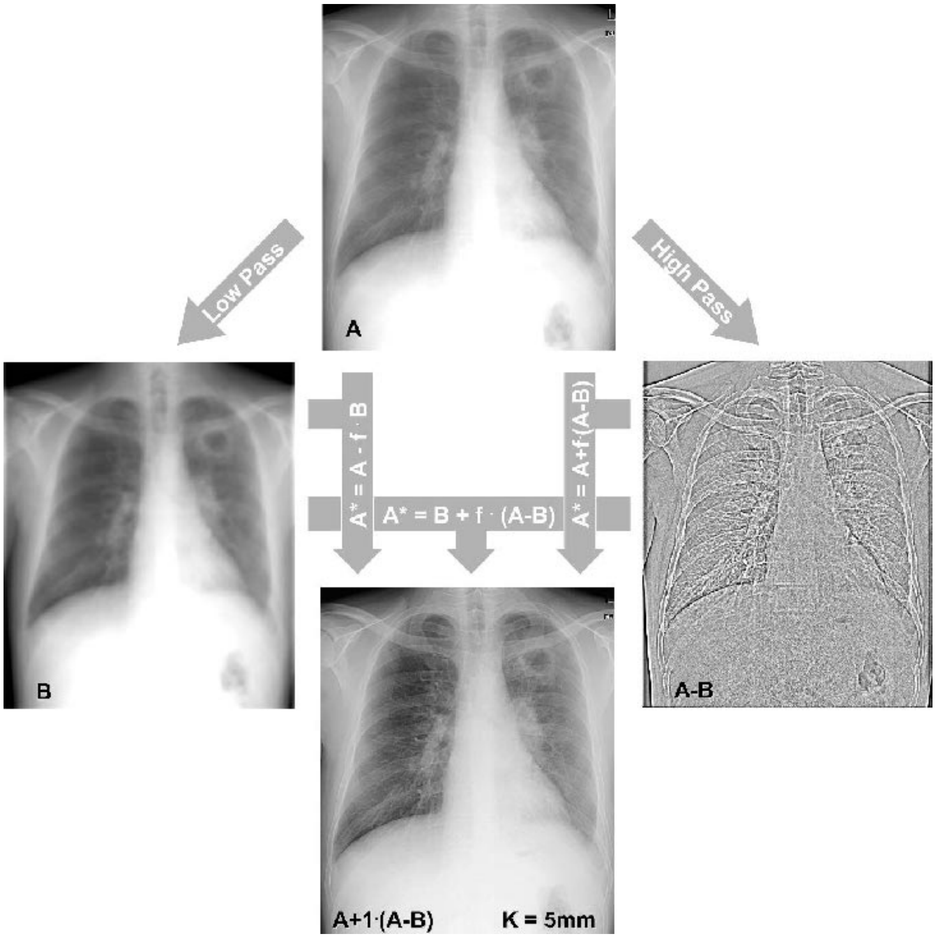


FIGURE 5. Unsharp masking is based on a low-pass version **B** that is created from the original image **A**. Subtraction yields a high-pass image (**A–B**). Three versions of weighted addition and subtraction are available that can produce very similar filtered images (**A***).

termines if it is visible in the high- or low-pass image. The transition between high- and low-pass images depends on the kernel size.

For the *final filtered image*, various combinations of the original, low-pass or high-pass images can be created in a weighted fashion. The classic technique used in most computed radiography systems is to add the (weighted) high-pass image to the original. This leads to an accentuation of the detail information contained in the high-pass image. The weighting factor f is then called "*enhancement factor*." In Fuji-based systems, which are most widespread in clinical use, it is abbreviated as "RE." The weighting factor can assume arbitrary values >0. A factor f of 0.5 corresponds to slight enhancement that is commonly used for chest radiography. Factors larger than 2 should in general be avoided in order not to create "over-enhancement" (Fig. 6).

Alternatively, the (weighted) low-pass can be subtracted from the original. Weighting factors f* are then between 0 and 1. The 2 factors are related as follows: f* = f/(1+f). As a final alternative, the high-pass image can be added to the low-pass image with a factor that is equal to 1+f. All these techniques are mathematically equivalent (given proper normalization) if f remains constant independent of local pixel values. Special effects (eg, dynamic range reduction, see below) can be achieved if images are subjected to a look-up table operation (gradational adjustment) before weighted addition.

The effects of unsharp masking are strongly dependent on the size of the filter kernel and the resulting "transition frequency" between high- and low-pass filtered images (Fig. 7). In general, *all* spatial frequencies that are higher than the transition frequency are enhanced while all frequencies lower than the transition frequency are relatively suppressed. It has been shown that kernel sizes of 20–30 mm give the best overall results for general chest imaging[13]; smaller kernels may mask diagnostically important information (see pitfalls below). Modifications are possible by using 2 different filter kernels to more selectively enhance or suppress specific details.

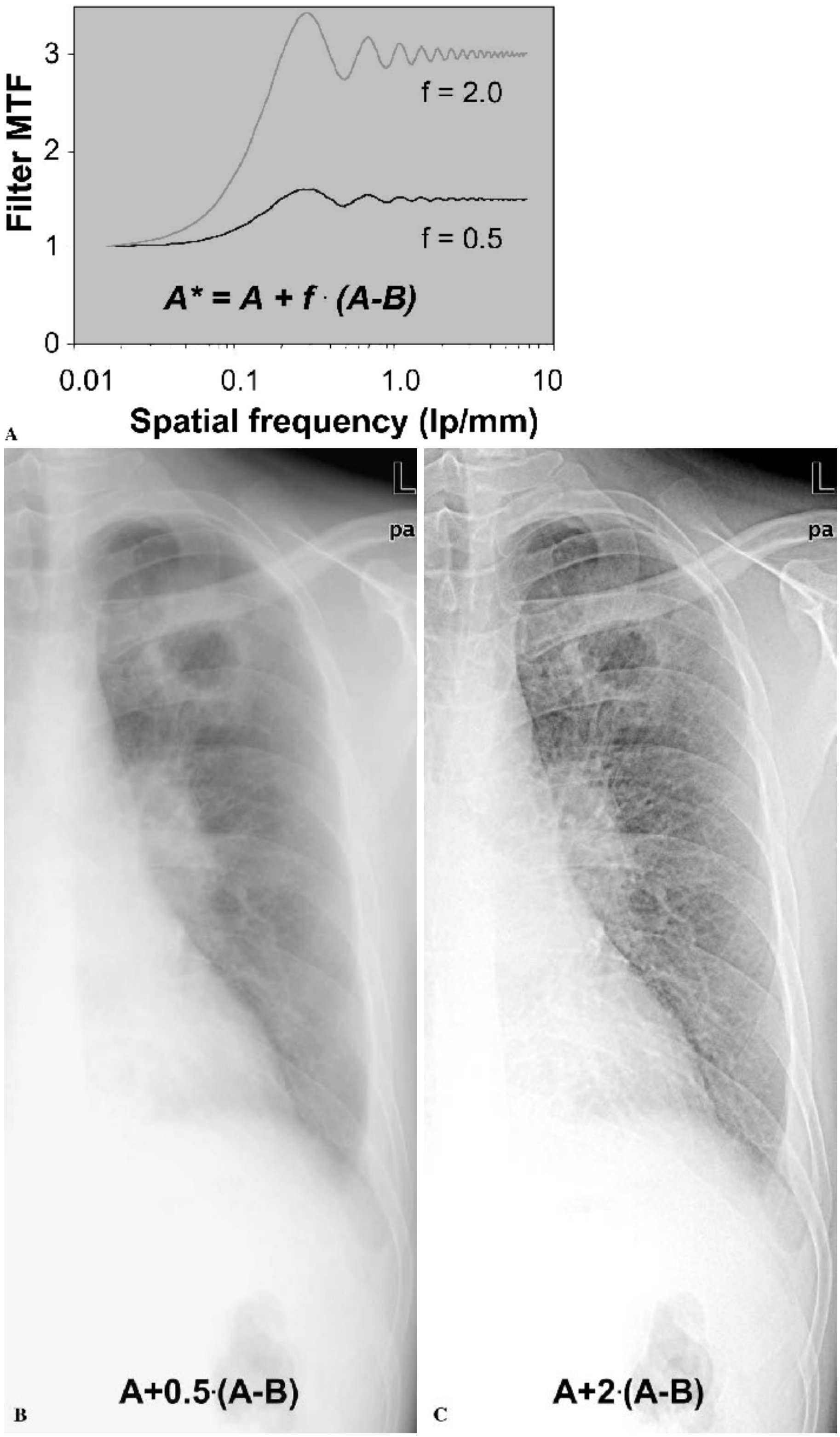


FIGURE 6. The filter MTF demonstrates how spatial frequencies are affected **A.** The enhancement factor f determines the strength of the filtering process. Compare a conventional looking image at f = 0.5 (**B**) to a strongly filtered image at f = 2 (**C**).

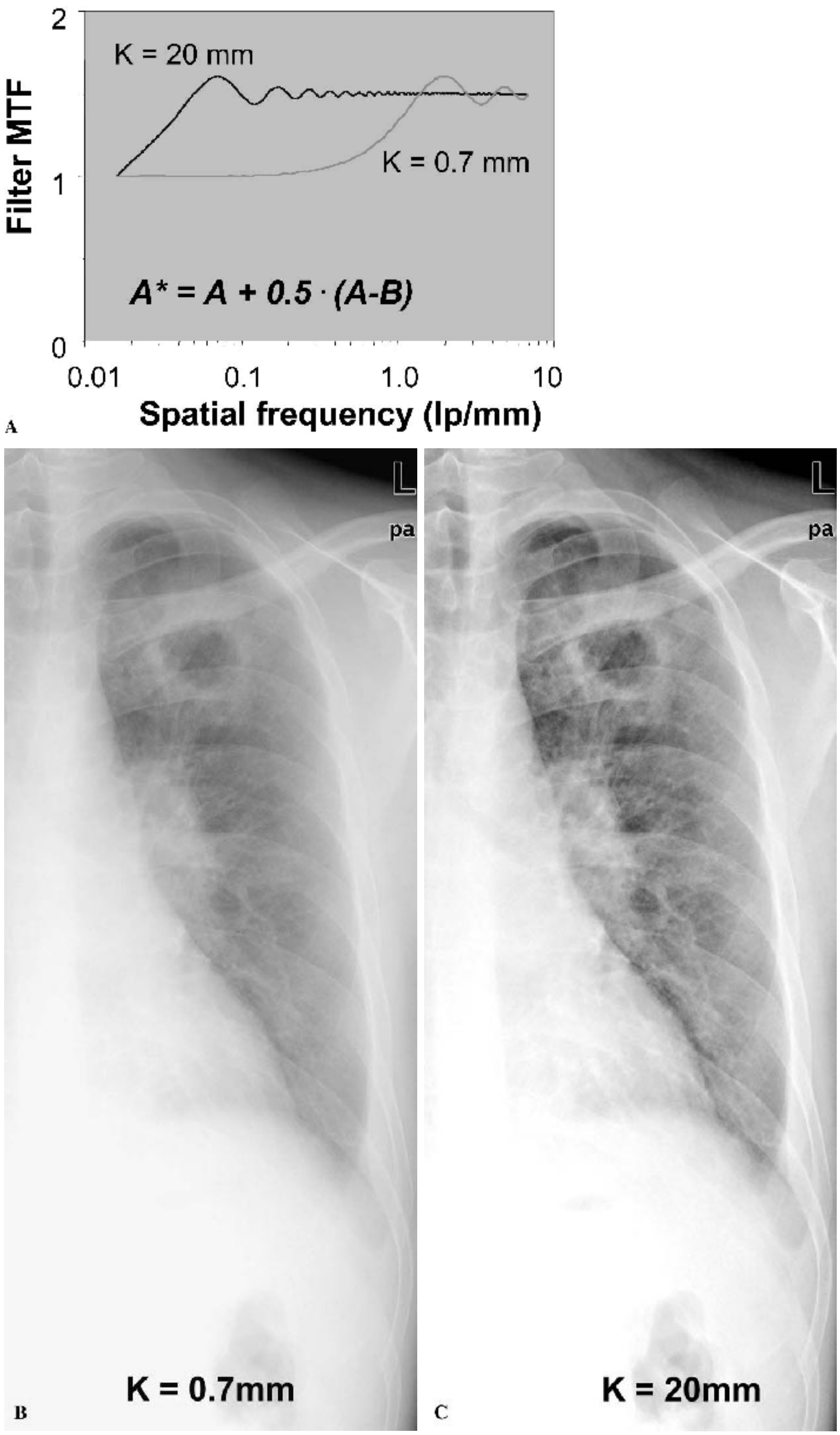


FIGURE 7. The kernel size K determines the cutoff between structures that are enhanced and suppressed (**A**). A small kernel (K = 0.7mm) only leads to enhancement of fine structures (hardly visible on the printed reproduction) and thus improves spatial resolution (**B**). A large kernel (K = 2 cm) enhances all structures that are smaller than its size (**C**) and thus leads to a general detail enhancement.

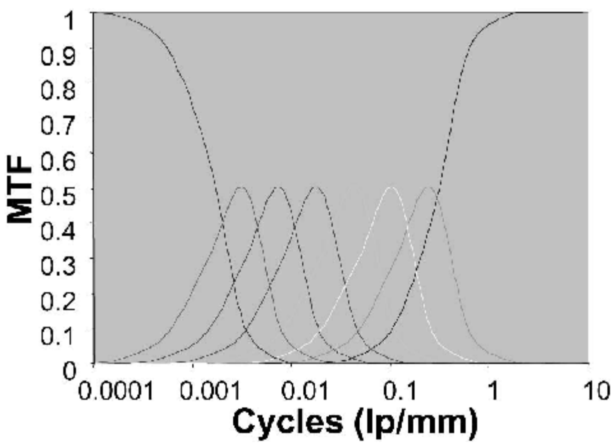


FIGURE 8. Multiscale processing is based on decomposition of the original image **A** into bandpass images B_n.

Multiscale Processing

Multiscale processing decomposes the original image into multiple frequency bands that contain information only from a particular structural size. For filtering, each of these sub-bands can be treated separately, which allows for a wide variety of processing options. Multiscale processing is available under various trade names, for example, MUSICA (Agfa),[10] MFP (Fuji),[7] and UNIQUE (Philips).[8]

The decomposition step can be implemented by repeatedly splitting the image into a high-pass component and a low-pass component.[14] This process starts with the highest frequency band. The resulting low-pass component is then taken as input to the next stage until 8 or more separate frequency bands have been created. All but the lowest of these frequency band images contain positive as well as negative pixel values (due to the subtraction process) that indicate the contrast of a structure within this specific frequency band. The lowest band contains only the large-area density differences in the image, for example, between lungs and chest wall/mediastinum (Figs. 8 and 9).

Multiscale processing is more flexible than simple unsharp masking. By enhancing individual frequency bands or groups of frequency bands, size-specific processing is possible. By applying the enhancement to a whole group of frequency bands (including the highest one), the same effects as with unsharp masking can be reached.

Most interesting are nonlinear techniques in which the images that correspond to the various frequency bands are subjected to look-up table operations (gradational adjustment) that can be identical for all (Fig. 10) or vary with spatial frequency (Fig. 11). Contrast-dependent enhancement modifies the contrast in each frequency band using appropriate LUTs, and thus allows for enhancement of low-contrast structures and suppression of high-contrast structures. Size-dependent enhancement is possible by enhancing the various sub-bands differently.

Multiscale processing yields the following advantages:

- hard-to-detect (low-contrast) structures are enhanced,
- conventionally over- or underexposed regions are better displayed,
- edge artifacts are eliminated, and
- the natural image impression is retained.

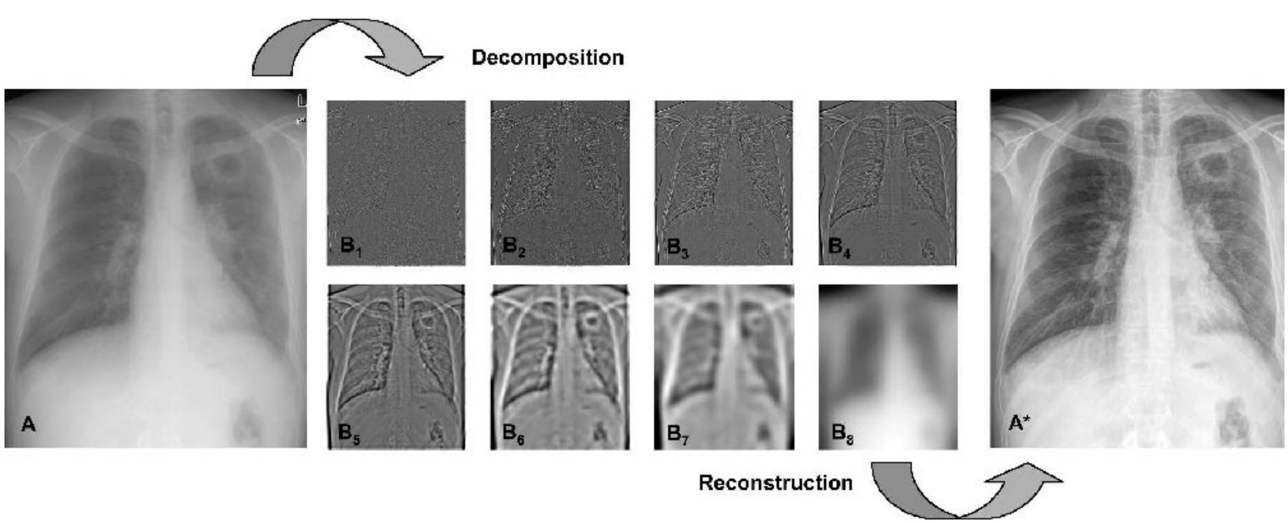


FIGURE 9. Multiscale processing: After decomposition of the original image **A** into bandpass images B_n, the filtered image is created by subjecting each frequency band to LUT operations that (for example) enhance low-contrast structures. The enhancement may be identical for each frequency band or may vary depending on the size of the structures to be enhanced or suppressed.

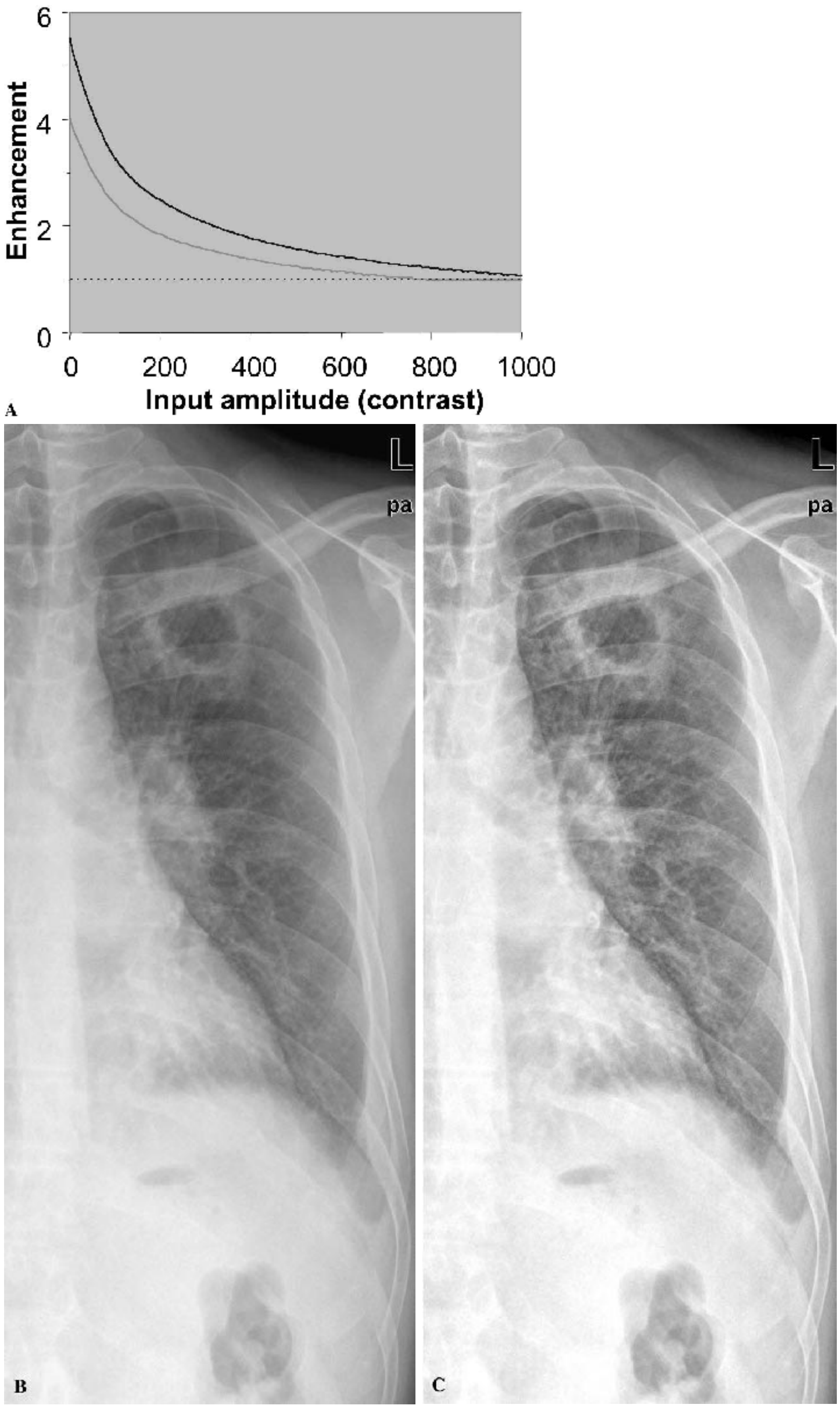


FIGURE 10. The enhancement curve determines how enhancement depends on local contrast. Enhancement is high for small contrast differences (input amplitude) and decreases for large contrast (**A**). Compare moderate (**B**) to strong enhancement (**C**).

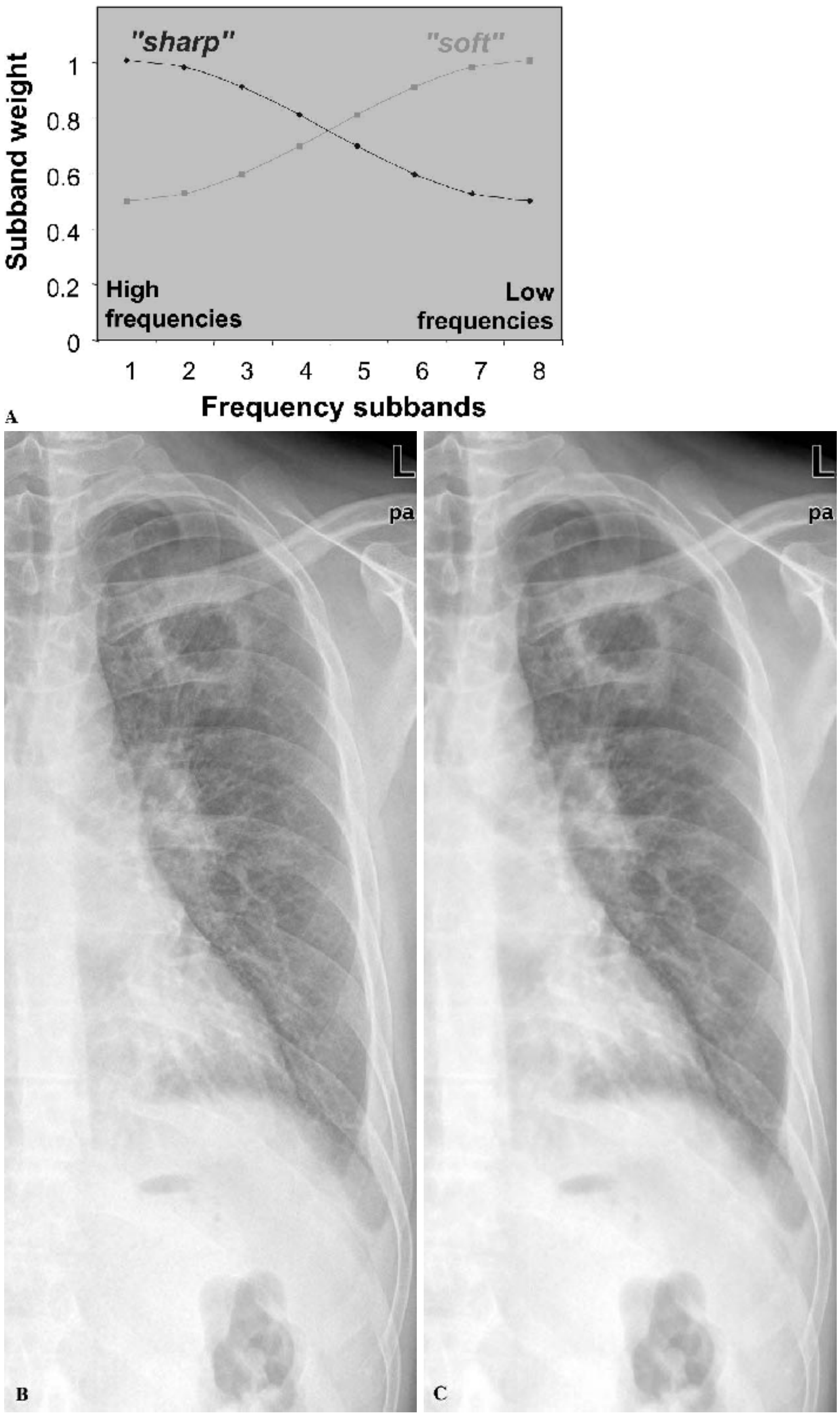


FIGURE 11. The sub-band weight can be used to relatively overenhance or suppress structures depending on their size (as defined by the sub-band that they are displayed in) (**A**). Compare the image character for relative stronger enhancement of small structures ("sharp," **B**) or larger structures ("smooth," **C**).

APPLICATIONS

Ideally, image processing provides dynamic range reduction to better display lungs and mediastinum, detail enhancement for improved visualization of subtle abnormalities, optimization of spatial resolution to compensate for the lower resolution of digital detectors, and noise reduction to compensate for the unfavorable quantum statistics in the mediastinum. Multiscale processing can provide all of the above. With unsharp masking, the radiologist has to select the most important tasks unless more than 1 kernel size can be used for processing.

Dynamic Range Reduction

Chest radiographs have to cope with a large dynamic range caused by large variations of absorption between the lungs and the mediastinum. Dynamic range reduction (DRR), also called dynamic range compression (DRC), is a processing technique that reduces the latitude of the image while keeping the detail contrast high. DRR is able to "harmonize" the image and to improve the visualization of structures in high and low absorption areas (Fig. 12).[11,12,15]

With computed radiography systems, processing is based on variants of unsharp mask filtering. DRR is then accomplished by nonlinear subtraction (using LUTs) of a low-pass image from the original. This low-pass image uses a large kernel (2–4cm) and only contains the gross density variations between lungs and surrounding soft tissues. By subtracting this information in a weighted fashion, the optical density within the mediastinum can be increased without changing regions with optimum density like the lungs. To achieve this, the weighting factors are varied depending on the pixel values in the original or low-pass image (depending on implementation). No subtraction is performed for the lungs while weighting factors of up to 0.5 (best around 0.2) are used for the mediastinum.

Alternatively, the high- and low-pass components can be subjected to different gradation curves before being added back together. By using a flat gradation curve that emulates a wide latitude film for the low-pass image and

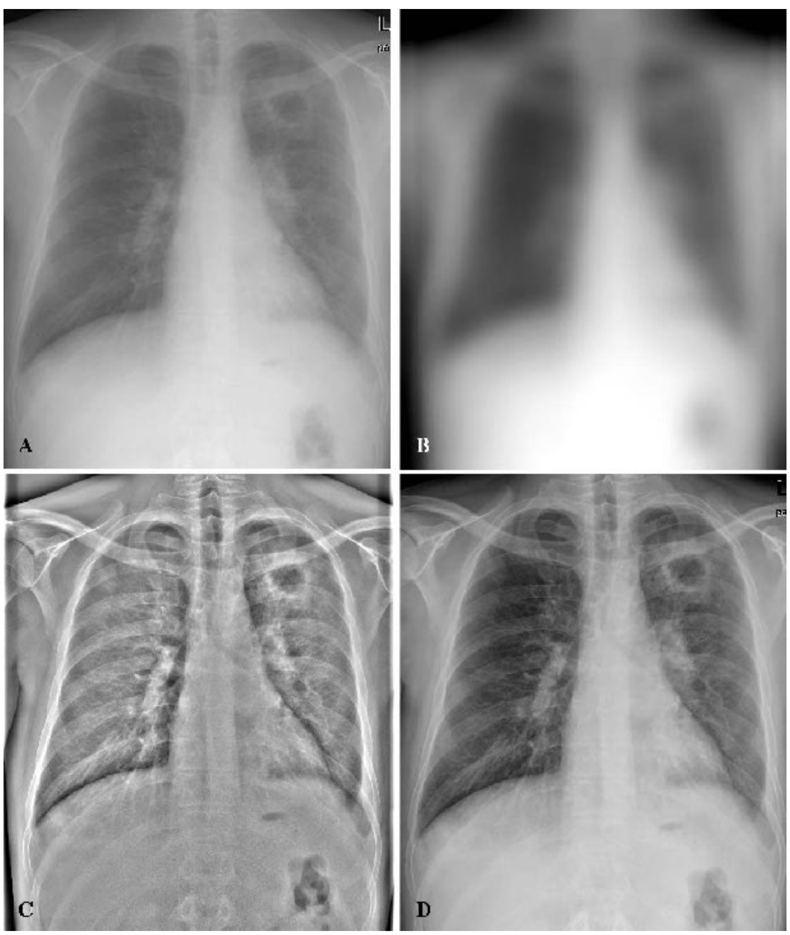


FIGURE 12. Dynamic range reduction uses a low-pass image (**B**) that only contains gross density differences to correct for very large differences between maximum and minimum attenuation (eg, lungs/mediastinum, soft tissue/bone, or thoracic/abdominal spine) Modifying it changes the latitude (compare Fig. 13). The high-pass image contains all detail information (**C**). Modifying it changes the *detail contrast* (also compare Fig. 13). The filtered image shows a harmonized image density and better contrast in the mediastinum (**F**).

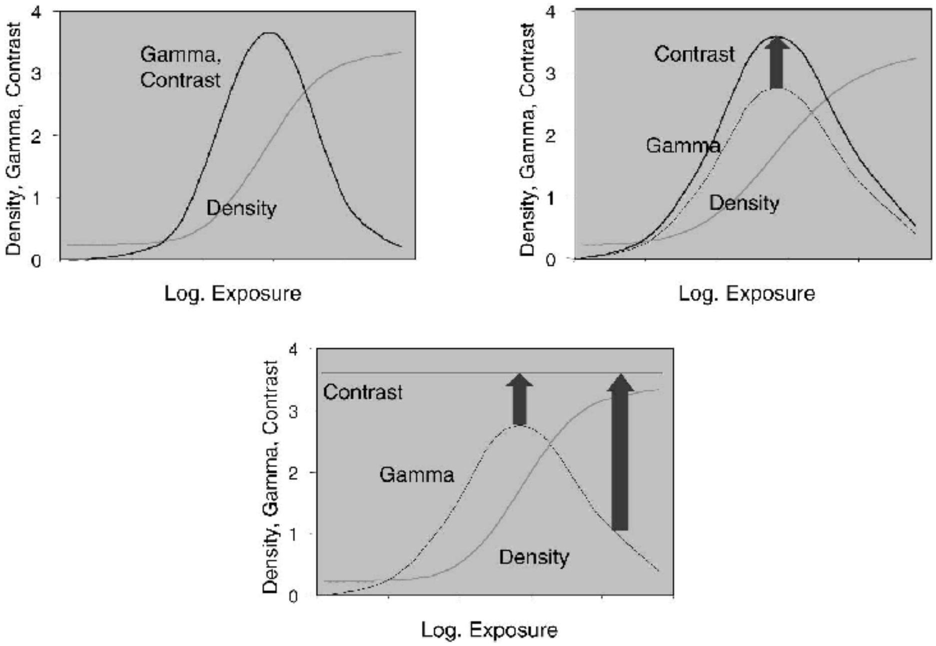


FIGURE 13. One of the main problems in conventional imaging is the coupling of the detail contrast to the density curve. For a film-like display the detail contrast is determined by the local gamma, that is, the derivative (slope) of the density curve (**A**). If a wider latitude is desired, the local gamma is lower. In conventional radiography this would lead to reduced detail contrast. With digital contrast enhancement (eg, by large-kernel unsharp masking) the detail contrast can be restored to the same level as with the higher gamma density curve (**B**). With standard enhancement algorithms the resulting detail contrast is still varying as a function of background density. Harmonized contrast designates a technique where the enhancement is adapted to the local density such that the same detail contrast is achieved in all image areas (**C**).

using a contrast curve (that provides the local contrast = first derivative of a gradation curve) of a G-film for the high-pass filtered image, the resulting image will combine the overall gradation characteristics of a wide latitude film (transparent mediastinum) with high local contrast (see Fig. 13b).[6,11]

With multiscale filtering, similar effects can be reached by subjecting the lowest frequency bands to gradation curves that decrease large contrasts and increase small ones (Fig. 14, see also Fig. 11).

Detail Enhancement

Structural enhancement is possible with unsharp mask filtering as well as with multiscale processing. Unsharp masking enhances all structures that are smaller than the kernel size independent of their intrinsic contrast, while

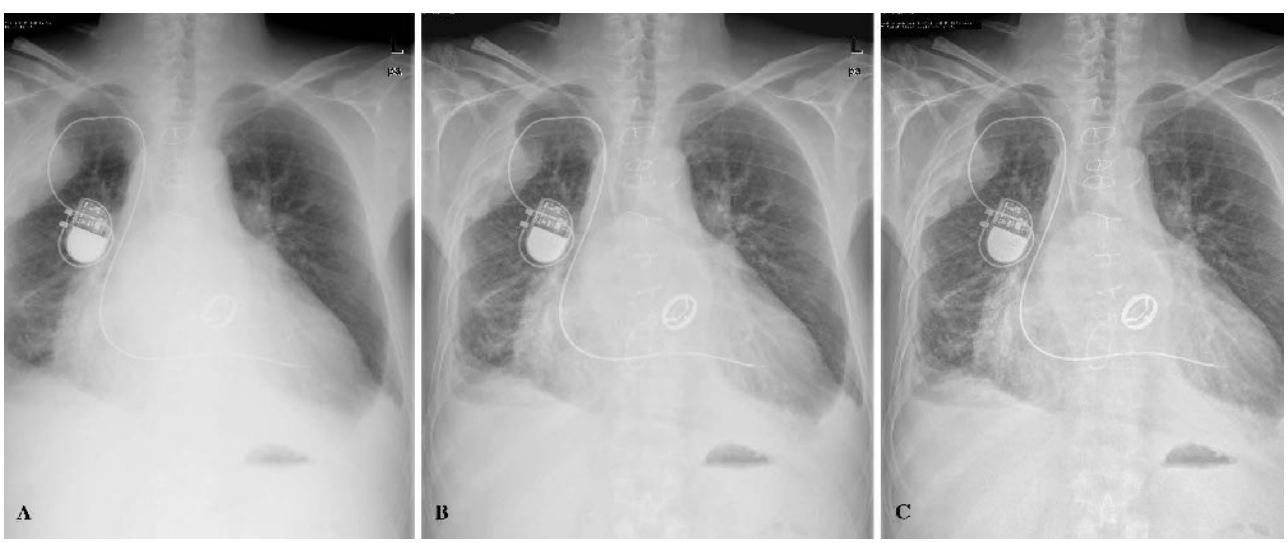


FIGURE 14. Compared with a (simulated) film image (**A**), unsharp mask filtering with a large kernel (**B**) and multiscale filtering with harmonized contrast (**C**) result in better detail visibility, especially in the mediastinum. Note the difference in conspicuity of the retrocardiac vessels, and the difference in sharpness of the cardiac pacemaker.

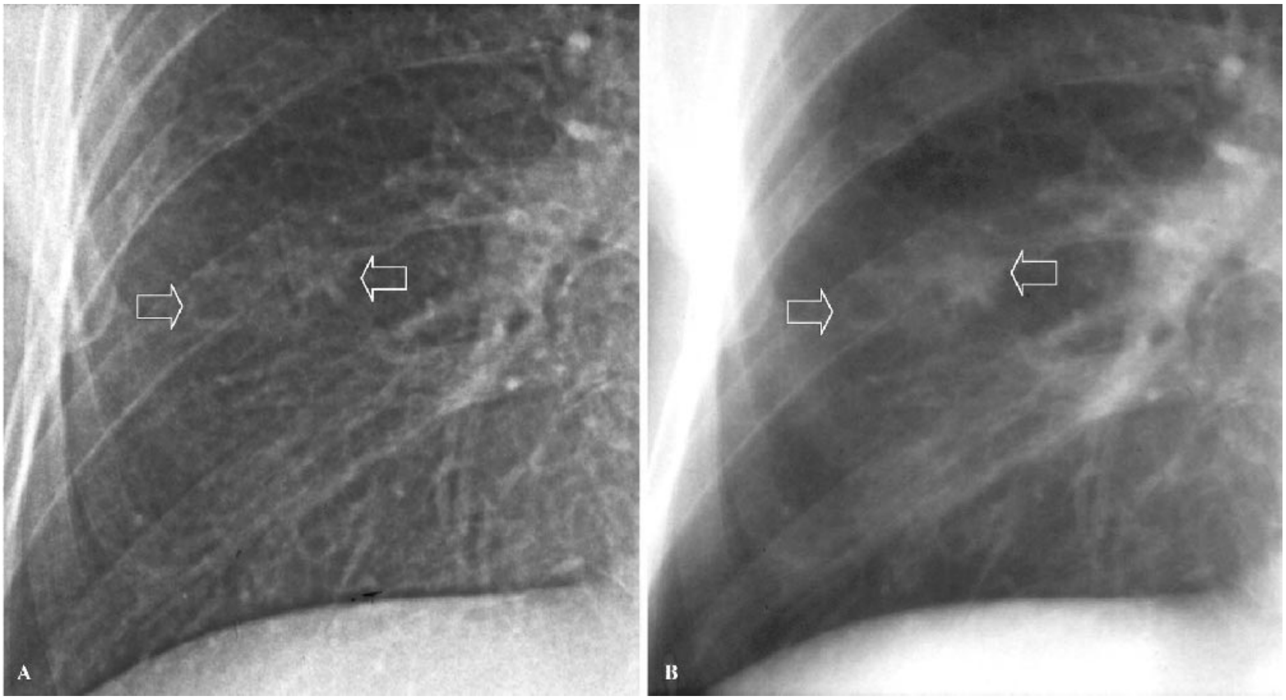


FIGURE 15. Unsharp masking with medium-sized kernel (5mm) for edge enhancement masks low-contrast opacities that lack well-defined edges, such as in this patient with septic emboli (arrows) (**A**). Filtering with large kernels (>1.5cm) enhances the whole septic focus (**B**).

multiscale filtering allows for contrast-dependent enhancement (Figs. 13 and 14).[11,13]

Detail enhancement can be used to improve the evaluation of intestinal markings or vessels in the chest and to enhance the conspicuity of lung nodules.

Multiscale processing is superior to unsharp masking and its nonlinear variants because local structural contrast can be enhanced without causing serious artifact that may suppress diagnostically important structures (see Fig. 15).[10]

With any type of image processing, however, care has to be taken not to create images that differ too strongly from "conventional" radiographs because the pattern against which radiologists compare suspicious findings may be disturbed (see Fig. 6c).

Optimization of Spatial Resolution

Unsharp mask filtering can be used to improve the MTF of a digital system (Fig. 16). It is important that the kernel size and the enhancement factor are optimized with respect to the native detector MTF to avoid over-enhancement that may cause edge artifacts.

Typically very small kernels (3–5 pixels) and enhancement factors of 1.0 to 1.5 are used. Some digital systems incorporate such MTF restoration filter into their standard preprocessing.[16]

Noise Suppression

All processing steps described above will also lead to a relative enhancement of image noise, especially in regions with a low optical density (high x-ray attenuation), such as the mediastinum, chest wall, and upper abdomen. Image noise becomes especially disturbing for low-dose applications or for detector systems that have a low de-

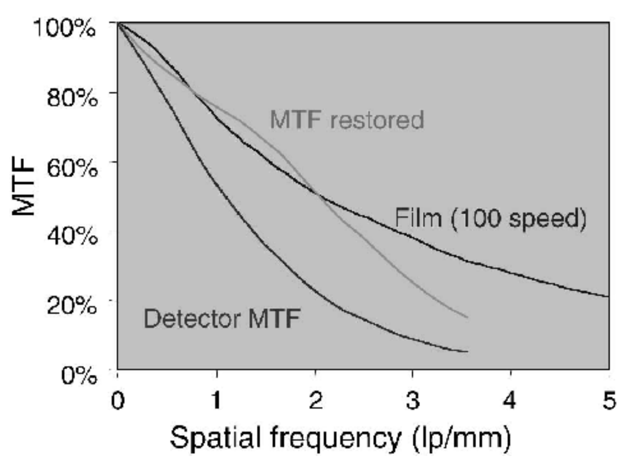


FIGURE 16. The MTF of a digital radiography system can be improved by unsharp mask filtering with a small filter kernel and moderate enhancement (MTF restoration).

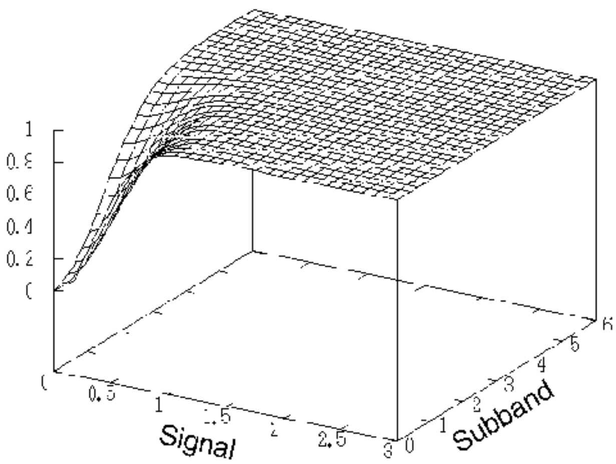


FIGURE 17. A relative noise suppression is possible by reducing contrast enhancement in regions with low signal, and reducing the low-contrast enhancement for high spatial frequencies.

tective quantum efficiency, such as first generation storage phosphor (CR) systems.

Noise suppression is possible by reducing the enhancement in regions with low detector signal. For unsharp mask filtering, the enhancement can be reduced nonlinearly in such regions (eg, β-curve in Fuji-based systems).

For multiscale filtering, the enhancement is reduced for low signals. In addition, the enhancement of low-contrast structures can be reduced for the sub-bands containing high spatial frequencies (Figs. 17 and 18).[17]

PITFALLS

Image processing may lead to a number of undesired effects that can cause an "unnatural appearance" of the chest radiograph but may even cause false positive or false negative interpretations of the images. Processing artifacts may lead to misdiagnoses but in the chest, only the suppression of low-contrast lesions is dangerous in clinical practice. Over-enhancement, on the other side, may cause radiologists inexperienced with the technique to miscall normal vascular markings as intestinal abnormalities.

Processing Artifacts

Processing artifacts are a consequence of selective over-enhancement of certain spatial frequencies or can occur in general with all types of image processing if too strong processing parameters have been chosen.

Edge Artifacts

Unsharp mask filters with too strong enhancement factors may lead to edge artifacts at high contrast edges (overshooting). These artifacts are most disturbing around metal prostheses where they might mimic a loosening process. Since metallic prostheses are uncommon on a chest radiograph, overshot artifacts are most prominent along the contour of the diaphragm, the heart, or the chest wall (as can be seen in the high-pass image Fig. 12c).

Edge artifacts only become disturbing in a chest radiograph if very large enhancement factors are chosen. Only in the first years of computed radiography, when double printouts ("conventional" and "filtered" look) were common, did these artifacts assume (a modest) clinical significance on the filtered image (enhancement factor 5). They then could rarely be confused with a pneumothorax, be it apical or subpulmonary. In general,

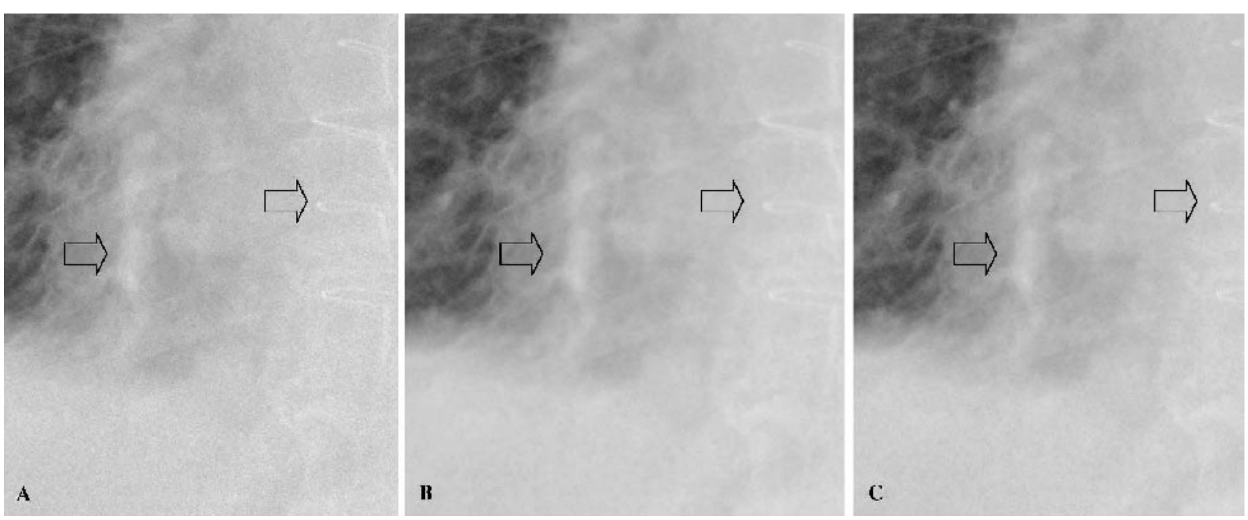


FIGURE 18. "Noisy" processed image (**A**), 100% suppression of enhancement (**B**), 75% noise suppression (**C**) Compare the noise in the regions marked by arrows.

their appearance is so characteristic that no mix-up is possible.

This type of artifact is caused by selective over-enhancement of higher spatial frequencies and can be avoided if only low to moderate enhancement factors are employed for unsharp masking. However, it can be completely avoided with multiscale processing, even if strong enhancement of low-contrast structures is used.

Suppression of Low-contrast Lesions

All structural enhancement also comes with the typical downside that a certain type of structures can only be enhanced if other structures are relatively suppressed. If this is done wisely (eg, by suppressing high-contrast and enhancing low-contrast structures), no diagnostically relevant structures should be lost and less visible structures should be enhanced. Such a techniques relies on nonlinear modification of unsharp masking or multiscale filtering. Most unsharp masking techniques do not rely on structural contrast but on spatial frequency composition and thus size as the criterion for relative suppression or enhancement of a structure. In general, all structures larger than the size of the filter kernel will be suppressed as long as they do not have sharp margins (that contain higher spatial frequencies), while structures (and contours) smaller than the filter kernel will be enhanced.

In the chest this effect is most prominent for the medium size filter kernel (RN = 4, K ≈ 5mm), often still used as the default parameters in the most common (Fuji-based) computed radiography systems. If these medium-size filter kernels in unsharp masking are combined with high enhancement factors, the accentuated structure in the image may mask ill-defined low-contrast lesions, in particular, unsharply marginated pulmonary nodules, interlobar effusions, or small infiltrates (Fig. 15).[13]

The artifacts are no longer too disturbing because the default enhancement of current systems is low (RE = 0.5), thus making the effect less prominent and reducing its clinical significance. It can be completely avoided by changing the default setting of the systems to a large filter kernel (RN = 0) that will enhance most nodular structures and suppress only very large structures.

Over-enhancement

Over-enhancement is caused by choosing too pronounced processing parameters that create an "unnatural" appearance of the filtered image. Over-enhancement can lead to misinterpretation, especially with unsharp mask filtering (see Figs. 6c and 15a), because normal structures may appear overly prominent or may no longer fit the known pattern of normal/abnormal that radiologists have been trained to distinguish. This "normal reference" is constantly refined as the experience of a radiologist grows and will be overthrown with too strong image processing.

This is especially important in chest radiographs, where image interpretation often depends on the detection of faint abnormalities such as increased interstitial markings or unsharply defined pulmonary vessels. Disruption of the normal image appearance will cause insecurity, and may even induce false negative or false positive diagnoses. Such false positives may be caused by over-enhancement of normal vascular structures (eg, by unsharp masking with a medium size filter kernel and larger enhancement factor), which may lead to a false positive diagnosis of an interstitial abnormality. At the same time, increasing the sharpness of vessel definition by unsharp masking may mask a perivascular edema. Even with multiscale filtering, too high a contrast will cause serious problems because any type of enhancement of one structure will invariably lead to relative suppression of another. Over-enhancement of the trabecular structure of the thoracic spine, for example, may suppress normal bone density and simulate osteoporosis.

As a consequence, image processing should be performed in a way to preserve the known appearance of a chest radiograph and to keep the "normal reference" intact. To this end, processing should use no extreme enhancement parameters and should always be kept as low as possible while still achieving the desired effects of MTF restoration, increased transparency of the mediastinum and improved contrast, especially for low-contrast structures.

Noise Enhancement

Image noise is most prominent in regions with high x-ray absorption and unfavorable quantum statistics, such as the upper abdomen, mediastinum, chest wall, and retrocardiac areas. In conventional radiographs, these regions correspond to the lowest portion of the gradation curve, which for these regions has a lower slope than for the lungs. As a consequence, noise is hardly visible. With digital radiographs, processing can enhance the image information in these regions. With the contrast enhancement, however, also comes an enhancement of the stochastic density differences of image noise. Noise, therefore, becomes an unwanted side effect of dynamic range reduction or other techniques that make the mediastinum look more transparent on processed radiographs. There is no loss of information (because noise was also present on the original images) but the prominence of image noise may become annoying and distracting.

For this reason, non-linear techniques for unsharp masking and multiscale filtering have been developed that use less enhancement in high-absorption regions and thus reduce image noise. Multiscale filtering, in addition, offers the possibility to even suppress image noise but these techniques should be used with care in order not to suppress faintly visible structures as well.

SUMMARY

Image processing is an essential part of digital radiography. Its characteristics are summarized in Table 1. All digital chest radiographs have to be processed to ascertain an automated optimization of contrast and density independent of the exposure dose. Digital processing can help to restore and improve spatial resolution even if the intrinsic MTF of the digital detector is limited. Processing is essential to take full advantage of the increased range of signals that can be captured by the detector. Dynamic range reduction is required to restore local image contrast. Digital processing can decouple contrast from latitude, and can simultaneously provide the advantages of a latitude film and a high-contrast film, making a high contrast in the lungs possible while maintaining good visualization of mediastinal structures. Processing can reduce disturbing effects of image noise, but wrong or too strong processing may cause misdiagnoses. Good image processing is characterized by retaining a "conventional look'' while overcoming the limitations of conventional radiography.

TABLE 1. *Image processing in chest radiography—an overview*

Goals of processing:
- To display the full range of attenuation differences in the lungs and mediastinum
- To optimize spatial resolution
- To enhance structural contrast
- To suppress image noise

To do:
- Try to maintain a "conventional look"
- Use dynamic range compression as the basic processing
- Add MTF restoration
- Use only moderate detail-enhancement
- Use the lowest enhancement to achieve the desired effect

How to do it:
- Nonlinear unsharp masking with large kernel or multiscale processing
- Unsharp masking with small kernel (<1 mm)
- Multiscale filtering
- Nonlinear filtering (unsharp mask/multiscale)

Caveats:
- Avoid over-enhancement
- Avoid "edge enhancement" with medium size kernels
- Too much noise suppression induces artifact
- If image noise in the original image is high (eg, for low-dose radiographs or low quantum efficiency), less enhancement is better

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