Abstract
This article starts with an overview of image processing techniques used in storage phosphor based Computed Radiography (CR) systems. Next it elaborates on a selection of image enhancement algorithms. Both the working principles and image quality issues are discussed. The main focus is on multiscale image enhancement, which has become state-of-the-art.

Introduction
Since the early days of CR technology developers have investigated solutions for bridging the gap between the very large dynamic range that characterizes the CR detector and the limited range of the output medium and viewing process. A considerable part of image processing functionality in current CR systems deals directly or indirectly with the issue of manipulating image contrast, in such a way that all relevant image features are rendered to an appropriate level of visibility, despite the restriction of viewing density range. CR equipment manufacturers have adopted basic image processing techniques, or they have developed dedicated solutions.

With this article it is our aim to provide a better understanding of the essential image enhancement techniques of CR systems, what their purpose is, how they operate, and how they affect image quality.

CR image processing OVERVIEW
A simplified diagram of the image processing operations in current CR systems is depicted in Fig. 3.1. The ensemble of operations applied to the stream of image data could be roughly entitled ‘image enhancement’. The role of image processing functions within this data path is to improve the visual quality of the CR image in terms of spatial resolution, sharpness, contrast resolution, dynamic range, SNR. The processing efforts in the main path have to do with maximizing the information transfer to the viewer. The enhancement of image contrast is the main topic of this article and will be elaborated in section 4.

The image processing operations are controlled by parameters, which often are assigned a value in accordance with the examination type. The predefined parameter values are stored in tables, with entries for each examination type. Specification of the examination type is done immediately before or after each exposure by means of an identification terminal, and hence is prone to human errors.

In current CR systems some image processing functions are controlled by internal parameters which are derived from the actual image data. The parameter values are
estimated by heuristic algorithms. This way the kind and degree of enhancement are adapted to the specific characteristics of the image in terms of density levels, dynamic range, noise level, or the presence of irrelevant regions like collimation borders. In most cases a reduced version of the original image is used as input for analysis, since this still contains the essential data for the task at hand. Although the final purpose of the analysis work is very similar among different equipment, most algorithms are proprietary, so that only little is published about their actual implementation.

![Diagram](image)

**Fig. 3.1** Generalized CR image processing flow diagram.

It is one of the crucial functions of CR image processing to extract the signal subrange spanned by the subject out of the full detector dynamic range, which is typically $10^3$ to $10^4$ in terms of exposure. The extracted range is considered to contain all relevant image data and is mapped into the viewing range. This is called signal normalization. Image quality is strongly affected by the proper operation of the subrange extraction algorithms. Contrast resolution will suffer if the extracted range is too large. On the other hand failure to cover the entire diagnostic subrange will cause some relevant image regions to be uniform white or black, since all signal values which exceed the extracted range are mapped to minimum or maximum optical density respectively. The process of subrange extraction is depicted in figure 3.2.

![Diagram](image)

**Fig. 3.2.** Subrange adjustment and gradation mapping in Agfa ADC.
The actual signal subrange (between the dashed lines) that corresponds to the relevant body portions is found by analysis of the histogram of signal values. The selected subrange is mapped into a quantity that drives the film printer or display monitor (in optical density units). This mapping is usually nonlinear. The shape of the mapping curve is designed to optimize the contrast rendering across the entire density range.

The current algorithms for subrange extraction may be fooled by the presence of low-density borders in collimated exposures, as shown in figure 3.3. For that reason the subrange extraction algorithm must confine its analysis exclusively to histogram data from the region of interest. Some CR systems provide advanced analysis software for finding the region of interest automatically. In addition these algorithms are able to detect multiple regions of interest in case of side-by-side exposures on a single imaging plate.

Fig. 3.3. In various examinations the X-ray beam is collimated in order to minimize the irradiated portion of the body. Image analysis techniques are applied to find the regions of interest without human intervention. A more extensive description of the imaging chain of CR systems is presented in 1.

**Image Contrast ENHANCEMENT**

This section concentrates on the question how image processing contributes to improving CR image quality.

The dynamic range of CR detectors is very large in comparison with the available density range of viewboxes or display screens under normal ambient light conditions. Without image processing this mismatch would cause a significant additional loss of pictorial information in the CR imaging chain. It will be shown in the next subsections how special enhancement techniques in addition to adjusting global density, contrast, and gradation can help to maximize the transfer of image information to the viewer.

**Edge enhancement**

The perceptibility of edges and small features can be improved by raising the amplitude of high spatial frequency components in the image, e.g. by a two-dimensional high-pass filter. Unsharp masking is the earliest and best-known method used in CR systems. In its basic form it is formulated as:
where \( X \) represents the pixel values of the original image, \( Y \) the resulting image, and \( \bar{X} \) a smoothed version of the original image obtained by a moving average operation. The difference image \( (X - \bar{X}) \) represents the high spatial frequency content of the image, and the enhancement factor \( \alpha \) determines how much of this component is added to the final image. In commercial CR equipment the enhancement factor is made data-dependent for better control:

\[
Y = X + \alpha \cdot (X - \bar{X})
\] (4.1)

In this case the factor \( \alpha \) determines the degree of enhancement, and the function \( \beta(X) \) adapts the degree of enhancement to the local image density. The type of function can be selected in order to match the typical examination requirements. This way, it is possible e.g. in chest images to suppress the enhancement in areas of low penetration such as the mediastinum, in order to avoid amplification of noise there.

The bandwidth of the spatial frequency band to be stressed is determined by the size of the filter kernel. If a small kernel size is specified, then all signal components of high spatial frequency are emphasized relative to medium and low-frequency image components. In case of a larger kernel size also the medium frequency components will be amplified. This is made clear by the curves plotted in fig. 4.1, which represent the frequency response of the basic unsharp masking filter (4.1) for different kernel sizes \( m \) (in pixels), given by:

\[
F_{\alpha}(f) = 1 + \alpha \cdot \beta(X) \cdot \left(1 + \frac{\sin(\pi \cdot m \cdot f)}{\pi \cdot m \cdot f}\right)
\] (4.3)

with the spatial frequency \( f \) expressed in cycles per pixel. The asymptotic amplification factor for high frequencies is \( 1 + \alpha \), and the lowest frequency at which the response reaches this value is given by \( f_c = 1/m \). In general, unsharp masking will stress any features smaller than the filter kernel. So, if it is the purpose only to sharpen the edges, then the kernel should be chosen slightly larger than the spatial extent of the edge transition. A feature can be thought of as consisting of a left edge followed by a right edge, hence the combined effect of emphasizing both edges is feature enhancement.
Although these filters are easily characterized in the frequency domain, it is not obvious to specify the most appropriate kernel size value for different examination types. Small kernel diameters are reported to be appropriate for improving image sharpness and the visibility of fine linear details, but at the same time the high-frequency portion of the noise spectrum is boosted, yielding a typical fine-grain appearance. A shortcoming of medium-kernel enhancement consists in hiding pathologic lesions that do not have contours to be enhanced, like pulmonary nodules. In case of a large kernel only large low-contrast objects run the risk of being suppressed. It is also advised to apply enhancement only to a mild degree, since extended low-contrast structures with ill-defined borders may be suppressed even with the largest available kernel size.

The fine-grain appearance caused by noise amplification in a narrow frequency band can be reduced with the density-dependent version of unsharp masking. This is achieved by selecting an enhancement curve $\beta(x)$ that avoids significant edge enhancement in areas of low penetration such as the mediastinum, where noise is most disturbing. The drawback, however, is that tips of catheters which are typically located in those areas are not accentuated neither.

Edge enhancement must be applied with caution, since severe enhancement may introduce artifacts. The amplification of high-frequency components of large amplitude step-like edges gives rise to the so-called “rebound” or “overshoot” artifact, in which the overenhancement creates a white or black line alongside of the actual edge. In addition, this rippling may obscure a low-contrast lesion in the immediate vicinity of a high-contrast edge.

**Dynamic range reduction**

The irradiation range of pelvic or shoulder images may be too large to display all image regions with sufficient contrast resolution. Unsharp masking with a very large kernel size may be applied for improving the contrast of all relevant image features by reducing the relative contribution of the very low frequency components in the image. It is assumed that the latter does not carry significant information.

$$y = (x - \bar{x}) + \alpha(\bar{x})$$  \hspace{1cm} (4.4)

with \hspace{1cm} $\alpha < 1$ \hspace{1cm} (4.5)
Here the blurred image $\hat{X}$ is obtained by a sliding average operator with a very large kernel (up to 255 pixels).

The expression (4.4) consists of a term in which the low-frequency component (i.e. the background) has been removed, and the second represents the background contribution. The first term corresponds to the band of medium to high spatial frequencies, and it carries all the relevant image features. This term passes without modification. The background term, on the other hand, is compressed.

**Multiscale contrast enhancement**

Many variants have been developed of the methods for edge enhancement and dynamic range reduction discussed so far. These techniques have in common that they all rely on a spatial neighborhood operator, in most cases a sliding average filter. As already mentioned above, there is quite some controversy in the medical imaging community about the selection of the kernel diameter. This question has no simple answer, and in many cases the optimum choice is only a compromise. In the above methods the image data are split into two different channels according to spatial frequency. Contrast is modified according to a different mechanism in each channel. This is not optimal, since in most images relevant features have frequency components across large parts of the spectrum. Small image features like edges and micronodules entirely belong to the high end of the spectrum. Intermediate and large-sized features like large nodules and anatomic structures normally cover the range of medium to high frequencies. Gross density variations caused by the transition between different body parts and background regions contribute mostly to the low end of the frequency range. A priori it is dangerous to neglect part of the frequency spectrum, since subtle lesions may exist at any scale. Detecting large but faint opacities may be equally important as finding tiny fractures or nodules.

The multiscale approach presented in this section does no longer adopt the object size or spatial frequency as a criterion for controlling the amount of enhancement, but rather the *radiation* contrast of the feature. Some image details are visible by themselves, others have very subtle contrast and are easily overlooked. This qualification applies to features of any size or *scale*, e.g. edges, textures, compact details, large structures or opacities. Visualization is generally improved by amplifying the contrast of subtle image features, and at the same time attenuating the strong components without the risk of omitting information. This is done *irrespective* of feature size, hence the term multiscale. It is the basic paradigm of multiscale contrast equalization, commercially known as MUSICA in the Agfa ADC system. The acronym stands for “MUlti Scale Image Contrast Amplification”.

**The multiscale representation**

The basic idea in multiscale enhancement is to decompose the image into components which represent individual details, and to improve the contrast by immediately operating on these components rather than on the original image. With a linear transform the image $X(i,j)$ is decomposed into a weighted sum of two-dimensional basis functions $A(i,j)$, and each transform coefficient $b_{k,l}$ represents the relative contribution of the corresponding basis function to the original image:

$$X(i,j) = \sum_{k,l} b_{k,l} \cdot A(i,j)$$ (4.6)
A linear transform with periodic basis functions such as the Fourier transform is not suited for the purpose of contrast enhancement since the basis functions extend across the whole image plane. Basis functions must be compact and localized in spatial domain in order to match individual features. Furthermore, they must encompass all scales in order to represent details of various sizes. Also they must be continuous, otherwise discontinuities in the resulting image will be introduced with any modification of the transform coefficients. Several instances of the wavelet transform fulfill the above criteria, except for the Haar transform, which does not meet the criterion of continuity. In MUSICA the image is decomposed according to the Laplacian pyramid transform\textsuperscript{11}.

![Fig. 4.2. Laplacian pyramid decomposition and inverse transform](image)

The decomposition and inverse transform are schematically depicted in Fig. 4.2. The original image is low-pass filtered with a 5x5 Gaussian kernel and subsampled by a factor of two. Next the intermediate result is interpolated to the original image size, and pixel-wise subtracted from the original image. This subband image is the finest Laplacian pyramid layer. The next layers are analogously computed starting from the intermediate subsampled image of the previous stage. This way, both image dimensions are halved each time. Decomposition goes on until a subsampled image comprising only one pixel, is obtained. This represents the DC component of the image.

In case of a $2k \times 2k$ image the pyramid comprises 11 layers. Any low-resolution copy of the original image is available as one of the gaussian smoothed images $g_i$ in the course of decomposition. In MUSICA the reduced image $g_4$ which is needed by the ROI finding algorithm is obtained this way.

The amount of detail is reduced at each subsequent stage of the pyramid decomposition. The difference between consecutive blurring operations is stored in the corresponding layer of the Laplacian pyramid. These difference images each represent pictorial detail corresponding to a specific scale. In the spatial frequency domain each pyramid layer corresponds to an octave of the original spectrum. The subbands have a considerable overlap as can be seen in Fig. 4.8.
The inverse transform is sketched in the right part of Fig. 4.2. This reconstruction process proceeds in reverse order starting from the largest scale image, which consists of a single pixel. Interpolation is carried out in order to magnify the image according to the dimensions of the next layer. Then the detail information associated with the current scale is read from the corresponding pyramid layer and added. This process of magnification and detail addition is repeated until an image of original size is obtained. The result is identical to the original image, if the interpolation filters used in the course of decomposition are identical to the filters of the inverse transform.

The Laplacian pyramid is a complete representation of the original image. Moreover, it is redundant, since there are 4/3 as many transform coefficients as there are pixels.

Fig. 4.3. Gaussian basis functions of the 2nd through 4th layers of the Laplacian pyramid transform. The finest scale layer (not drawn) consists of identical pulses of unit width at each pixel location. Basis functions overlap, and are uniformly spread across the image plane.

The basis functions have a Gaussian profile with limited spatial extent, there is a partial overlap, and the ensemble encompasses the entire spatial domain at all scales, as can be seen in the plots of Fig. 4.3. These properties are essential in view of the basic purpose of the decomposition; i.e. to analyze the image into components that can be numerically manipulated in order to modify the contrast of image features on an individual and local basis.
Contrast equalization

Contrast improvement is achieved by modifying the coefficients of the Laplacian pyramid. Small coefficients represent subtle details. These are amplified in order to improve the visibility of the corresponding details. The strong density variations on the other hand have a major contribution to the overall dynamic range, and these are represented by large-valued coefficients. They can be reduced without risk of information loss, and by compressing the dynamic range, overall contrast resolution will improve. The entire process of multiscale decomposition, contrast equalization and reconstruction is pictorially exemplified in Fig. 4.4.

Contrast is equalized by applying the following nonlinear amplification to the transform coefficients of pyramid layers:

$$y(x) = a \frac{x^\gamma}{|x|}$$  \hspace{1cm} (4.7)
The coefficients $x$ are normalized to the range $[-1, 1]$. The factor $a$ is needed for rescaling the resulting image to the original dynamic range. The exponent $p$ controls the slope of the amplification curve, and hence also the amount of contrast enhancement when the image is reconstructed by applying the inverse transform to the modified pyramid coefficients. The required sigmoid shape for equalization is obtained if the exponent is chosen $p < 1$. This ensures that the smaller values are amplified relative to the larger ones. Best results are obtained in the range 0.7 through 0.85. Experience has learned that there is no need for applying stronger enhancement beyond this range, since this will not reveal any additional information, but only increase the noise impression. The curves are plotted in Fig. 4.5 and the corresponding pictures after reconstruction of a hip examination in Fig. 4.6. The leftmost picture corresponds to the original image, which results from setting the exponent to one.

![Graph](image)

Fig. 4.5. Nonlinear amplification applied to the pyramid coefficients. $p = 1.0, 0.7, 0.5$.

![Images](image)

Fig. 4.6. Left: original hip image ($p = 1$). Middle, right: multiscale contrast equalized images ($p = 0.7, p = 0.5$).
The most notable effect of multiscale contrast equalization is the uniformly improved visibility of subtle features throughout the image, without really diverging from the original ‘look’. Sharpness increases, but also low-contrast opacities benefit from improved rendition. The enhancement is noticeable in poorly penetrated areas like the mediastinum, but in the lungs as well. In skeletal examinations also the soft tissue regions are properly visualized.

Like with other contrast enhancement techniques the noise is amplified simultaneously with image details. However, multiscale contrast enhancement is not affected by the fine-grain appearance typical of edge enhancement.

Enhancement techniques based on a sliding neighborhood operator such as unsharp masking show peculiar behavior in zones of strong density transitions. The width of the transition zone is exactly the kernel radius. This regular appearance is probably what makes the rebound artifact conspicuous. Multiscale contrast equalization on the other hand is more robust against step responses because all frequency bands are manipulated in a similar way. There is essentially no transition zone, since all scales are involved. Years of experience with clinical images have confirmed that MUSICA does not suffer from the rebound artifact at the edges of metallic implants or at interfaces between bone and soft tissue.

In ADC systems multiscale contrast equalization is used for all examination types because it combines the benefits of improved contrast with natural look and the absence of artifacts.

**Frequency processing**

The multiscale representation is very suited for implementing conventional filters such as edge enhancement or low-frequency attenuation. The latter is used for reducing the dynamic range, and is further referred to as latitude reduction. In fact, any frequency response can be easily synthesized by appropriately weighing the Laplacian pyramid coefficients according to the layer to which they belong, since each layer is associated with an octave of the spatial frequency spectrum.

In MUSICA contrast equalization is the basic mode of enhancement, and in most examination cases it is the only mode. If additional edge enhancement or latitude are required, they are realized by concatenation. Starting from the Laplacian pyramid coefficients, contrast equalization (4.7) is applied to all layers. Next, each layer is multiplied by scale-dependent factors $a_{ek}$ and $a_{lk}$ for edge enhancement and latitude reduction respectively. A diagram of a few layers of the combined multiscale enhancement flow is shown in Fig. 4.7.

![Diagram of multiscale enhancement](image)

**Fig. 4.7.** Combined multiscale enhancement, including contrast equalization, edge enhancement and latitude reduction.
Edge enhancement is implemented by multiplying the pyramid coefficients of the small-scale layers with a scale-dependent factor $ae_k$:

$$ae_k = f \left( \frac{(k-n_e)}{n_e} \right), \quad 0 \le k < n_e$$

$$ae_k = 1, \quad k \ge n_e$$

in which $f_e$ is the parameter that controls the degree of edge enhancement at the finest scale, i.e. at which the scale index $k$ is zero, and $n_e$ is the number of pyramid layers (octaves) to which edge enhancement is applied. This means that the enhancement increases at a rate of $\sqrt[k]{f_e}$ per octave, equally distributed among the $n_e$ octaves of the high end of the spectrum. With this gradual filter characteristic it is possible to minimize rebound in the vicinity of steep density transitions.

In a similar way, for carrying out latitude reduction the pyramid coefficients of the large-scale layers are multiplied by a scale-dependent factor $al_k$ defined by:

$$al_k = 1, \quad k < L - n_l$$

$$al_k = f \left( \frac{(L-k-n_l)}{n_l} \right), \quad L - n_l \le k < L$$

In this case $n_l$ of a total of $L$ layers are involved, and in the corresponding spatial frequency band the contrast is reduced at a rate of $\sqrt[k]{f_l}$ per octave. The amount of latitude reduction is controlled by the parameter $f_l \ge 1$.

In MUSICA the number of layers involved in edge enhancement $n_e$ is three, and for latitude reduction $n_l$ is five. With contrast equalization (MUSI-contrast) as the basic enhancement mode the added value of edge enhancement is rather limited. As a matter of fact, most of the edges in an image are weak, and hence these will already be accentuated by contrast equalization. Some additional sharpening is used in a few examination cases such as extremities, with moderate enhancement (edge contrast parameter equal to 3, which corresponds to $f_e = 1.7$). Stronger edge enhancement is not useful in most cases, and it will unnecessarily raise the noise level.

Latitude reduction is applied in a few examinations, like shoulder, in which there exists a significant density transition across the image. A moderate reduction factor is suitable, $f_l = 1.4$, since contrast equalization by itself reduces the dynamic range considerably where needed. This is because the large density variations across the image, which contribute most to the dynamic range, are attenuated most.

The effects of edge enhancement and latitude reduction in the spatial frequency domain are plotted in Fig. 4.8. The dash-dotted lower curves represent the frequency response of the Laplacian pyramid layers, from DC through the finest scale. Together these responses sum to unity, which corresponds to perfect reconstruction starting from the unmodified pyramid. The solid curve is the frequency response of edge enhancement by a factor 1.7, and the dashed curve corresponds to latitude reduction by 1.4. These responses are equal to the weighted sum of the individual layer responses, with weight coefficients given by (4.8) and (4.9) respectively. The effect of contrast equalization is not taken into account here. The spatial frequency domain is not suited for demonstrating the effect of contrast equalization, because the latter is essentially a nonlinear operation carried out within
each frequency band, whereas edge enhancement and latitude reduction are linear operations.

![MTF graph](https://example.com/mtf.png)

**Fig. 4.8.** Spatial frequency response of edge enhancement (solid) and latitude reduction (dashed). The dash-dotted curves below represent the frequency responses of the Laplacian pyramid layers, from DC (left) through the finest scale (right).

The heel images of Fig. 4.9 demonstrate how contrast equalization, edge enhancement and latitude reduction each affect image quality in a particular way. Exaggerated parameters settings have been chosen so that the differences are clearly visualized. Enhancement factors, gains and offsets have been adjusted in such a way that the processed images have comparable fine-scale contrast, and approximately the same signal range within the region of interest. This can be verified on the density profiles. The edge-enhanced image (top right) shows strong emphasis of the very fine details, but leaving the spikes out of consideration, the density profile is very close to the original.

![Heel images](https://example.com/heel_images.png)

**Fig. 4.9.** Comparison of enhancement modes. Top left: original heel image. Top right: edge enhancement. Bottom left: contrast equalization. Bottom right: latitude reduction. The density profile across the center line is plotted on top of each image.
The rebound effect is also clearly visible as the false edge at the collimation border. Contrast equalization (bottom left) shows improved contrast of details and features of any size, both in bone and soft tissue regions. This is also exemplified by the density profile, which emphasizes all subtle short-, mid- and long-range variations of the original curve. There is no rebound artifact despite the strong enhancement.

A similar effect can be observed in the latitude-reduced image (bottom right), in that the contrast of small and medium-sized features has improved, but not to the same degree as with contrast equalization, and not in all image regions. Best results are found in the joint region, but the toes in the upper image half remain invisible. The relative contrast improvement is obtained by suppressing the long-range increase of density from left to right. This corresponds to a lowering of the slope on to the density plot, with a simultaneous amplification of density variations along the profile. Unlike contrast equalization, latitude reduction tends to ‘flatten’ the images, in that it may remove density gradients that exist across adjacent macroscopic regions.

**Conclusion**

The multiscale contrast enhancement techniques presented here have proven to be very valuable in rendering medical X-ray images. The perceptibility of low-contrast features of arbitrary sizes is improved throughout the image, without creating artifacts.

The concept has a general scope, and because of its apparent benefits it is being evaluated in other medical modalities as well, such as computed tomography and digital mammography, and in non-medical application areas, such as non-destructive testing.

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**References**


