

Chapter 5

Current Post-Processing Methods in Digital Radiography

Digital radiographic images (DR) are acquired by a variety of means, and the acquisition of these types of images was detailed in Chapter 4. After the DR image has been acquired, a series of steps must be undertaken prior to the image being visualised on a computer monitor. These pre-display processes depend upon the type of DR system.

Direct and indirect DR systems usually undergo more pre-display processes than those required in computed radiography (CR). These additional processes compensate for detector faults.

Several pre-display processes common to all DR techniques are undertaken prior to the image being displayed (Seibert, 1999). A typical first step is to locate the image area within the image matrix. Image detectors, such as imaging plates in CR, and the image acquisition process control the maximum size of the DR image. In many cases the x-ray field is collimated to a size smaller than the DR image matrix. Locating the image in the detector matrix is achieved by finding the borders of the x-ray collimation area. The area inside the collimated area is included in further image analysis. An example of this edge localisation is provided in Figure 5.1 (Seibert, 1999).

On occasions, radiographic technique may require multiple image exposures on a single plate prior to image processing and display. Radiographers preset the number of exposures per plate into the DR system. This information is also used to assist in edge localisation.

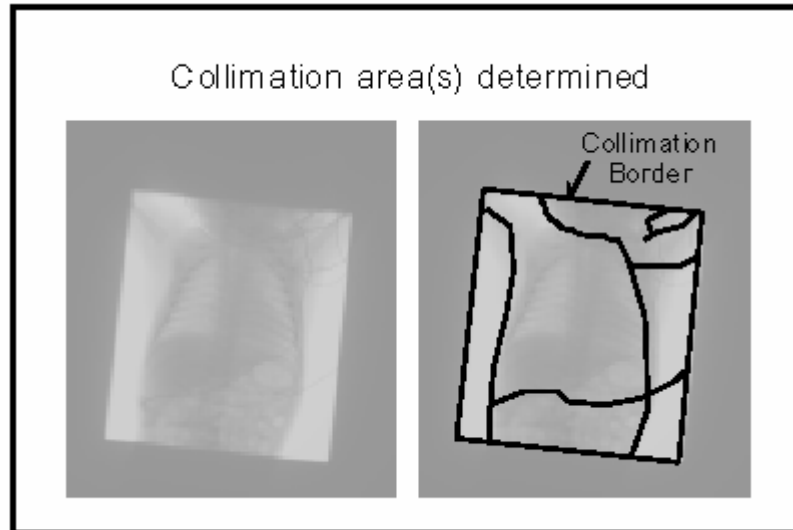


Figure 5.1 Localisation of edges of x-ray collimation (Seibert, 1999, p.4)

Once the area of x-ray exposure has been localised, histogram analysis of the exposed area can be undertaken. A histogram is produced by computing the frequency of the pixel values and is displayed as the frequency or number of pixels versus the pixel values (Baxes, 1994). The histogram is used to define pixel values that represent unattenuated x-ray exposure and pixel values that represent areas of non-exposure. An example of such a histogram is provided in Figure 5.2 (Seibert, 1999). Following histogram analysis, the range of useful signal levels can be identified.

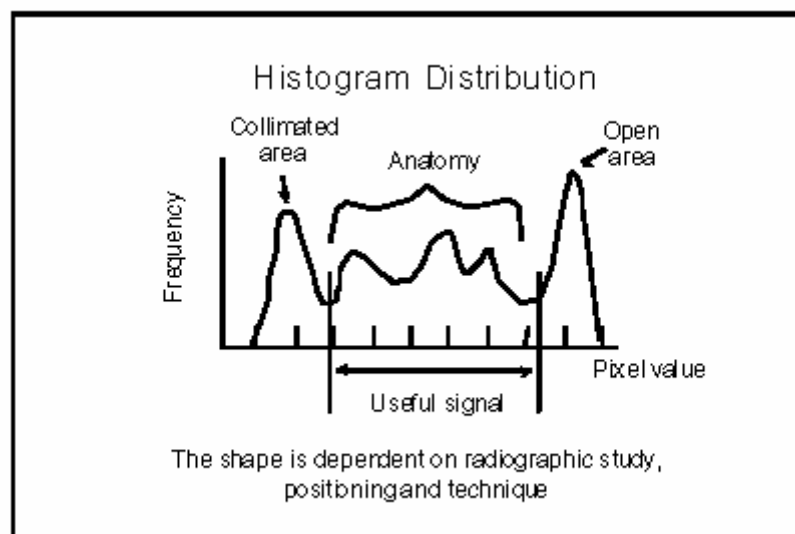


Figure 5.2 Histogram distribution of pixel values from the image plate (Seibert, 1999, p.5)

Prior to the examination being undertaken, the radiographer defines the anatomical region for the radiographic exposure and stores this information in the DR system. Predetermined processing functions based on anatomical regions are used in conjunction with the useful signal range to determine the final pixel values in the image. DR images are usually 12 bits deep. This implies that there are 4096 (2^{12}) possible values for each pixel. Possible pixel values range from 0 to 4095. The pixel values of the image are then inverted so that areas of high x-ray exposure appear black and areas of low x-ray exposure appear white. This process produces a negative image that is typical in medical imaging. Comparison of a paediatric chest x-ray with and without full pre-processing functions is shown in Figure 5.3 (Seibert, 1999).

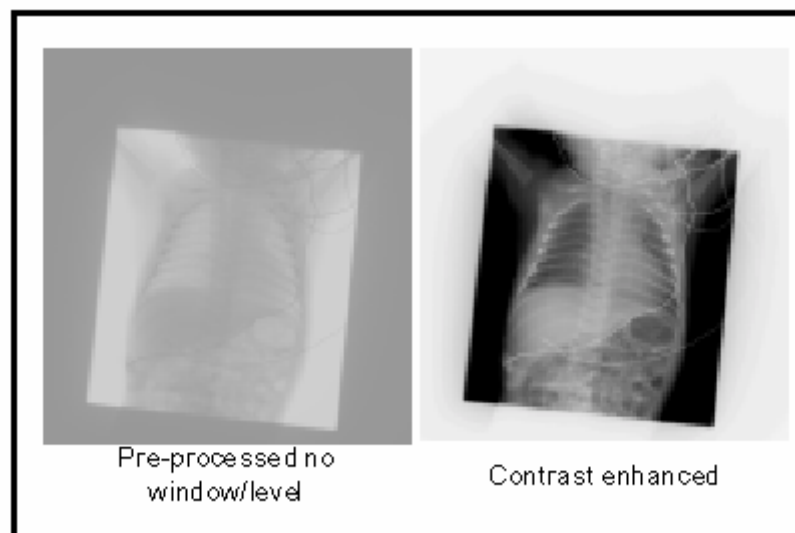


Figure 5.3 Comparison of a paediatric chest x-ray with pre-processing (right) and without pre-processing functions applied (left) (Seibert, 1999, p.5)

Once the pixel values are determined, the DR image is stored as an image file. The standard for DR images and other medical images is the Digital Imaging and Communications in Medicine (DICOM) file format. The DICOM file format is part of a larger group of DICOM standards. These standards were established by the American College of Radiologists (ACR) and the National Electrical Manufacturers Association (NEMA). The DICOM standards define transfer and storage protocols,

define objects such as “patient” and “study”, and define services such as “find”, “get” and “store”. These protocols and objects provide a standard method for transferring and storing images across medical imaging modalities and images produced by different manufacturers (Bushberg *et al*, 2001; DICOM: Digital Imaging and Communications in Medicine, 2003; DICOM: The value and importance of an imaging standard, 2004).

A DICOM image file, like other image file formats, has two main sections. These are the image pixel values and the file header. An image file header contains information such as the size of the image (number of rows and columns), type of image (grey scale or colour), compression details and where the first pixel value is located. The DICOM image file header also contains other information such as patient demographics, type of examination, examination factors, type of equipment used, where the examination was performed, and date and time of examination, and may also contain clinical indications and findings (Clunie, 2004).

There are many digital image processing (DIP) operations that may be undertaken when viewing DR images. These are based on general DIP operations. General DIP operations for viewing of images fall under three broad categories: contrast and brightness enhancement, perceived spatial enhancement, and image resizing.

5.1 General Digital Image Processing – Image Resizing

Image resizing is an operation that enlarges or reduces the size of the displayed image. There is little standardisation of terminology used in this area. Baxes’ (1994) and Gillespy & Rowberg (1994) used “image scaling”, Schowengerdt (1997) used “reduce/expand” and Gonzalez & Woods (1992) used “zooming”. The two main purposes of resizing images are to increase the size of the object within the image for ease of visualisation and to decrease the size of the image so that the entire image can be visualised on a monitor or display device with a lower spatial resolution than the image itself.

During the process of resizing, image pixels are mapped onto a smaller or larger image matrix. Various methods to achieve the remapping of pixels are used. The common methods are nearest neighbour interpolation and bi-linear interpolation. Another more time-consuming method is bi-cubic or cubic convolution. Detailed discussion of these methods can be found in Gonzalez & Woods (1992), Baxes (1994), Russ (1994), and Castleman (1996).

The other DIP operations of contrast and brightness enhancement and perceived spatial enhancement are more relevant to this work. The developed radiographic contrast-enhancement mask (RCM) algorithms use contrast and brightness enhancement processes to achieve their desired effect. Some perceived spatial enhancement methods were created to have a desired effect of dynamic range compression similar to the RCM algorithms. These methods are discussed in more detail later in this chapter.

5.2 General Digital Image Processing – Contrast and Brightness

Image processing operates in either the spatial domain or another transform domain such as the frequency domain. Contrast and brightness adjustments are more easily undertaken in the spatial domain.

Contrast and brightness adjustments in a displayed image are achieved through point operations or processes. This is a global process. In a global process, all pixels in the image have the same operation applied equally across the image. Point operations can be considered as linear or non-linear.

Point operations of contrast and brightness adjustments alter each pixel value in the image without any influence on neighbouring pixel values. Point operations typically start in the top left corner of the image matrix. A new pixel value results from an operation on the original pixel value. This new value is mapped to the same (x, y) spatial location in a new image matrix of the same size. The operation is then shifted to the next pixel and repeated until the end of the row. The operation is repeated on the pixels in the next row until all the pixels in the image have undergone the point

operation (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994). The general formula for point operations is given in Equation 5.1. The new or output image $O(x, y)$ results from an operation on each of the pixels in the image $I(x, y)$.

$$O(x, y) = p \circ I(x, y) \quad \dots\dots\dots 5.1$$

where: $O(x,y)$ is the new or output image;

$I(x,y)$ is the original image;

$p \circ$ is the point operation that is applied to each pixel.

Examples of linear point operations are addition and multiplication. The point operations of contrast and brightness adjustment in medical imaging are commonly called window width (WW) and window level (WL) (Bushberg, 2002; Seeram, 2001). Non-linear point operation examples are exponential or logarithmic functions (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994).

New pixel values result from point operations. The new image, $O(x, y)$, can be saved as a new file or displayed for viewing on a monitor or other such display device. Linear point operations of addition/subtraction alter the brightness in the image. Linear point operations of multiplication/division alter the displayed contrast.

Look-up tables (LUTs) are typically used to map pixel values to the display device. A LUT can be considered as a two-column matrix. The left column consists of all possible pixel values in $I(x, y)$. If $I(x, y)$ is a 12 bit depth image, the left column in the LUT will have values from 0 to 4095. The point operation is performed on these values and the results are stored adjacent to the original value in the right column. The use of LUTs reduces the time required for point operation calculations, especially when the operation is complex and the image is large. Typically LUTs are displayed graphically with the X axis as the input values and the Y axis as the output values (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994).

Figure 5.4 is an example of three LUTs for 8 bit depth data. The solid line represents an operation of multiplying the existing pixel values by 1 and the dotted line represents multiplying the existing pixel values by 1.5. Image contrast in $O(x,y)$ will differ between these two examples. Note that clipping has occurred at the maximum pixel value of 255 for the operation of multiplying by 1.5. Clipping occurs when the output value exceeds the bounds of the bit depth of the new image. These values are “clipped” to the maximum or minimum pixel value as appropriate. The contrast can be inverted with the use of a LUT. In Figure 5.4, the dashed line represents a LUT that will invert the contrast in the image.

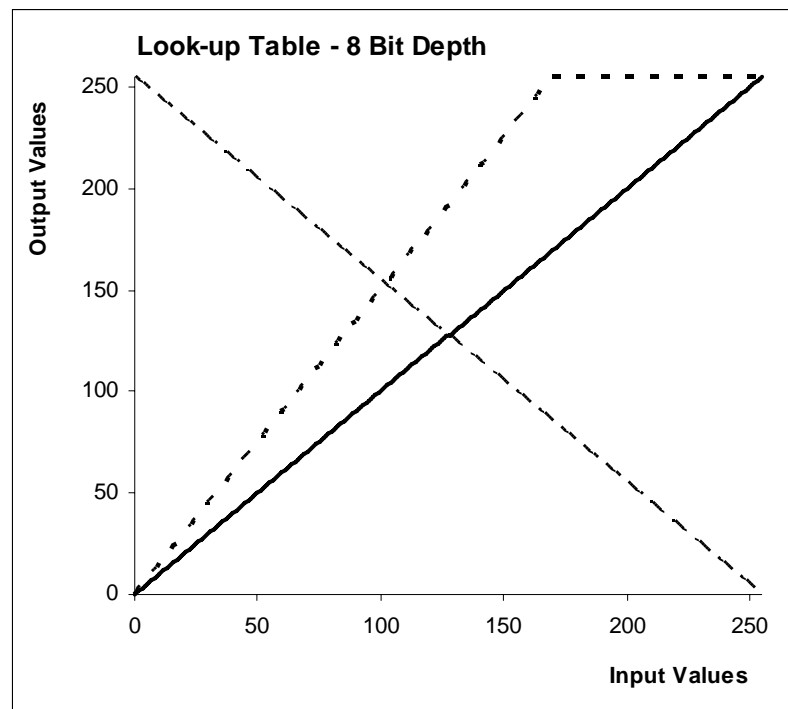


Figure 5.4 Look-up table for contrast adjustment

Dotted line: multiply by 1.5;

Solid line: multiply by 1;

Dashed line: invert image

Figure 5.5 shows examples of a point operation of addition which result in image brightness adjustment. The dotted line represents an operation of subtraction of 80 from the existing pixel values, which produces an image that appears darker than the original. The solid line is an operation of adding 25 to the existing pixel values, which produces an image that appears lighter than the original (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994).

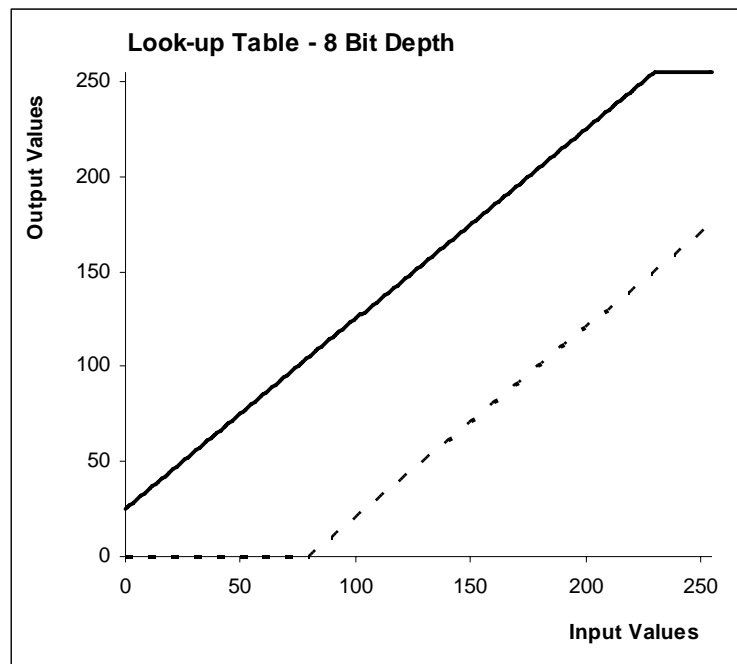


Figure 5.5 Look-up table for brightness adjustment

Dotted line: addition of - 80;

Solid line: addition of 25

An example of a LUT of WW and WL is shown in Figure 5.6. WW is a measure of the range of the input values that will be mapped to output values within the dynamic range of the new image. The WW value provides an indication of the displayed contrast of the image. Large values of the WW imply low displayed contrast and small values of WW imply high displayed contrast. Pixels that have input values above the WW range will appear white and those with input values below the WW range will appear black. WL is the midpoint value in the WW, which provides an indication of the brightness of the displayed image (Bushberg, 2002; Seeram, 2001;). The dotted line LUT represents a WW of 128 and a WL of 66. The solid line LUT represents a WW of 196 and a WL of 150. In this example there would be differences between the output images that result from the use of these LUTs in both displayed brightness and displayed contrast.

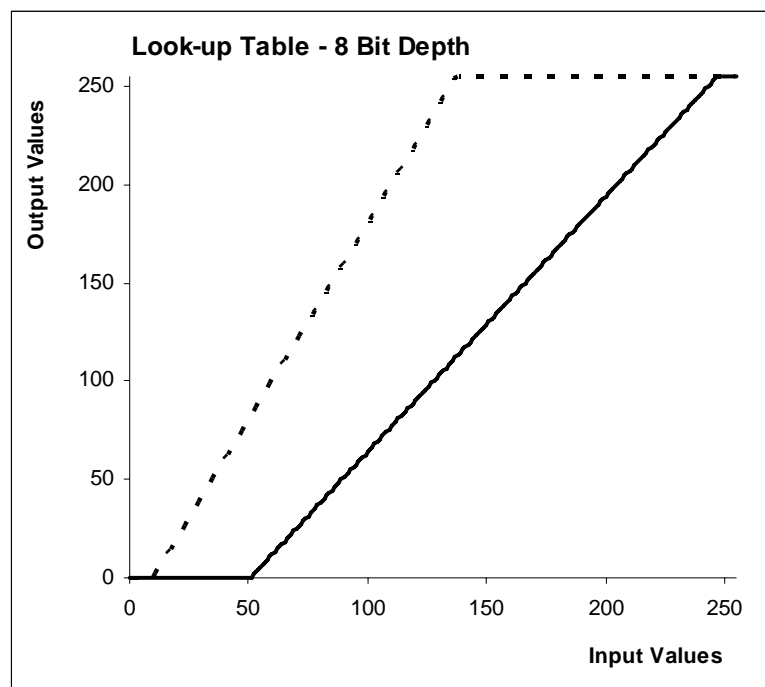


Figure 5.6 Look-up table with examples of window width and window level

Dotted line: WW = 128, WL = 66;

Solid line: WW = 196, WL = 150

Figure 5.7 provides examples of non-linear LUTs. The solid line represents a LUT with an exponential function and the dotted line represents a LUT with a logarithmic function. The use of an exponential function enhances displayed contrast in the brighter areas of the image whereas the logarithmic function enhances displayed contrast in the darker areas of the image. Use of a non-linear LUT will allow high displayed contrast in some areas of the image while other areas have low displayed contrast.

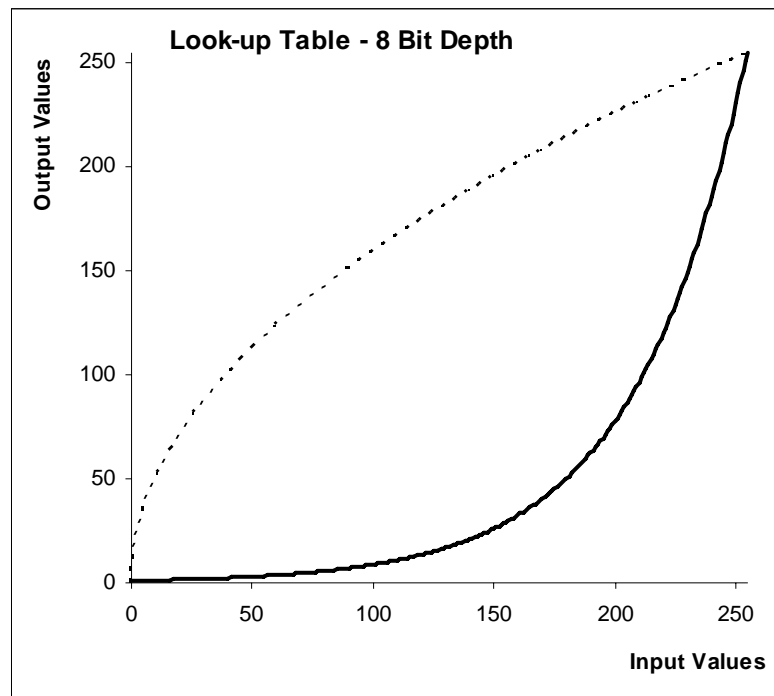


Figure 5.7 Look-up table with examples of non-linear functions
Dotted line: Logarithmic function;
Solid line: Exponential function

5.3 Digital Radiography – Contrast and Brightness

In medical imaging the WW and WL functions are common means of adjusting the displayed contrast and brightness of images. WW and WL functions are more commonly used in computed tomography (CT) and magnetic resonance imaging (MRI) than in DR (Bushberg, 2002; Seeram, 2001).

In DR contrast and brightness adjustment, complex LUTs are used, their prime purpose being to display DR images with similar appearances to film/screen (F/S) radiographic images. S-shaped LUTs are commonly used in DR. A common desired result is that the S-shaped LUT matches the S-shaped characteristic curves of F/S and produces results that are similar in appearance to F/S images (Barski *et al*, 1998; Computed radiography: Advanced processing capabilities, 2000; Freedman & Artz, 1997b; Huda *et al*, 1996; Van Metter & Foos, 1999a, 1998b). A typical S-shaped LUT is shown in Figure 5.8.

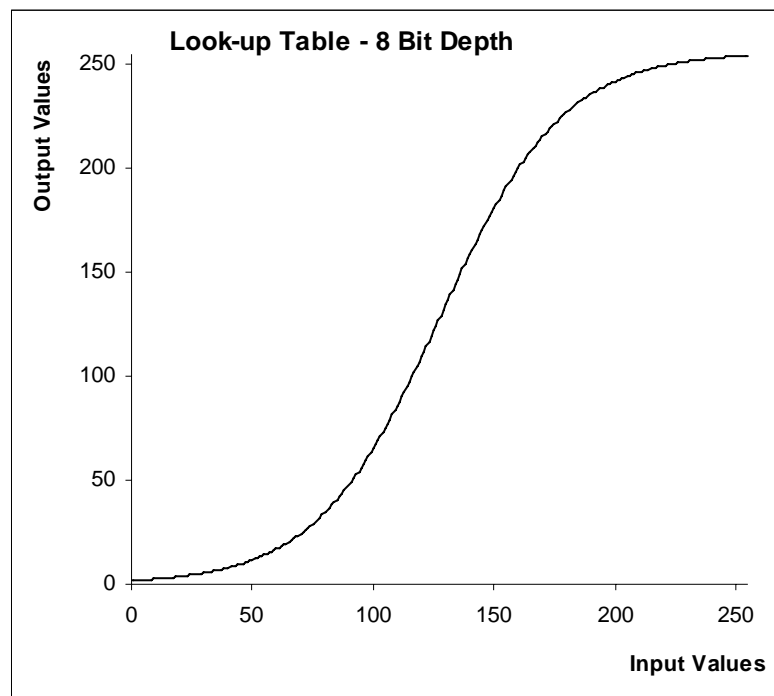


Figure 5.8 Look-up table with an example of an S-shaped function

The three major DR equipment manufacturing companies, Fuji Medical, Agfa-Gevaert and Kodak, differ in their approach to contrast and brightness adjustment. Fuji Medical allows users to adjust the shape of the S-shaped curve through adjustment of parameters that control the curve shape or gradient type (GT), the curve's rotation amount or gradient angle (GA), the curve's rotation central point or gradient centre (GC), and the brightness adjustment or the curve's gradient shift (GS)

(Barski *et al*, 1998; Computed radiography: Advanced processing capabilities, 2000; Freedman & Artz, 1997b; Huda *et al*, 1996).

Figure 5.9a & b (Computed radiography: Advanced processing capabilities, 2000) shows the range of S-shaped curves that are available to users of the Fuji Medical systems. These are user-selectable gradient types.

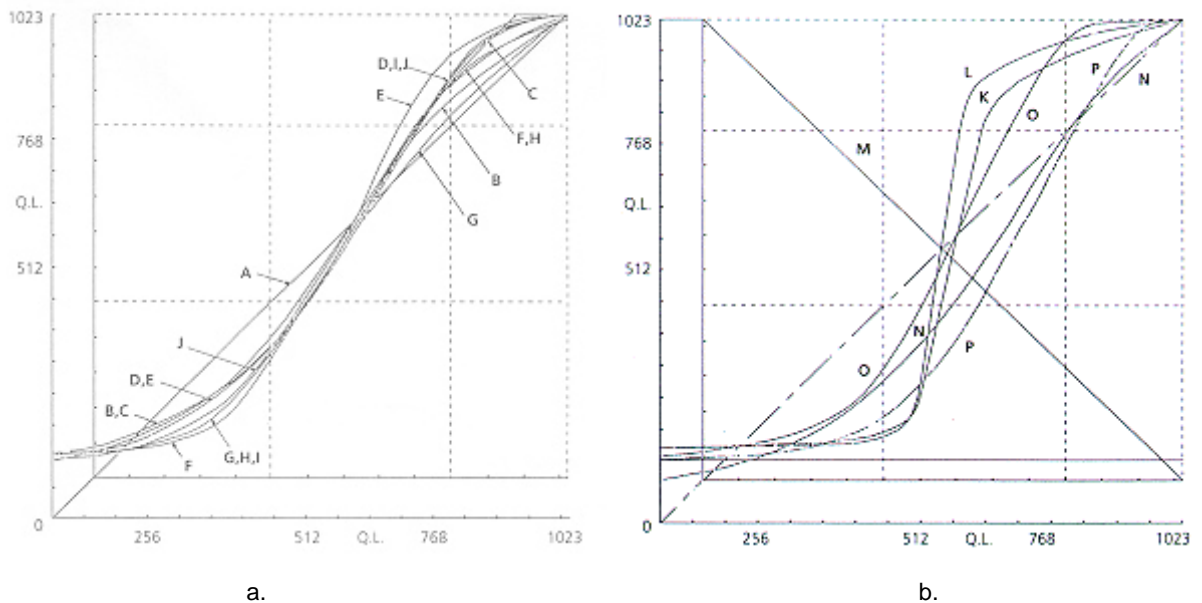
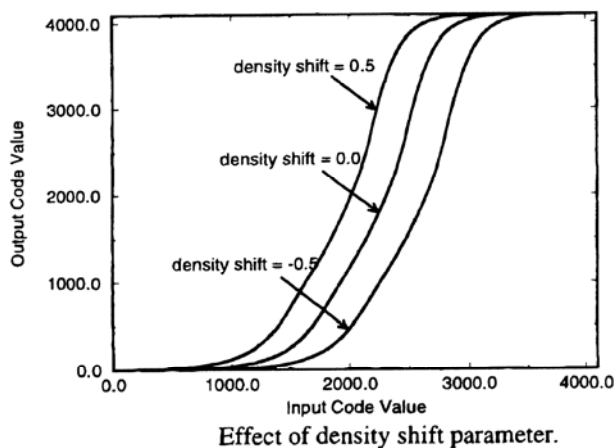
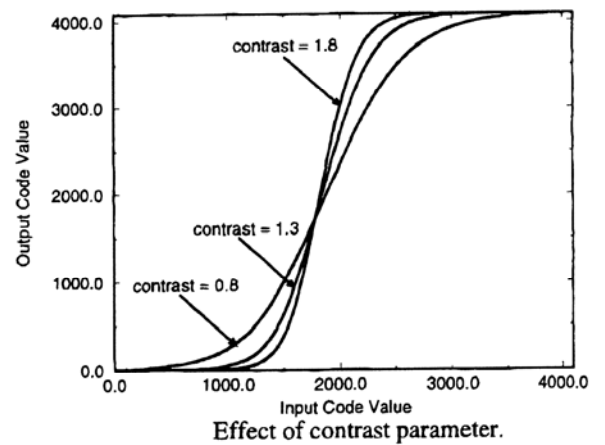


Figure 5.9 Look-up table S-shaped curves gradient types offered by Fuji Medical (Computed radiography: Advanced processing capabilities, 2000)

Other manufacturers' controls are simpler. The common contrast adjustment among the three major manufacturers is the slope of the straight line portion of the curve and is typically called gamma, γ . The terminology is the same as that which is used in F/S imaging. Similar to WW, γ controls the displayed contrast in the image. Increasing the γ increases the displayed contrast of the image. Conversely, decreasing γ decreases the displayed contrast. Brightness of the image is controlled through shifting the curve to the right along the X-axis or to the left along the X-axis of a LUT plot (Barski *et al*, 1998; Van Metter & Foos, 1999a). Figures 5.10a & b (Barski *et al*, 1998) show the brightness adjustment (Figure 5.10a) and contrast adjustment (Figure 5.10b) that can be made through user alteration of density shift parameters and contrast parameters respectively.



a.



b.

Figure 5.10 Look-up table S-shaped curves offered by Kodak:
 a. Brightness adjustment; b. Contrast adjustment
 (Barski *et al*, 1998, p.169)

Specific LUTs can be applied automatically to an image. This can be advantageous in that the need for manipulation of brightness and contrast can be reduced. Sailer *et al* (2004) compared the use of automatically applying specific LUT curves for given patient types and a semi-automated approach with user input when viewing chest DR images. The γ of the LUT was automatically applied depending upon the patient size being above or below a defined weight. The automated approach was preferred when the lateral projection of the chest was undertaken. The use of the semi-automated approach was preferred when viewing the posterior-anterior (PA) projection of the chest. This was due to the large density differences that exist within this projection of the chest image.

5.4 Histogram Equalisation

A method of automatically adjusting displayed brightness and contrast in digital medical images is adaptive histogram equalisation. This method is based on classic histogram equalisation. Classic histogram equalisation uses the histogram of the entire image. A histogram is a frequency distribution plot of pixel values in the image. An example of a histogram, that of the DR image of a foot in Figure 2.4 b. (8 bit depth for display and printing purposes), can be seen in Figure 5.11. Classic histogram equalisation modifies the pixel values such that the new histogram contains uniform frequency across all potential pixel values in the image. Using classic histogram equalisation, all pixel values are equally represented in the histogram-equalised image. Figures 5.12a, b and c provide examples of histograms before and after histogram equalisation. The main purpose of histogram equalisation is to use an automated process to adjust for contrast differences in the image. Following the use of histogram equalisation, the displayed contrast of the new image is lower than that of the original image (Castleman, 1996).

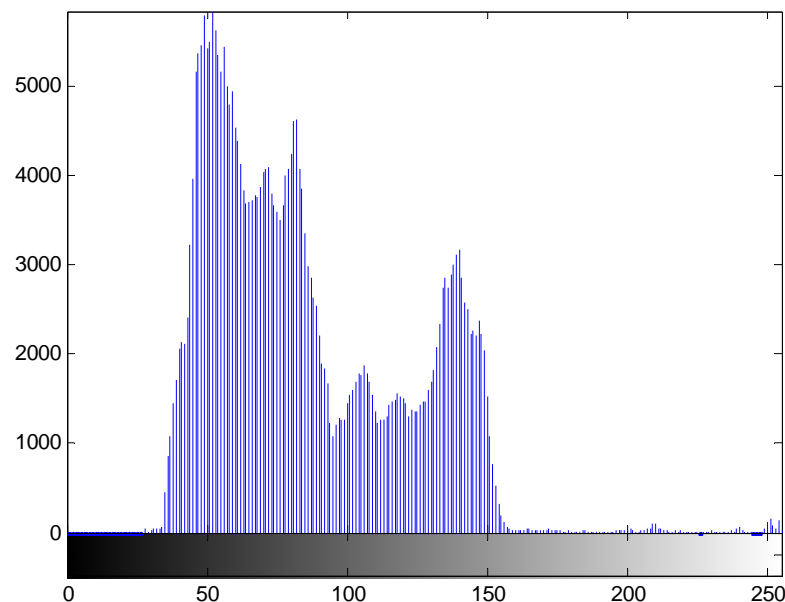
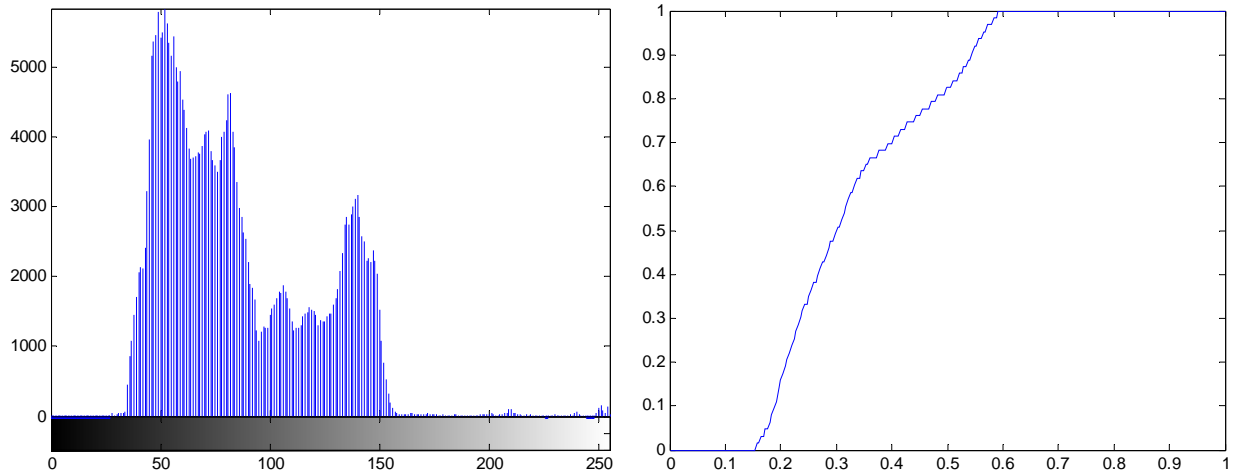
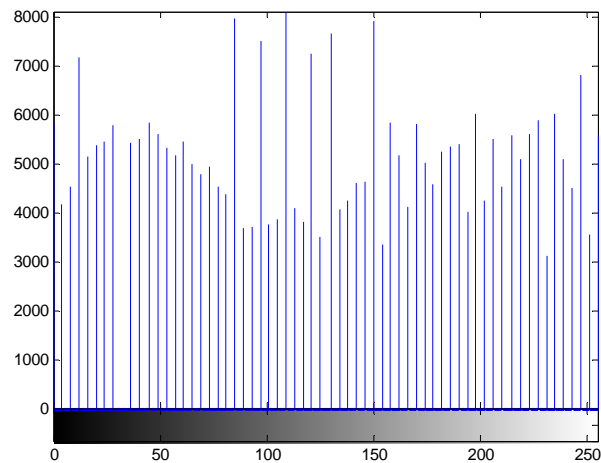


Figure 5.11 Histogram of Figure 2.4b (an 8 bit depth image)



a.

b.



c.

Figure 5.12 Histogram equalisation: a. Original histogram (Figure 5.11); b. Non-linear equalisation LUT; c. Final histogram after histogram equalisation operation

Adaptive histogram equalisation is more locally based than classic histogram equalisation. Adaptive histogram equalisation uses the values of surrounding pixels to alter the local values and hence local contrast. This method has been reported for use in CT examinations (Gillespy & Rowberg, 1994). The use of adaptive histogram equalisation has also been reported in digital mammographic examinations (Pisano *et al*, 1998, 2000; Santos *et al*, 2002; Sivaramakrishna *et al*, 2000).

Freedman & Artz (1997b) reported that adaptive histogram equalisation was used by Fuji Medical and Agfa-Gevaert in DR images. They noted that adaptive histogram equalisation is called dynamic range control by Fuji Medical and MUSICA processing by Agfa-Gevaert. However, as discussed later in this chapter, these processes are actually different from adaptive histogram equalisation. No other literature discussing adaptive histogram equalisation in DR, other than in mammography, has been found.

The adaptive histogram equalisation method can produce unexpected results and is therefore not generally popular in non-mammographic medical imaging examinations (Bushberg *et al*, 2001; Gillespy & Rowberg, 1994). Chuang, Chen & Hwang (2001) reported on loss of definition of the edges of the object and over-enhancement of noise in the images when adaptive histogram equalisation is used in digital imaging. Possible reasons for its non-use in DR are the unexpected results of the displayed image and lack of user control of the displayed contrast and brightness of the image.

5.5 Spatial Enhancement

Spatial enhancement is undertaken to improve the perceived spatial information within the image or to change the spatial features displayed in the image. This can be achieved through operations undertaken in the spatial domain or in other transform domains such as the frequency domain. Operations undertaken in the spatial domain usually have a corresponding frequency domain operation that will produce the same or a similar effect on the image. Some desired effects on the image, such as specific types of noise reduction, are best undertaken in the spatial domain and others are best undertaken in other domains. For example, periodic noise is best reduced in the frequency domain whereas so-called “salt and pepper” noise is best reduced by spatial domain operations. The choice of domain under which the operation will be performed is often dependant on the efficiency of the operation in that domain (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994).

Enhancement operations in the spatial domain are based on operations being performed on a local neighbourhood of pixels of the original image. A single new pixel value is the result of the local neighbourhood operation. The concept of spatial enhancement in spatial domain processing, using a 3x3 kernel, is illustrated in Figure 5.13 (Bushberg *et al*, 2001).

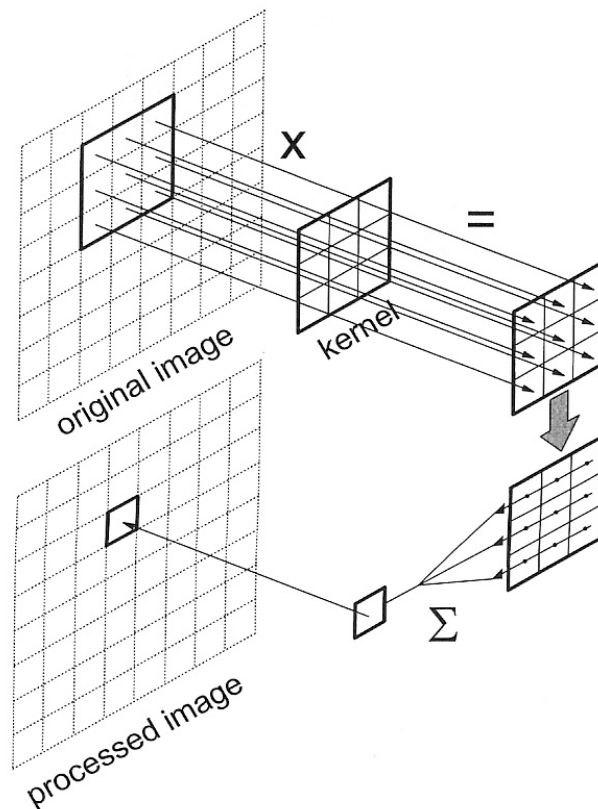


Figure 5.13 Concept of spatial enhancement in the spatial domain using a 3x3 kernel (Bushberg *et al*, 2001, p.312)

The process of spatial domain enhancement is the convolution of a kernel, of size greater than 1x1, with the image. Pixel values in a local neighbourhood are multiplied by values in the kernel. The outcomes of these multiplications are summed to form a new pixel value. The new pixel is located in the new image at the location corresponding to the centre of the neighbourhood of the original image. The convolution process usually starts in the top left of the image. The kernel is moved

one pixel to the right and repeated until the end of the row is reached. The process is further repeated by moving the kernel one row below the original starting point and repeated until all pixels in the new image have a new value (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994). The general form of spatial domain enhancement operations is shown in Equation 5.2.

$$O(x, y) = k(a, b) * I(x, y) \quad \text{..... 5.2}$$

where: $O(x,y)$ is the new or output image;

$I(x,y)$ is the original image;

$k(a,b)$ is the convolution kernel.

Types of spatial enhancements that may result from such convolutions are smoothing or blurring of features in the image or so-called low pass filtering, enhancement of the edges of features in the image or so called high pass filtering, and detection of edges of features in the image leaving only the edges visible and the background suppressed, also called high pass filtering (Baxes, 1994; Castleman, 1996; Gonzalez & Woods, 1992; Jain, 1989; Russ, 1994).

The type of spatial enhancement that results from the convolution depends upon several factors relating to characteristics of the kernel. The values of the kernel determine whether low or high pass filtering occurs. If all the values in the kernel are greater than zero, low pass filtering results. If both positive and negative values exist within the kernel, or the kernel has a so-called zero cross-over, high pass filtering results. Contrast in the new image can also be altered with the convolution process. If the sum of the kernel values is 1, the overall contrast in the new image will be similar to that of the original image. When the kernel's summed value is other than 1, the contrast of the image will differ from that of the original image. The resulting contrast will be similar to that which would result from a point operation of the same value. If the sum of the kernel values is zero, background suppression will result, leaving only the edges of features visible. Examples of different kernels are shown in

Figure 5.14. Image examples using the kernels in Figure 5.14 are shown in Figure 5.15.

0.111	0.111	0.111
0.111	0.111	0.111
0.111	0.111	0.111

a.

-1	-1	-1
-1	9	-1
-1	-1	-1

b.

-1	-1	-1
-1	8	-1
-1	-1	-1

c.

Figure 5.14 Various convolution kernels of size 3x3

- a. smoothing filter;
- b. edge enhancement filter, sum total = 1;
- c. edge detection and background suppression filter, sum total = 0.

The size of the kernel also affects the spatial enhancement of the new image. Increasing the size of the neighbourhood of the original image that is used in the convolution process increases the effect of the contribution of the neighbouring pixels. In low pass filtering, increasing the size of the kernel increases the amount of smoothing of the features within the image. In DR imaging, the size of the filter is typically given in millimetres rather than the matrix size.