CHAPTER 14

NUCLEAR MEDICINE IMAGE DISPLAY

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14.1. INTRODUCTION

The final step in a medical imaging procedure is to display the image(s) on a suitable display system where it is presented to the medical specialist for diagnostic interpretation. The display of hard copy images on X ray film or photographic film has largely been replaced today by soft copy image display systems with cathode ray tube (CRT) or liquid crystal display (LCD) monitors as the image rendering device. Soft copy display requires a high quality display monitor and a certain amount of image processing to optimize the image both with respect to the properties of the display device and to some psychophysiological properties of the human visual system. A soft copy display system, therefore, consists of a display workstation providing some basic image processing functions and the display monitor as the intrinsic display device. Display devices of lower quality may be used during intermediate steps of the acquisition and analysis of a patient study. Display monitors with a quality suitable for diagnostic reading by the specialist medical doctor are called primary devices, also known as diagnostic devices. Monitors with lower quality but good enough to be used for positioning, processing of studies, presentation of images in the wards, etc. are referred to as secondary devices or clinical devices.

Nuclear medicine images can be adequately displayed even for diagnostic purposes on secondary devices. However, the increasing use of X ray images on which to report jointly with images from nuclear medicine studies, such as those generated by dual modality imaging, notably by positron emission tomography (PET)/computed tomography (CT) and single photon emission computed tomography (SPECT)/CT, requires display devices capable of visualizing high resolution grey scale images at diagnostic quality, i.e. primary display devices. Both grey scale and colour display devices are used, the latter playing an important role in the display of processed nuclear medicine images and in the display of overlaid images such as from registered dual modality imaging studies.

Owing to the advances of picture archiving and communication systems (PACSs), the location of a display device which used to be adjacent to the gamma camera or PET scanner now widely varies, and can, for example, be in special reporting rooms, patient wards and operating rooms. An important requirement for display devices connected to a PACS is that, regardless of the display device and the imaging modality used, the display of an image should be consistent in appearance and presentation among the monitors and printers used at different locations and under different environmental lighting conditions. A special standard arising from this need was created specifying the requirements to display grey scale images on different display devices [14.1]. For colour displays, similarity in visual appearance is achieved by using an industry standard colour management system (CMS) [14.2]. The requirements for inclusion of a CMS into a PACS are addressed in Ref. [14.3].

The technology of soft copy display devices is presently experiencing a rapid transition from CRT based monitors to LCD technologies, the latter potentially offering better image quality, improved stability, reduced weight and reduced costs.

Regardless of the technology, display hardware needs to operate at maximum performance and needs to deliver consistent and reproducible results. Quality assurance of a display device, both at the time of installation and at regular intervals during its lifespan, ensures stability and optimum performance, and should be considered an important component of the quality system in nuclear medicine.

14.2. DIGITAL IMAGE DISPLAY AND VISUAL PERCEPTION

Emissive display devices such as CRTs or LCDs display a soft copy of an image. Hard copies consist of transmissive film or of reflective prints, both produced by printing devices.

The digital image to be displayed is represented as a rectangular matrix of integer values. Each matrix element corresponds to a pixel of the image. Regardless of the modality, the original intensity values as produced by the acquisition device are integer values with a pixel depth of between 8 to 16 bits. For display, the original images are stored in the memory of a computer, the display workstation. The soft copy display system of a workstation consists of the monitor which produces the visible image on the screen, and of the associated display controller, also known as a graphics card or video card, which holds the intensity values and carries out the conversion of the digital values to the analogue signals which control the electronics of the display device.

Originally, there is no colour in the image, but pseudo-colours are routinely used in nuclear medicine images to improve the visibility or to emphasize special features within an image. Another common use of colour is with overlay display techniques of pairs of registered images, e.g. originating from dual modality acquisitions such as PET/CT or SPECT/CT. Colour display devices, therefore, play an increasingly important role in the visualization of medical images.

14.2.1. Display resolution

The resolution of a digital display device is commonly described in terms of the number of distinct pixels in each dimension that can be displayed. Displays are designed in such a way that individual pixels have an approximately quadratic shape. The basic question arising is how many pixels are needed for adequate visualization. The desirable lower limit of pixel size would be reached if the human eye would not be able to distinguish between the individual pixels which make up the screen. Assuming a spatial resolution of the human eye of 1 arc minute and an average reading distance for an X ray image of 65 cm, a human observer would not be able to discern two adjacent pixels as being different for pixels smaller than about 0.18 mm. A modern 3 megapixel (MP) colour LCD monitor with a screen diagonal of 540 mm (~21 in) has pixel matrix dimensions of 1536 × 2048, resulting in a pixel size of 0.21 mm which is close to the resolution limit of the eye, so that individual pixels are almost indistinguishable. Monitors of this quality are already used routinely in radiology as primary display devices except for mammography. A 5 MP LCD monitor of the same screen size would have pixel matrix dimensions of 2048 × 2560 and linear pixel dimensions of 0.17 mm. Monitors with such high resolution are accepted as primary devices for reading digital mammography images. Limits of display resolution can be overcome by magnification and interpolation techniques included as standard features in the display workstation.

LCD monitors are composed of a large number of liquid crystals. This number represents the ‘native’ resolution of the display device since each pixel can be addressed individually to change brightness and colour. The display of an image is best if each pixel of the image maps to a pixel of the screen. If the mapping requires interpolation of the image pixels to the screen pixels, the image loses sharpness. A CRT display, in contrast, can change screen pixel size without loss of image sharpness by changing deflection and modulation frequencies. Several display resolution values can, therefore, be used with equal sharpness.

A display issue specific to nuclear medicine arises from the fact that the matrix dimensions of the original acquired or reconstructed image are much smaller than the display resolution would permit to display. The small original image size is due to both the poor spatial resolution of a nuclear medicine

imaging device and its noise characteristics. As a rule of thumb, the sampling size used for a nuclear medicine image preserves the information content in the image when it is approximately one third of the full width at half maximum (FWHM) of the spatial resolution of the acquisition system. Thus, a scintillation camera equipped with, for example, a high resolution collimator, a system spatial resolution of 8 mm FWHM and a field of view of 540 mm × 400 mm would require a matrix size of at most 256 × 256 pixels to preserve the information content transmitted by the camera. Even a current commercial off the shelf (COTS) display device, on the other hand, has minimum pixel dimensions of 1024 × 1280. Displaying the original image matrix at native pixel resolution would result in an image too small for visual interpretation. It is, therefore, essential for the image to be magnified to occupy a reasonable sized part of the available screen area. Straightforward magnification using a simple interpolation such as the 'nearest neighbour' interpolation scheme would result in a block structure with clearly visible square elements which strongly interfere with the intensity changes created by the true structure of the object generating artefacts in the interpretation. Magnification is necessary and can be done without artefact generation by a suitable interpolation algorithm that generates smooth transitions between screen pixels while preserving the intensity variations within the original image. It is the task of the display workstation to provide software for visualizing this type of image.

14.2.2. Contrast resolution

This refers to the number of intensity levels which an observer can perceive for a given display. It is referred to as perceived dynamic range (PDR).

Brightness refers to the emitted luminance on screen and is measured in candelas per square metre (cd/m^2). The maximum brightness of a monitor is an important quality parameter. Specifications of medical display devices also include the calibrated maximum brightness which is lower but is recommended for primary devices to ensure that the maximum luminance can be kept constant during the lifespan of the display device. Typical values for a primary device LCD monitor are 700 and 500 cd/m^2 for the maximum and the calibrated maximum luminance, respectively.

The dynamic range of a display monitor is defined as the ratio between the highest and the lowest luminance (brightness) a monitor is capable of displaying. The dynamic range is highest if measured in the absence of ambient light. It is then called contrast ratio ($CR = L_H/L_L$) and is the figure usually quoted by vendors in the specifications. A typical CR of a grey scale primary LCD monitor is 700:1, measured in a dark reading room. If luminance values are measured with ambient light present, which is the scenario in practice, CR is replaced by the luminance

ratio (LR = L'_H/L'_L), which is the ratio of the highest and the lowest luminance values including the effect of ambient light. It can be considerably smaller than the CR, since the effect of ambient lighting is added as a luminance L_{amb} to both the minimum and the maximum luminances. The CR is related to the PDR, but its potential usefulness as a predictor of monitor performance suffers from the lack of standardized measurement procedures and the effect of ambient light. The dark room CR is a common performance parameter quoted by manufacturers.

The PDR is the number of intensity levels an observer can actually distinguish on a display. It can be estimated based on the concept of just noticeable differences (JNDs). The JND is the luminance difference of a given target under given viewing conditions that the average human observer can just perceive. The measured JND depends strongly on the conditions under which the experiment is performed, for example, on the size, shape and position of the target. The PDR is defined as the number of JNDs across the dynamic range of a display. The PDR for grey scale displays has been assessed in Ref. [14.4] to be around a hundred. The number of intensity values which a pixel of the digital image can hold is much higher. The pixel of an image matrix is usually 1 or 2 bytes deep. It is an integer number between 256 and 65 536, and is given by the pixel depth of the image matrix. It is a further task of the display system to scale the original intensity values to a range compatible with the performance of the human observer. It is common to use 256 intensity values to control the brightness of a display device as this is sufficient to produce a sequence of brightness levels perceived as continuous by the human observer.

Colour displays using pseudo-colour scales can extend the PDR to about 1.5 times that of a grey scale display. This has been demonstrated for a heated-object scale which has the additional advantage of producing a ‘natural’ image [14.4]. Owing to the enormous number of possible colour scales and the fact that the majority of them produce ‘unnatural’ images, the concept of JNDs, while valid in principle, cannot be transferred directly to colour displays.

14.3. DISPLAY DEVICE HARDWARE

14.3.1. Display controller

The display controller, also known as a video card or graphics card, is the interface between the computer and the display device. Its main components are the graphical processing unit (GPU), the video BIOS, the video memory and the random access memory digital to analogue converter (RAMDAC). The GPU is a fast, specialized processor optimized for graphics and image processing operations. The video memory holds the data to be displayed. The capacity of

current COTS graphics cards ranges from 128 MB to 4 GB which is sufficient to store even a large sequence of images. This permits the rapid change between images on the screen for, for example, cine display, rapid browsing through a sequence of tomographic slices or the use of overlay techniques to display simultaneously additional information such as text, markers or regions of interest without having to modify the image data. The output signals generated by the RAMDAC depend on the monitor type. For CRT monitors, the RAMDAC generates analogue positioning and intensity signals and additional control signals for the deflection and synchronization of the electron beam. The output of the graphics card is via a video graphics array connector. For current LCD displays, the graphics card provides standardized digital output signals via the digital visual interface connector which avoids image distortion and electrical noise, and can be configured to use the native spatial resolution of the LCD display directly.

Lookup tables (LUTs) constitute a crucial part of the video memory because they play an important role in the implementation of intensity transformations and in the display of colours. An LUT contains the digital values which are converted by the RAMDAC to the analogue intensity values driving the monitor. These values are called digital driving levels (DDLs). The range of DDLs is typically 8 bit, i.e. from 0 to 255. For medical displays, the range can be larger, up to 12 bit. The maximum value contained in an LUT produces the maximum brightness the screen can display. Since there is some latitude in the maximum brightness a screen can display, it is adjusted by the monitor hardware controls or firmware, and in practice follows manufacturer's recommendations to ensure optimum performance with respect to both image quality and life expectancy of the monitor. The luminance values generated by the sequence of available DDLs (e.g. 0…255) produce the characteristic curve of the display device.

The intensity values of an image stored in video memory are mapped to the values in an LUT. The mapping transformation associates each intensity with an LUT index. In the case of a colour display, a triple of DDLs, one each for the three primary colours red, green and blue, is used for a pixel. The LUT consisting of triples of primary DDL values is referred to as a colour lookup table (CLUT). For a medical image, which by nature is a grey scale image, a pseudo-colour image is generated by a mapping transformation which associates the intensity value of an image pixel to an appropriate index of a CLUT.

14.3.2. Cathode ray tube

The CRT is a vacuum tube containing an electron gun and a fluorescent screen (Fig. 14.1). The electron gun produces electrons which are focused by a system of magnetic and electrostatic lenses into a pencil beam. The electron

beam is accelerated by a positive high voltage applied to the anode towards the fluorescent screen. The screen is covered with a crystalline phosphorescent coating that produces a visible light spot when hit by the electron beam. The light distribution of the spot is given by a 2-D Gaussian function. The directional dependence of the intensity of the emitted light follows Lambert's cosine law, implying that the apparent luminance of the screen is independent of the viewing angle. Using the deflecting coils attached to the collar of the tube, the beam is made to scan the screen area in a rectangular pattern. At the same time, the intensity of the electron beam is controlled by the control grid, thereby producing different light intensities. The digital image matrix containing the numerical intensity values is visualized by synchronizing scanning motion and intensity modulation given by the LUT to produce an intensity pattern, the visual image, on the screen. The elements of the matrix occupy a rectangular grid on the screen, with the luminance of the centre of each grid point corresponding to the LUT value of the corresponding matrix element.

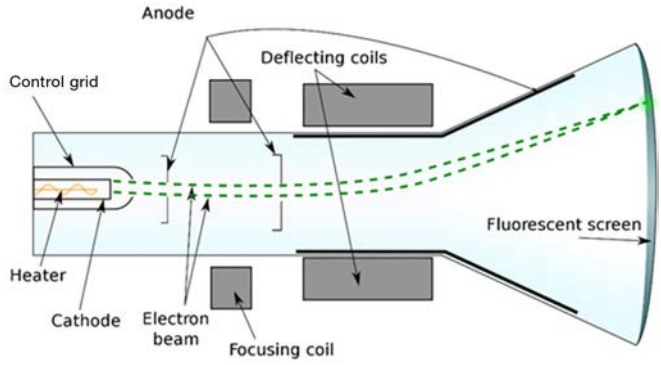

FIG. 14.1. Diagram of a cathode ray tube.

Colour CRTs use three different phosphors which emit red, green and blue light, respectively. The phosphors are packed together in clusters called 'triads' or in stripes. Colour CRTs have three electron guns, one for each primary colour. Each gun's beam reaches the dots of exactly one type of phosphor. A mask close to the screen absorbs electrons that would otherwise hit the wrong phosphor. The triads or stripes are so small that the intensities of the primary colours merge in the eye to produce the desired colour.

14.3.3. Liquid crystal display panel

An LCD display panel consists of a rectangular array of liquid crystal cells in front of a light source, the backlight. Each cell acts as a tiny light valve which transmits the backlight to an extent determined by a voltage applied to the liquid crystal. The image on the screen is formed by applying voltages to each cell separately, thereby modulating the light intensity into the desired intensity pattern.

A typical liquid crystal cell (Fig. 14.2) consists of a liquid crystal in twisted nematic phase between two glass plates, G, coated with alignment layers (not shown) that precisely twist the liquid crystal by 90° when no external field is present (left diagram). Light from the backside is polarized by polarizer P_1 and rotated by the crystal structure. The second polarizer P_2, set at 90° to P_1, then permits the light to pass. If a voltage is applied to the two transparent electrodes, E1 and E2, the nematics realign themselves (right diagram) and the polarized light is blocked by P_2. Partial realignment is achieved by varying the voltage and permits the transmitted intensity to vary.

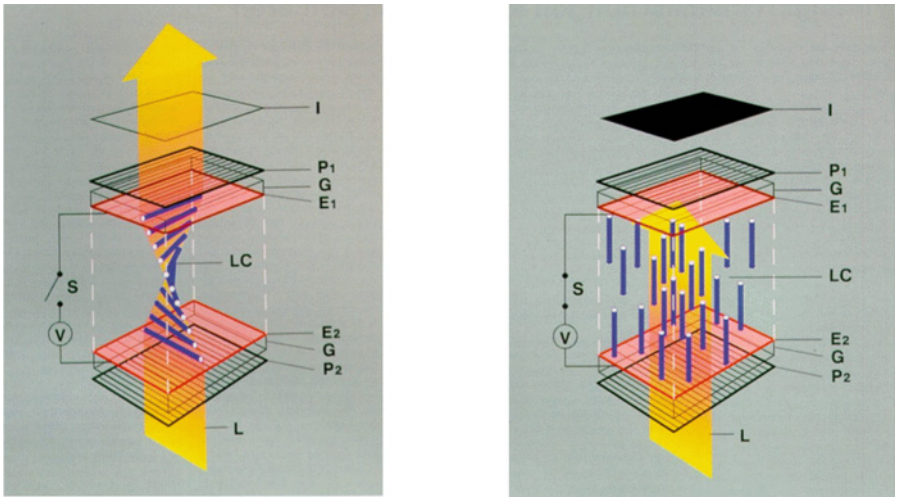

FIG. 14.2. Illustration of construction and operation of a single pixel of a twisted nematic liquid crystal cell. No voltage applied = OFF state (left diagram); voltage applied = ON state (right diagram) (courtesy of M. Schadt).

In colour LCDs, each individual pixel is divided into three cells, or subpixels, which are coloured red, green and blue, respectively, by additional filters. Each subpixel can be controlled independently, so that thousands or millions of possible colours can be obtained for each pixel.

An active matrix LCD is the predominant type of flat panel display. It is the standard display device used as general computer displays, in notebooks and increasingly as high quality displays for medical imaging. The active matrix design permits switching each pixel individually by applying a row and column addressing scheme. It is implemented by thin film transistor technology which supplies each pixel of the display with its own transistor. The circuit is made by depositing a thin film of silicon on the glass surface where the transistors are fabricated. The transistors take up only a small fraction of the surface and the rest of the silicon film is etched away to let the light pass through (Fig. 14.3).

LCD displays using twisted nematic liquid crystals exhibit a strong dependence of display brightness and colour on viewing angle. Developments in technology have considerably improved the angular viewing response. The preferred technique used for medical display devices is currently the in-plane switching (IPS) technology. IPS aligns the crystals horizontally and applies the voltage to realign the liquid crystal structure to both ends of the cell. Non-uniformity of both luminance and luminance ratio of an LCD is expressed as a function of viewing angle (from the normal to the display surface) for horizontal and vertical directions separately.

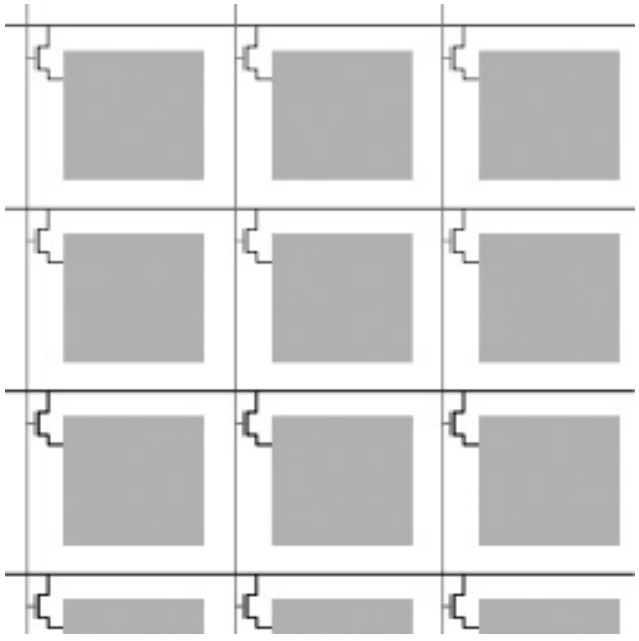

FIG. 14.3. A diagram of the pixel layout. Each liquid crystal pixel is connected to a transistor which provides the voltage that controls the brightness. The pixel is addressed by a row–column scheme.

14.3.4. Hard copy devices

Although reporting is increasingly performed using soft copy displays and PACS facilities, there is still a need for hard copies of images, such as for use in the operating room or to be sent to referring physicians. Hard copy images of diagnostic quality are printed on X ray film and use high resolution laser printing and dry processing. In nuclear medicine, where the original images of a diagnostic procedure often use pseudo-colours for the final display of the results, even cheap mainstream colour printers may adequately be used. The long term storage properties of hard copy media are not an issue when used in connection with a PACS, where images are stored in electronic form. Therefore, technologies such as dry laser film, thermal printers, colour laser printers or inkjet printers are all acceptable hardcopy output devices.

The spatial resolution of a printer is given in dots per inch (dpi), a measure of spatial printing density. It is defined as the number of individual dots that can be placed within the span of 1 in (2.54 cm).

14.3.4.1. Film laser imager

Dry laser imagers print X ray images on transparent film with the same quality as available with conventional X ray film. Spatial resolution is up to 650 dpi, adequate for diagnostic quality output for all imaging procedures, including mammography. Contrast resolution depends on film quality and can reach a D_{max} of up to 4.0.

14.3.4.2. Colour printer

Mainstream colour laser printers produce cheap grey scale and colour output of images with a spatial resolution of typically 600–1200 dpi. For normal paper, the CR is low. Image quality can be improved by using special paper with a smooth surface for improved toner gloss and sharpness.

Inkjet printers can have a spatial resolution of up to 9600 × 2400 dpi. With special photo paper, excellent image quality equivalent to colour photographs can be achieved. The stability of the printout is known to be somewhat fragile, with the image fading within a couple of years even under optimal storage conditions.

14.4. GREY SCALE DISPLAY

Nuclear medicine images do not require the same high quality grey scale displays as is necessary for the display of X ray images. This can be attributed to

the fact that the nuclear medicine image is a low count image with considerable statistical fluctuations, making the comparison of tiny intensity differences meaningless. A main difference to diagnostic X ray reporting is the fact that colour in the image was recognized early as a helpful technique to improve diagnostic reading and a tradition of visualization in colour has been established. Therefore, images and analysis of results, in particular curves and functional or metabolic images, are preferably displayed using colours. The display is usually done on workstations with special nuclear medicine software and using current COTS colour LCD screens as standard display devices. Typical screen sizes are from 20 to 24 in and native display resolutions from 1024 × 1280 pixels to 1200 × 1600 pixels. Depending on the capabilities of the nuclear medicine display workstation's software, several monitors can be used simultaneously.

The need to be able to perform concurrent diagnostic reporting on X ray images, generated by dual mode acquisition techniques such as PET/CT and SPECT/CT, and the inclusion of images from other modalities via PACS in the reporting session, requires the use of grey scale display devices of diagnostic quality (primary devices) at the nuclear medicine display workstation.

Both CRT and LCD display devices are available with spatial resolution and CRs satisfying the requirements for a primary device. LCD displays are rapidly replacing CRT displays for several reasons:

— LCDs typically have about twice the brightness of CRTs. An overall brighter image is less sensitive to changes in the level of ambient light and is preferred for reporting.
— LCD monitors exhibit no geometric distortion.
— LCDs have a weight that is about one third of that of a comparable CRT.
— LCDs are less prone to detrimental ageing effects.
— LCDs are less expensive.

Furthermore, high quality colour LCD devices can be used as grey scale primary devices, which is not feasible for a colour CRT monitor.

14.4.1. Grey scale standard display function

Today's ubiquity of PACSs enables deployment of display devices at all locations where access to medical images is needed. The main challenge when using different display devices in a PACS is to ensure that an image presented to an observer appears identical irrespective of the display device used, be it a CRT based or LCD based soft copy display, or hard copy displays, such as film laser printers or paper printers. The Digital Imaging and Communications in Medicine (DICOM) grey scale standard display function (GSDF) offers a strategy that

ensures that a medical image displayed or printed at any workstation or printing system for which the GSDF is implemented has the same visual appearance, within the capabilities of the particular display device [14.1]. This implies that a display device complying with the GSDF must be standardized and calibrated, and that a scheme of regular quality control is required for the display systems on the PACS. Colour display systems may also be used for the purpose of displaying grey scale images if calibrated to the GSDF [14.3].

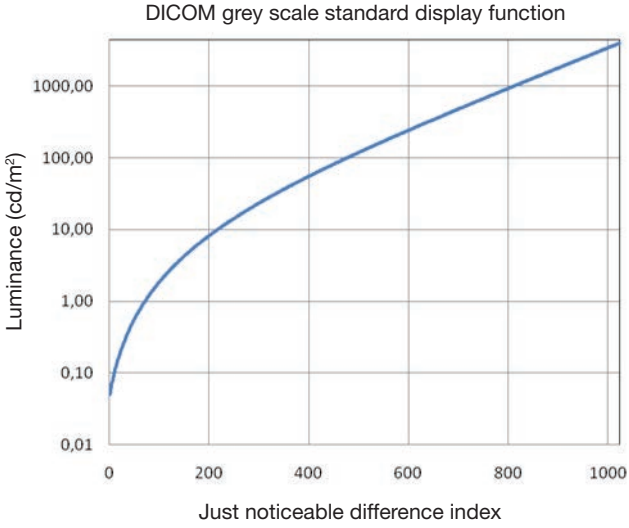

FIG. 14.4. Digital Imaging and Communications in Medicine (DICOM) grey scale standard display function.

The visual appearance of a native image as produced by the acquisition device (gamma camera, PET scanner, CT scanner) depends, if no corrections are applied, on the characteristic curve of the particular display device used at a display workstation. An image displayed using the characteristic curves inherent to a particular device could be significantly different in visual perception from the optimal presentation. Part 14 of the DICOM standard [14.1] standardizes the display of grey scale images. It does so by introducing the GSDF which can be seen as a universal characteristic curve (Fig. 14.4). The GSDF is based on human contrast sensitivity. It covers a luminance range from 0.05 to 4000 cd/m^2. The minimum luminance is the lowest that can be used in practice with a CRT display, whereas the maximum luminance is slightly above the luminance of a very bright unattenuated light box used for the examination of mammography X ray films, so that it covers the range of luminance values of all display

devices in current use. Human contrast sensitivity is non-linear within this range. Perceptual similarity of a displayed image is achieved by linearizing the GSDF with respect to contrast sensitivity. This is done by introducing the JND index. One step in the JND index corresponds to a luminance difference that is just noticeable, regardless of the mean luminance level. The DICOM standard contains the standard GSDF as a tabulation of luminance (brightness) against the JND index. Table 14.1 shows the first and the last few JND indices of the tabulation. It can be seen clearly that the relative changes of luminance need to be much larger on the dark side of the curve than on the bright side to achieve a JND.

TABLE 14.1. TABULATED JUST NOTICEABLE DIFFERENCE INDICES OF THE GREY SCALE STANDARD DISPLAY FUNCTION

Just noticeable difference	Luminance (cd/m^2)
1	0.0500
2	0.0547
3	0.0594
4	0.0643
—	—
—	—
—	—
1021	3941.8580
1022	3967.5470
1023	3993.4040

Note: The relative difference between the luminance of consecutive just noticeable difference indices is much higher for low indices (~9%) than for high indices (~0.6%).

An individual display device with a luminance range L_{min}–L_{max} and a DDL range of, for example, 8 bits exhibits a characteristic luminance curve as a function of the DDL as shown in Fig. 14.5. The device specific characteristic curve will usually not match the corresponding segment of the GSDF. A transformation is needed, which is implemented as an LUT. The LUT will map a DDL D_s which should produce the standard luminance value L_s to the value D_m, so that for input level D_s the transformed value D_m will produce the correct luminance as required by the GSDF. The transformation LUT correcting for the

deviations of the characteristic curve of a specific display system from the GSDF may be implemented directly in the display device or in the video memory of the display controller. The result of the transformation is that the modified DDLs operating the display will generate a characteristic curve identical to the GSDF.

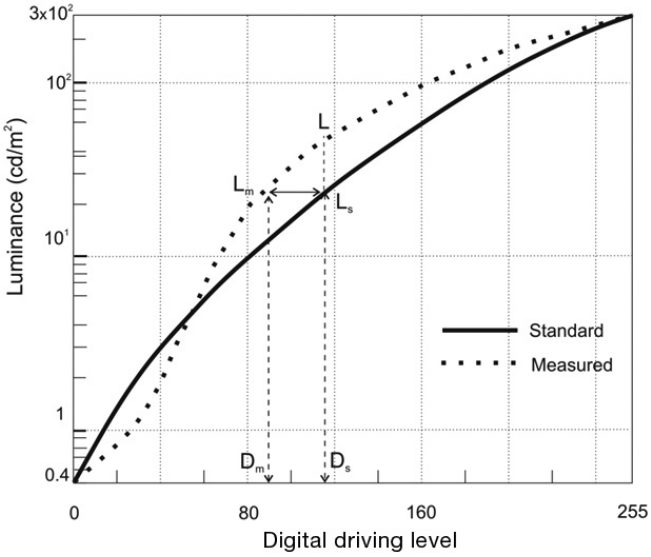

FIG. 14.5. Mapping of digital driving level D_s to the value D_m, so that for input level D_s which should produce the standard luminance value L_s the transformed value D_m will produce the correct luminance as given by the grey scale standard display function.

14.5. COLOUR DISPLAY

The human eye can distinguish millions of different colours. The full range of colours the average human can see is given by the spectrum of sunlight. Each colour can be characterized by three coordinates representing the colour as a mixture of three primary colours in a colour space.

One of the first colour spaces introduced in 1931 is the International Commission on Illumination (CIE) xyz colour space [14.5]. It is derived from a model of human colour perception and uses three tri-stimulus values to compose a colour. It is designed in such a way that one of the three coordinates defines luminance (brightness); the other two coordinates represent the chromaticity (hue). This leads to the well known CIE 1931 chromaticity diagram with its typical horseshoe shape in which all colours the human visual system can perceive are represented as a function of two coordinates, x and y (Fig. 14.6). The third

coordinate, representing brightness, would only change the saturation of a colour, so, for example, varying the brightness coordinate for the chromaticity 'white' would run through all levels of grey from black to the maximum white a display device can render.

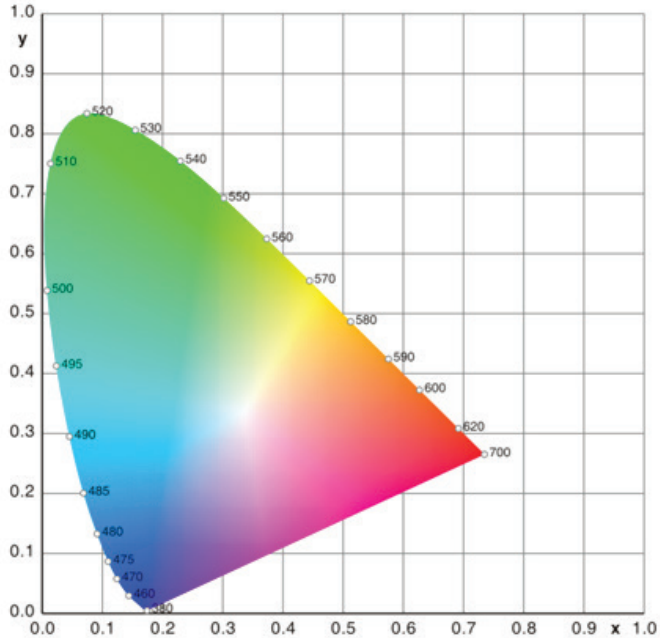

FIG. 14.6. International Commission on Illumination xy chromaticity diagram. The outer curved boundary is the spectral locus, with wavelengths shown in nanometres.

Another frequently used colour space is the red, green, blue (RGB) space, a natural colour space for a CRT or LCD colour monitor. It uses as coordinates the intensities of the red, green and blue primary colours to generate a colour pixel (Fig. 14.7).

The colour space used for hard copy printers is the cyan, magenta, yellow, key (black) (CMYK) space.

The quality of the colour image depends on the colour depth (the range of colour intensities) with which each subpixel contributes. Colour quality increases with subpixel depth. A common classification of the display controller's ability to reproduce colours is: 8 bit colour (can display 256 colours), 15/16 bit colour (high colour: can display 65 536 colours), 24 bit colour (true colour: can display 16 777 216 colours) and 30/36/48 bit colour (deep colour: can typically display over a billion colours).

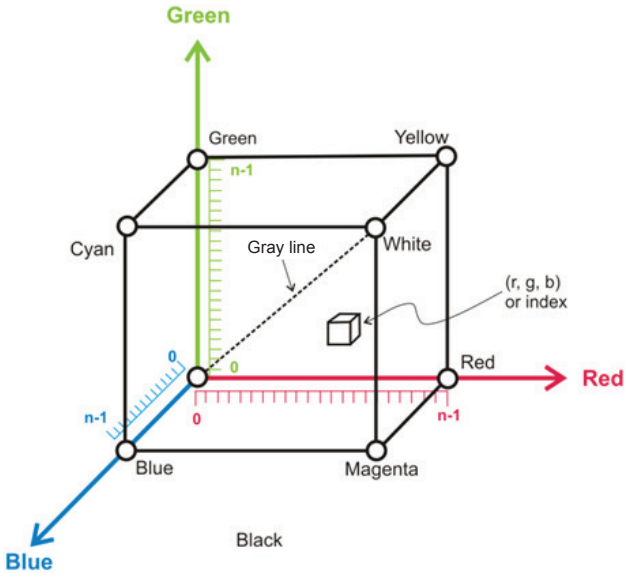

FIG. 14.7. Red, green, blue colour cube with a grey line as diagonal. The number of (r, g, b) voxels, i.e. colours available, depends on the bit depth of each coordinate. A depth of 8 bits for each component would result in 16 777 216 colours.

A nuclear medicine display controller can usually handle true colour pixel depths, with 8 bits available for each primary colour.

Colour was utilized in nuclear medicine already at an early stage of the development of digital displays. Since the original image data contain no colour information, the allocation of colour to an image pixel can be freely chosen. The allocation takes the form of a CLUT. Conceptually, the CLUT is an array structure containing the colour coordinates for each colour included in the table. A colour is defined by three values representing the intensities of the red, green and blue subpixels. Each pixel intensity in the image maps to an array index of the LUT, so that each intensity is associated with a particular colour. This is accomplished by a transformation algorithm. The transformation is usually carried out by the GPU of the display controller. The CLUT is stored in the memory of the graphics card. The LUT is usually much smaller in size than the image. Usual CLUTs contain 64–256 elements, respectively colours. An advantage of a CLUT is that colours can be changed by changing the LUT, resulting in better display performance. It is worthwhile noting that for a real world colour image, the colour of each pixel is determined by the image itself and cannot be arbitrarily associated with a colour such as is the case for pseudo-colour display. Thus, the quality of a real world image increases the larger the number of colours that can be reproduced. Using a CLUT for a colour image of the real world implies a loss of quality, as

can easily be seen on images on the Internet which use CLUTs with typically 64 colours to save on image size. The addition of colour information to native nuclear medicine and X ray images always results in a pseudo-colour image, with the colours chosen by the user.

A modern nuclear medicine system typically uses 16–32 different CLUTs. The choice of colours is a complex issue. A continuous colour scale can be achieved if the individual components vary slowly and continuously. Pseudo-colour can be used to increase the PDR relative to grey scale; other CLUTs may emphasize regions with a specific intensity as, for example, in the case when performing a Fourier analysis of the beating heart to highlight amplitude and phase information.

14.5.1. Colour and colour gamut

As with grey scale images, it is expected that a colour image displayed on a PACS display device has the same colour appearance regardless of the type or the individual characteristics of the display device. Fortunately, the problem of producing digital colour images with the same perception of colours regardless of the display device, including display monitors and hard copy printers, has already been resolved by the printing industry and the photographic industry.

Since each colour is a unique entity, it is to be expected that unambiguous transformations exist between the coordinates representing the colour in different colour spaces. Such transformations are indeed available and are the basis of a CMS. The purpose of a CMS is to produce a colour image that is perceived as being the same by a human observer regardless of the output device used.

The gamut or colour gamut is defined as the entire range of colours a particular display device can reproduce. The gamut depends on the type of display and on design characteristics. The number of vertices of the gamut is given by the number of primary colours used to compose a colour. In the case of an LCD or a CTR monitor, the three primary colours, red, green and blue, are used to produce a colour. For a printer, the colours of several inks or dyes can be mixed to produce a colour on paper. Most printers can create dots in a total of six colours which are cyan, yellow, magenta, red (which combines yellow and magenta), green (yellow plus cyan) and blue (cyan plus magenta). Typical gamuts for an LCD monitor and for a printer are shown in Fig. 14.8. It is obvious that the monitor can display colours unavailable to the printer and vice versa. The International Color Consortium (ICC) has published procedures including colour transformations (CMS) that ensure that a colour image that is displayed on, for example, a monitor has the same appearance on, for example, a colour printout [14.6]. The system is based on describing the colour properties of a colour display device by an ICC colour profile. The colour profile contains

manufacturer provided or, preferably, the measured gamut of the individual display device in a format which permits transformation of the colours between the representation on the display device and an intermediary colour space, for which the CIE xyz space or the CIE lab space are used. The intermediary colour space acts as a colour reference and is called the profile connection space (PCS). The PCS is utilized by DICOM [14.3], analogous to the GSDF for grey scale displays, as a reference space for the transformation of colours from one colour display device to another. Unlike the GSDF, it does not, however, claim to linearize perceived contrast. The colours of an individual display device can be transformed with the help of the ICC profiles to any other display device, including hard copy colour printers while maintaining the same visual perception of the colours. For colours available on one device but not on the other, the PCS substitutes colours similar in perception to the missing ones. In order to make the PCS work, it is necessary that all display systems involved have their ICC profiles available. The DICOM standard formalizes the information required for colour management by adding the necessary tags to the data dictionary, so that in a medical PACS the colour transformations are carried out by the CMS in a manner transparent to the user.

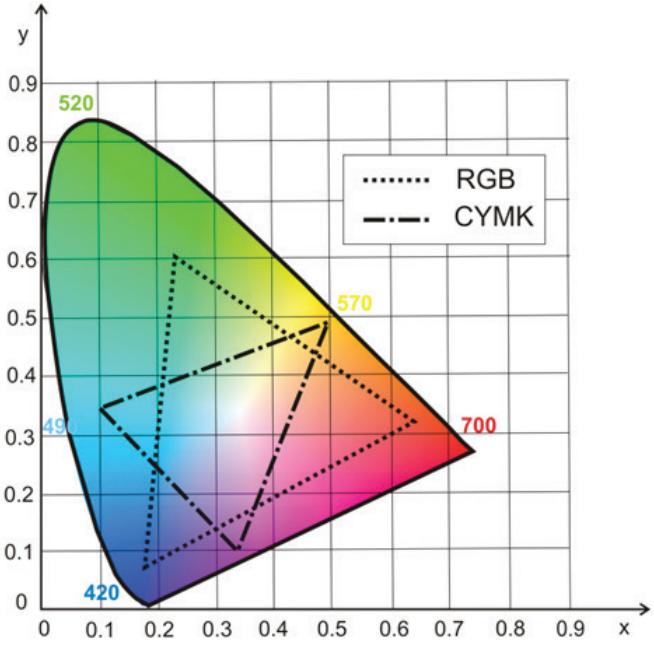

FIG. 14.8. Typical gamuts for a liquid crystal display monitor and a colour printer. The large non-overlapping areas of the colours which cannot be reproduced by the other device and must be substituted by similar colours should be noted.

14.6. IMAGE DISPLAY MANIPULATION

14.6.1. Histograms

The intensity histogram of an image represents the distribution of the grey values in an image. It is obtained by dividing the range of grey values into intervals of equal width, the bins, and calculating the number of pixels with intensity values falling into each bin. The number of bins to store the frequencies can be freely chosen, but the most informative displays are obtained with bin numbers between 32 and 256. The graphical display of the histogram transmits a rough idea about the distribution of intensities (Fig. 14.9).

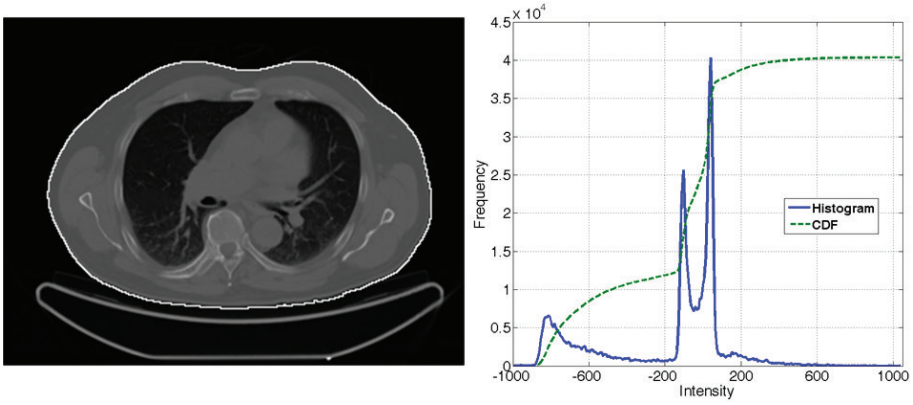

FIG. 14.9. Transverse CT slice at the height of the heart (left), with the corresponding intensity histogram, using only the pixels within the region surrounding the trunk. The number of bins is 256. Even when excluding all background pixels, the unequal distribution of intensities is seen, especially the lack of high intensity values which is in agreement with the small proportion of bony structure in the image.

14.6.2. Windowing and thresholding

The most basic intensity transformations used in image display are to transform a pixel intensity I to a grey scale intensity value r within the available grey scale range Q of the display monitor:

$$r = T(I) \tag{14.1}$$

Q is normally in the range 0...255. The transformations do not take into account the values of surrounding pixels; each pixel is processed independently.

Windowing and thresholding are linear intensity transformations of that type. They provide an easy method to emphasize contrast and visibility of structures in areas of interest by only mapping intensity values within an intensity window defined by a threshold and a window width to the available range of brightness values. Values below the threshold are set to black; values above the upper level are set to maximum display intensity. Thus, for an intensity threshold level t and a window width w, the pixel intensity I is transformed into the grey scale value r according to:

$$r = \begin{cases} \beta I, t \le I \le t + w \\ 0, I < t \\ Q, I > t + w \end{cases} \tag{14.2}$$

with $\beta = Q/W$.

Windowing and thresholding may be hardware implemented, i.e. the values may be changed by turning knobs on the monitor or the console or, more frequently, by software implementation using mouse movements, sliders or the arrows on the keyboard of the display workstation. The diagnostic value offered

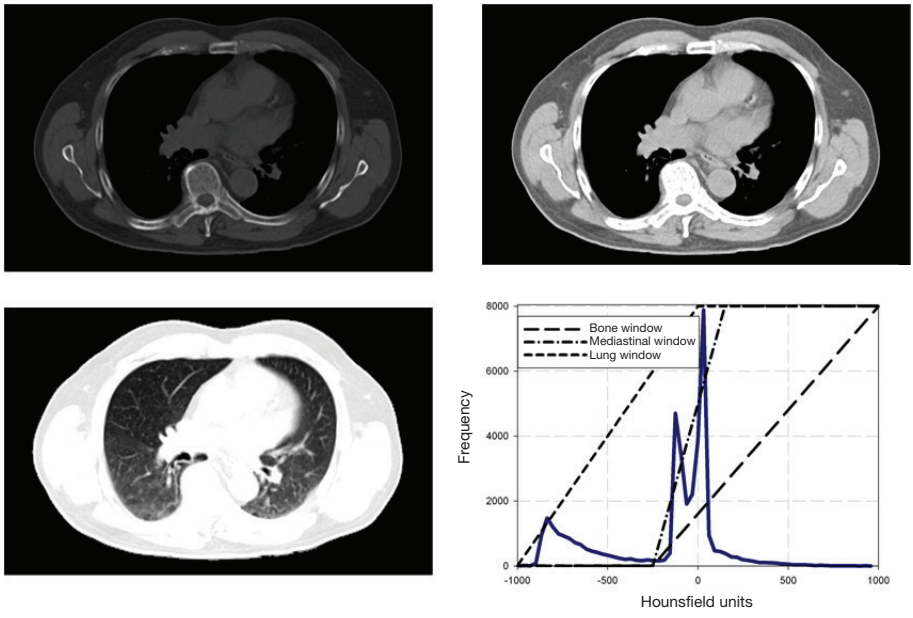

FIG. 14.10. CT slice from Fig. 14.9 with typical lung and mediastinal windowing (upper row from left to right), a bone window (bottom left) and a histogram with corresponding linear window functions (bottom right).

by windowing and thresholding can be appreciated when using typical CT windows for bone, mediastinum and lung. This is shown for a transaxial CT slice through the chest, together with the range of intensity values actually visualized out of the total histogram (Fig. 14.10).

Simple algorithms may be successfully used to perform automatic windowing and thresholding using histogram data, such as minimum and maximum intensity or setting the threshold and window width in such a way that a small percentage of the lowermost and uppermost intensity values are discarded. Suitable values are between 0.5 and 2%.

14.6.3. Histogram equalization

Image intensity values may utilize the range of display intensities inefficiently. The CT slice of Fig. 14.9 is a typical example of a medical image and demonstrates that most of the intensity values are the Hounsfield units for soft tissue and the lung.

Histogram equalization aims at utilizing each grey scale level available for display with the same frequency. If all intensity values were present in equal numbers in the image, the histogram would be flat, and the corresponding cumulative density function would increase linearly. A redistribution of intensity values s to approximately equally distributed grey scale intensity values r can be achieved using the transformation:

$$r_{eq} = \mathrm{CDF}(I) \times (Q - 1)/(M \times N) \tag{14.3}$$

where

CDF(I) is the cumulative density function of the original image;
Q is the available range of grey scale values;

and the image size is $M \times N$ pixels.

For more details, see Ref. [14.7]. Figure 14.11 demonstrates the effect of histogram equalization using the standard algorithm of the image processing software package ImageJ [14.8] on the CT slice of Fig. 14.9. In the processed image, the structures of both the bronchi of the lung and the ribs are visualized in the same image without underflow or overflow and with approximately the same information content as in the three windowed images of Fig. 14.10 together. The drawbacks of the method are that the visual appearance of an image depends on the shape of the histogram and may, therefore, be significantly different between patients, and the fact that the resulting intensity data can no longer be

used to extract quantitative information. The latter is nicely shown by the range of intensity values in the histogram of Fig. 14.11, which no longer shows the familiar range of CT numbers.

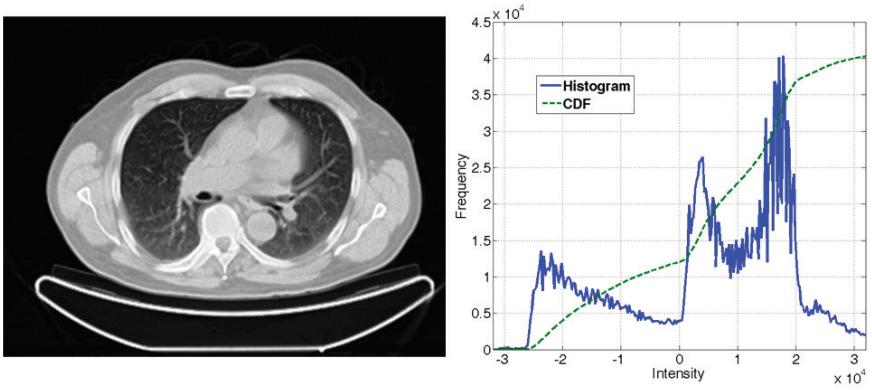

FIG. 14.11. CT slice from Fig. 14.9 after histogram equalization with a corresponding intensity histogram, using only the pixels within the region surrounding the trunk. The number of bins is 256. The cumulative density function is now approximately linear. The intensity values are no longer related to Hounsfield units. Owing to the better distribution of the intensity values, all of the structures of interest, including the bone and the bronchi of the lung, are visualized simultaneously.

14.7. VISUALIZATION OF VOLUME DATA

14.7.1. Slice mode

An image volume dataset consists of a series of adjacent image slices through the patient's body. The slices can be displayed sequentially with manual stepping through the images or automatically as a movie, or they can be displayed simultaneously as a montage of several images. Specialized viewing software offers easy ways to manipulate the display further and permits, for example, zooming and panning. Panning consists of quickly moving around a zoomed image too large to be displayed completely on the screen by utilizing the mouse, a joystick or a scroll wheel.

From the original slices, orthogonal views can easily be calculated by rearranging the pixel matrix. Presenting the orthogonal views simultaneously on the screen facilitates the anatomical location of structures. Slices with oblique orientations can also be calculated. In myocardial SPECT and PET, reorientation along the long and the short axes of the left ventricle are the standard display for

visualization (Fig. 14.12). A substantial gain in anatomical information can be achieved by using 'linked' cursors. The technique consists of moving a cursor in one image while a second cursor is simultaneously moved by software to identical locations in the other views.

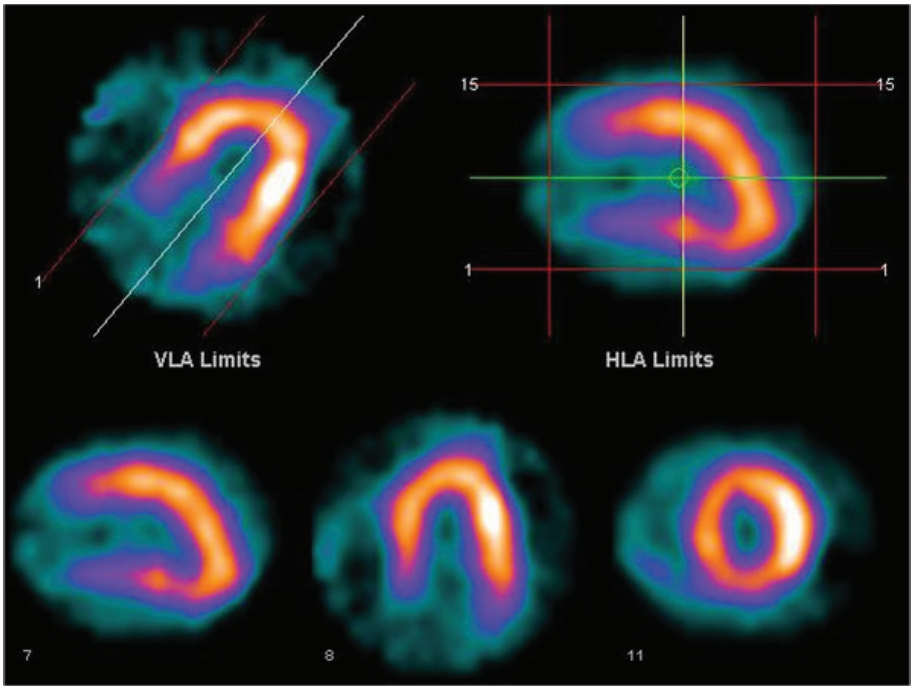

FIG. 14.12. Orthogonal views of myocardial perfusion SPECT with orientation of the slices along the long axis of the heart. The upper row shows an original transaxial slice through the myocardium with the white line indicating the long axis (left) and a sagittal slice (right). The bottom row shows the reoriented views with the vertical and horizontal slices through the long axis, and a slice perpendicular to the long axis (from left to right). (Courtesy of B. König, Hanuschkrankenhaus, Vienna.)

14.7.2. Volume mode

Volume mode display refers to techniques which extract the information about structures in the 3-D image dataset by selecting intensity information directly from the volume dataset and projecting the selected values on the display screen. The ray casting technique projects a line from a viewpoint through the data starting from a pixel on the display screen. It calculates the value of interest using the image intensities along its path (Fig. 14.13). Less frequently used in nuclear medicine are splatting techniques consisting of the projection

of possibly modified image voxels on the display screen; these techniques will not be considered further here. Details can be found, for example, in Ref. [14.9]. The dominant ray casting geometry in nuclear medicine applications and in dual mode imaging is parallel projection. Perspective projection is predominantly used in virtual endoscopy and is not yet used routinely in dual mode imaging.

14.7.2.1. Transmission type volume rendering

Maximum intensity projection (MIP) consists of projecting the maximum intensity value encountered along the trajectory of the ray through the data volume on the corresponding screen pixel. It improves the visualization of small isolated hot areas by enhancing the contrast (Fig. 14.14). MIP is successfully employed for lesion detection in PET oncological whole body studies. Its efficiency for the detection of lesions is further increased by displaying the MIP projections as a sequence of projection angles in cine mode.

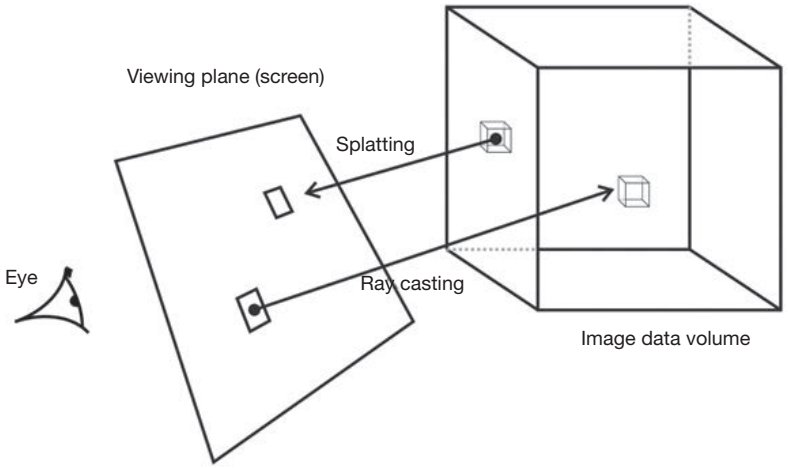

FIG. 14.13. Principle of ray casting and splatting geometry. In ray casting, the ray collects intensity transformation throughout its trajectory. A voxel is usually hit outside its centre which has to be corrected for by interpolation. Splatting starts from the centre of a voxel and distributes its intensity on several screen pixels.

Summed voxel projection produces the rendered image by summing all intensities along a ray trajectory. If applied to a CT volume, it is known as a digitally rendered radiograph. If central projection geometry is used, the projection image simulates a conventional X ray image. Digitally rendered radiographs of CT data are used in radiotherapy for the positioning and registration of patients. Tomographic datasets from nuclear medicine displayed as digitally rendered

radiographs may be used to compare lesion extensions with planar X ray images of the patient.

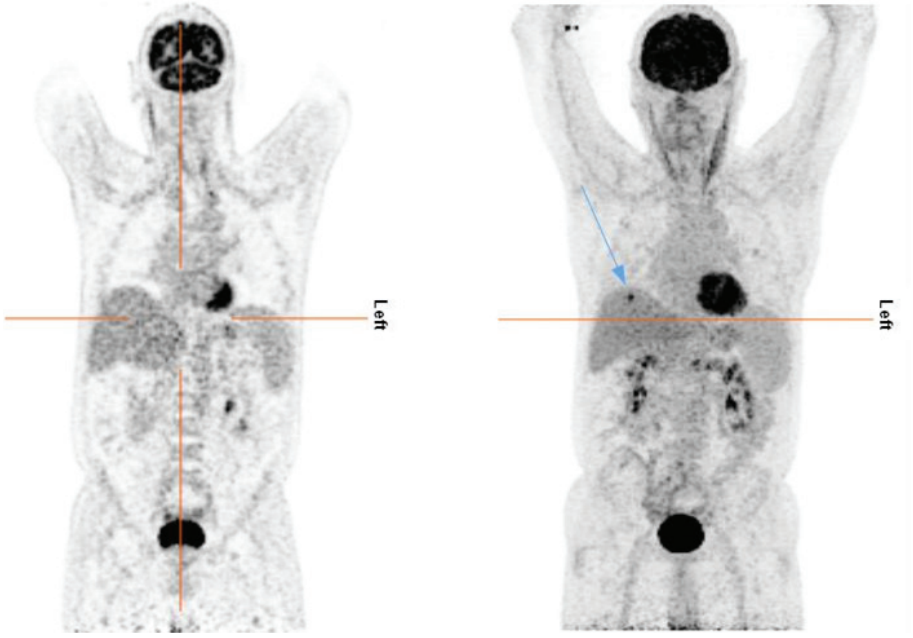

FIG. 14.14. A maximum intensity projection (right) compared to a standard coronal slice. A solitary lesion is clearly visible in the maximum intensity projection image (arrow) while it is missing in the coronal standard view.

14.7.2.2. Reflection type volume and surface rendering

The purpose of surface and volume rendering is to visualize structures in volume datasets as objects and display them in a pseudo-3-D mode. Volume rendering techniques extract information about objects directly from the 3-D volume data. They start by casting rays through the image volume data and by processing the intensities of the voxels along the ray trajectory. Depending on the handling of the intensities, different types of 3-D display can be generated.

Three dimensional rendering utilizes standard computer graphics techniques, such as illumination, shading and the application of textures, to produce a realistic appearance of anatomical structures and tumours. This is useful for the visualization of complex anatomical relationships, can improve the orientation of surgeons and resolve ambiguities of localization.

For registered images originating from different imaging techniques, such as images from magnetic resonance and from PET, anatomical and functional

data can be displayed simultaneously, thereby taking advantage of the excellent morphological resolution of one modality and of functional, blood flow or metabolic information of the other modality. Such combined images are capable of displaying the spatial relationships between different objects, such as, for example, the surface of the left ventricle together with the location of the coronary arteries, or the surface of the brain grey matter rendered from a magnetic resonance study combined with the blood flow obtained by an HMPAO (hexamethylpropyleneamine oxime) SPECT study.

Surface rendering traditionally starts from a sequence of contours extracted from the object of interest. The surface is obtained by fitting a mosaic of triangles followed by illumination and shading. The relatively small number of parameters needed to describe an object permits real time visualization of transformations, useful, for example, for interactive surgical planning. Analytical descriptions of the objects can also be generated, which can be used by other programs, such as CAD/CAM, or which can be used to produce physical models of the objects of interest using lithographic techniques.

Three dimensional surfaces are generated by specifying an intensity threshold. The method is closely related to the generation of contours. When a ray encounters the threshold value along its trajectory, the location of that voxel is interpreted as a surface point of the structure of interest. The appearance of a 3-D image is improved further by utilizing illumination and shading techniques. In order to apply these effects, additional knowledge about the orientation of the surface element is required for which gradient techniques with various levels of sophistication are employed.

Voxel gradient shading is the most successful technique to produce illuminated and shaded surfaces. It calculates a gradient vector from a voxel neighbourhood and renders a realistic pseudo-3-D image by calculating diffuse reflective illumination from an external light source and applying shading. Noise in the surface is reduced by smoothing (Fig. 14.15, middle).

Volume compositing can be considered a generalization of voxel gradient shading. It aims at visualizing internal structures beyond the limit given by a threshold by using information from all voxels along a ray. The method consists of calculating a gradient for each voxel along the ray, thereby assigning a surface to each voxel, and applying lighting and shading to that surface. The light transmitted and reflected by each voxel is then collected into the pixel value on-screen by assigning opacities to each voxel and summing the results along the ray (Fig. 14.15, right). Volume compositing is by far the most complex and time consuming rendering method. The results are similar to the voxel gradient method. Under favourable conditions and through careful selection of the rendering parameters, volume compositing can visualize several objects simultaneously in the rendered image.

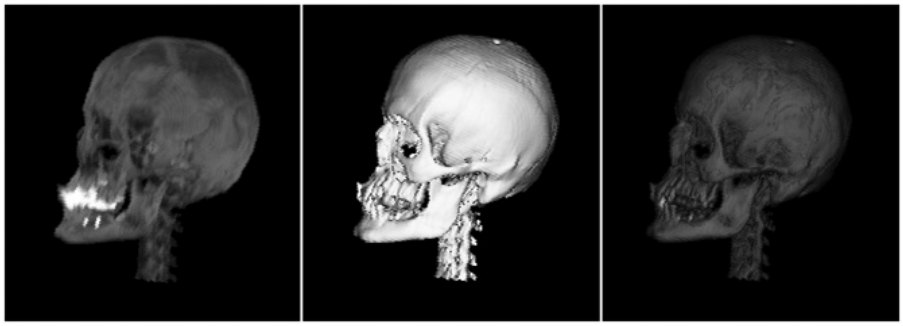

FIG. 14.15. Surface of a skull from CT image data using (from left to right) maximum intensity projection, voxel gradient shading and volume compositing for rendering. Rendered images were produced with ANALYZE© 9.0.

14.7.3. Polar plots of myocardial perfusion imaging

Myocardial perfusion imaging is a tomographic technique using a myocardial perfusion tracer such as the potassium analogue ^{201}Tl or ^{99m}Tc-MIBI (methoxyisobutylisonitrile) to produce SPECT images of the perfusion of the left ventricle. Myocardial perfusion is reduced or absent in ischaemic and infarcted areas. The size and severity of perfusion defects is of high diagnostic and prognostic value. Therefore, myocardial perfusion imaging is among the most frequent nuclear medicine investigations. Unfortunately, visual interpretation of the tomographic slices is difficult and suffers from high inter-observer variability due to the poor spatial resolution of SPECT studies in general and to the added blurring of the images by the motion of the heart during acquisition. Display methods tailored to a more reliable detection of perfusion defects were, therefore, developed shortly after the introduction of myocardial perfusion SPECT.

The initial visual presentation of the tomographic images makes use of a coordinate system natural to the anatomy of the left ventricle. One coordinate axis passes through the long axis of the heart; the other two are perpendicular to the long axis and to each other (Fig. 14.16). The standard display consists of slices perpendicular to the long axis, the short axis slices and two sets of slices parallel to the long axis. The left ventricle has an annular shape in the short axis slices. The annuli can be aggregated into one image using a polar map representation [14.10]. In a first step, each annulus is reduced to a circumferential profile. The methods of choosing the circumferential profile vary, for example, using the maximum intensity at each angular step only or taking a mean intensity, and the thickness of the annulus can also be considered at each angular step. Thereafter, all of the profiles are arranged into one image, starting with the

profile representing the apex at the innermost position, with each following profile surrounding the previous one. The resulting display is referred to as a bull's-eye display or polar map (Fig. 14.17). The latter name refers to the fact that the intensity along a given annulus can easily be handled by a polar coordinate system. The intensities displayed for each annulus correspond to the myocardial perfusion in that slice or a segment thereof. Absolute perfusion values cannot be derived from the intensities. The method to obtain an estimate of the degree of hypo-perfusion and of the location of the perfusion defect consists of comparing the relative intensity values in different segments of the annuli to the maximally perfused segments of the same patient, and then comparing the pattern of relative perfusion of the individual study with normal perfusion patterns. This permits an estimate of both the degree and extent of the perfusion defects as well as a good anatomical allocation to the coronary arteries causing the hypo-perfusion.

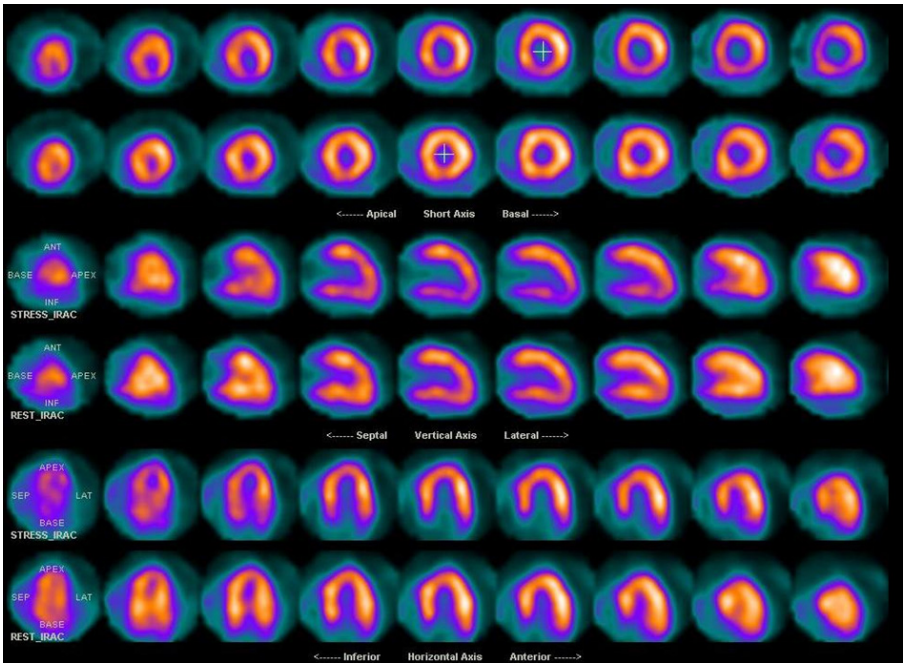

FIG. 14.16. Results of a stress–rest perfusion study with ^{99m}Tc-MIBI (methoxyisobutylisonitrile) with the orientation of slices adapted to the long axis of the heart. Images show hypo-perfusion of the inferior wall. The upper rows are stress images; the lower rows are images at rest. (Courtesy of B. König, Hanuschkrankenhaus, Vienna.)

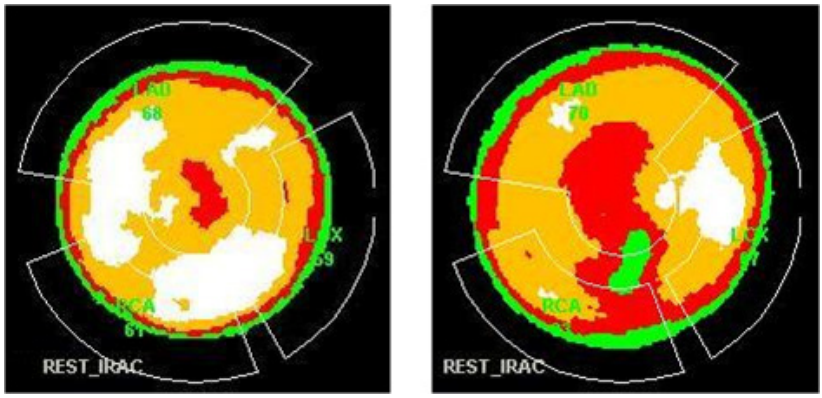

FIG. 14.17. Bull's-eye displays of myocardial SPECT perfusion studies. Normal perfusion (left) and hypo-perfusion of inferior wall (right). The colours indicate the degree of perfusion, with white — normal, orange — acceptable, red — hypo-perfused, and green — no perfusion. Also indicated are the perfusion areas for the main coronary vessels LAD, LCX and RCA. (Courtesy of B. König, Hanuschkrankenhaus, Vienna.)

14.8. DUAL MODALITY DISPLAY

Several techniques have been developed to display registered images originating from different modalities, such as from a PET/CT study. The simplest technique is to display the images belonging together side by side. Anatomical information can be gained easily by using the linked cursor technique. The CT image which has superior spatial resolution is, thus, used to determine the anatomical location of a lesion visible in the PET image. The linked cursor technique, while providing precise anatomical information, is impractical if several lesions are present in the image as often is the case in oncological studies. In such situations, alpha blending is helpful, which combines both the CT and the PET image into one composite image.

Alpha blending consists of adding the images pixelwise with different weight. The weight is called the transparency factor α, with $0 \leq \alpha \leq 1$. The composite pixel I_{CS} is given by:

$$I_{CS}(m, n) = \alpha \times I_{BG}(m, n) + (1 - \alpha) \times I_{FG}(m, n) \quad (14.4)$$

where

I_{BG} is the intensity of the background pixel;

and I_{FG} is the intensity of the foreground pixel.

When using the native grey scale images for both modalities, it is difficult to distinguish clearly which intensity comes from which modality. The composite display becomes much easier to interpret if one of the images uses a CLUT. In this case, the formula has to be applied to each colour component separately:

$$R_{CS}(m, n) = \alpha \times R_{BG}(m, n) + (1 - \alpha) \times I_{FG}(m, n) \quad (14.5)$$

$$G_{CS}(m, n) = \alpha \times G_{BG}(m, n) + (1 - \alpha) \times I_{FG}(m, n) \quad (14.6)$$

$$B_{CS}(m, n) = \alpha \times B_{BG}(m, n) + (1 - \alpha) \times I_{FG}(m, n) \quad (14.7)$$

where

R, G and B are the colour components of the background image;

and I is the grey value of the foreground image.

In PET/CT alpha blending, the background image is usually the PET image whereas the CT image as the foreground image retains the grey scale (Fig. 14.18). The composite display can be further improved by changing thresholds and windows for each modality separately and interactively. For CT, the traditional windows are usually employed.

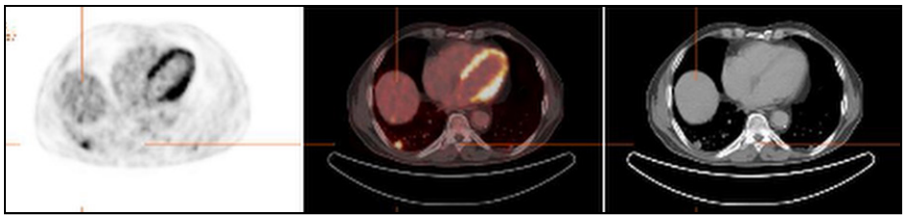

FIG. 14.18. PET/CT fused image display with a PET image on the left showing a hot lesion at the border between the lung and rib cage. The fused image in the middle shows the location inside the lung close to the pleura; the CT image on the right confirms and permits close inspection of the position. Linked cursors point to the position of the lesion. The transparency factor α is 0.5.

14.9. DISPLAY MONITOR QUALITY ASSURANCE

Several standards define the performance parameters of display systems (see, for example, Refs [14.11, 14.12]). The Report of task group 18 of the American Association of Physicists in Medicine offers an exhaustive up to date

set of performance parameters for current medical display monitors, together with procedures and accompanying test images to assess performance [14.13]. It contains limits of acceptance for all parameters, distinguishing between primary devices and secondary devices.

In addition to quality assurance aimed at the individual display device, a major component of quality assurance is to ensure a consistent display of the image at all display workstations of a PACS, including different ambient light conditions. This is resolved by including the calibration and validation of the DICOM GSDF into the quality assurance framework.

Quality control at regular intervals of a display device is required because the performance may change over time, due to ageing processes of the display device, both for CRT and LCD displays, and due to changes in environmental lighting with time.

14.9.1. Acceptance testing

The purpose of acceptance testing is to ensure that the performance of purchased equipment complies with the specifications established in the sales contract. The user should clearly specify in the contract the required performance, the test procedures to assess the performance parameters and the limits of acceptability. Reference [14.13] lists a set of performance parameters which completely characterize the performance of a soft copy display device. These are summarized in Table 14.2.

For each of the parameters, several tests at various levels of sophistication are described in detail. Most of the parameters can be assessed visually by analysing the test images listed in Table 14.2, possibly with additional use of templates on transparency sheets, such as for the assessment of distortions. An exhaustive set of test images has been made electronically available for these tests, both in Joint Photographic Experts Group (JPEG) and DICOM format [14.3]. For quantitative tests, such as for the calibration of luminance characteristic curves, of the DICOM GSDF and of chromaticity values, luminance meters and colorimeters with computer readout of the measured values and special software are necessary.

14.9.2. Routine quality control

In order to make sure that a display system meets the expected performance during its economic lifetime, assessment of performance parameters at regular intervals is necessary. Reference [14.13] recommends that a subset of the tests in Table 14.2, namely geometric distortions, reflection, luminance response, luminance dependencies, resolution and chromaticity, be performed at monthly to quarterly intervals, depending on the monitor's stability. Tests for

TABLE 14.2. PERFORMANCE PARAMETERS OF DISPLAY MONITORS AND EQUIPMENT FOR MEASUREMENT
(according to Ref. [14.13])

Test	Major required tools	
	Equipment	Patterns
Luminance response	Luminance and illuminance meters	TG18-LN TG18-CT TG18-MP
Luminance dependencies	Luminance meter	TG18-UNL TG18-LN TG18-CT
Reflection	Luminance and illuminance meters	TG18-AD
Resolution	Luminance meter, magnifier	TG18-QC TG18-CX TG18-PX
Geometric distortions	Flexible ruler or transparent template	TG18-QC
Noise	None	TG18-AFC
Veiling glare	Baffled funnel, telescopic photometer	TG18-GV TG18-GVN TG18-GQs
Chromaticity	Colorimeter	TG18-UNL80

geometric distortions and for resolution are more important for CRTs, whereas the dependence of luminance on the viewing angle is important only for LCD displays.

In addition, Ref. [14.13] recommends that a daily check prior to clinical work be performed by the user. It consists of evaluating anatomical test images or a suitable geometrical test image such as a TG18-QC test image (Fig. 14.19) to verify adequate display performance. The instructions for assessing the quality of a display device when using the TG18-QC test pattern are given in Box 14.1.

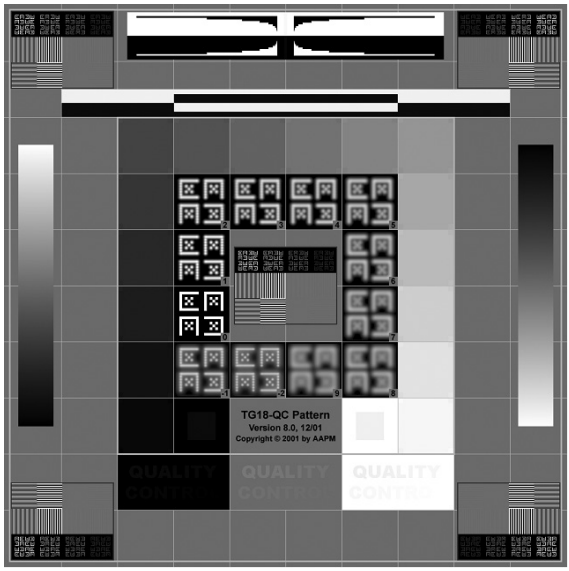

FIG. 14.19. Test pattern TG18-QC suitable for daily quality control of display monitor performance using visual inspection [14.13].

Box 14.1. Instructions for visual assessment of image quality using the TG18-QC test pattern as part of daily quality control by the user [14.23]

1. General image quality and artefacts: Evaluate the overall appearance of the pattern. Note any non-uniformities or artefacts, especially at black-to-white and white-to-black transitions. Verify that the ramp bars appear continuous without any contour lines.
2. Geometric distortion: Verify that the borders and lines of the pattern are visible and straight and that the pattern appears to be centered in the active area of the display device. If desired, measure any distortions (see section 4.1.3.2).
3. Luminance, reflection, noise, and glare: Verify that all 16 luminance patches are distinctly visible. Measure their luminance using a luminance meter, if desired, and evaluate the results in comparison to GSDF (section 4.3.3.2). Verify that the 5% and 95% patches are visible. Evaluate the appearance of low-contrast letters and the targets at the corners of all luminance patches with and without ambient lighting.
4. Resolution: Evaluate the Cx patterns at the center and corners of the pattern and grade them compared to the reference score (see section 4.5.3.1). Also verify the visibility of the line-pair patterns at the Nyquist frequency at the centre and corners of the pattern, and if desired, measure the luminance difference between the vertical and horizontal high-modulation patterns (see section 4.5.3.1).

The most frequent influence on reporting comes from changes in ambient lighting. An increase in the level of ambient light results in poorer discrimination of structures in the darker parts of the image. This has to be compensated for by adapting the GSDF to the current light level. Modern medical display monitors, therefore, provide automatic measurement and recalibration features. A typical, high performance LCD display used as a primary device includes a luminance meter covering a small area of the monitor to continuously control the display characteristic curve and the maximum brightness level. A second photometer records the average ambient lighting. With such an arrangement, it is possible to adjust the GSDF continuously to the DICOM required luminance values, taking into account the changes in L'_{min} and L'_{max}, the minimum and maximum luminance values including the luminance L_{amb} added by ambient light.

As an annual quality control of a display device, the TG18 working group recommends performing all tests carried out during acceptance [14.13].

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