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No. 19

Quality Assurance Programme for Computed Tomography: Diagnostic and Therapy Applications

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QUALITY ASSURANCE PROGRAMME FOR COMPUTED TOMOGRAPHY: DIAGNOSTIC AND THERAPY APPLICATIONS

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QUALITY ASSURANCE PROGRAMME FOR COMPUTED TOMOGRAPHY: DIAGNOSTIC AND THERAPY APPLICATIONS

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FOREWORD

The application of radiation in the diagnosis and treatment of disease is an important component of the work of the IAEA. In the area of diagnostic radiology, this work is currently focused on quality assurance (QA) methods to promote the effective use of radiation for a diagnostic outcome through achieving and maintaining appropriate image quality, and on dose determination to allow the monitoring and reduction of dose to the patient.

The role of computed tomography (CT) in modern medicine is well established as a means of diagnosis and also as an essential precursor to radiation therapy treatment. The clinical relevance of the technology and the recent rapid technological developments have brought about extensive increases in the use of this diagnostic tool generally, and in an increasing number of Member States. The complexity of this technology continues to increase, as does its potential to deliver substantial doses to patients. Consequently, the need for QA to acquire the maximum clinical information at acceptable radiation dose levels is critical.

The current publication is unique in that it contains advice applicable to both diagnostic and therapeutic applications of CT, in recognition of the fact that in many facilities the use of a CT scanner for both diagnostic and therapeutic applications may be common. Currently, there are a small number of QA protocols in CT that apply to limited national and regional settings. Many Member States, therefore, have requested guidance in this area. In responding to these requests, the current publication was written with the aim of presenting an internationally harmonized approach to QA in the field. This approach will allow Member States to implement QA in CT in a standardized way.

This publication on QA for CT was developed as a companion to complement the recently published Quality Assurance Programme for Screen Film Mammography (IAEA Human Health Series No. 2) and Quality Assurance Programme for Digital Mammography (IAEA Human Health Series No. 17) and follows the same format and style.

The IAEA acknowledges the contribution of the drafting committee. The technical officer responsible for the preparation of this publication was I.D. McLean of the Division of Human Health.

EDITORIAL NOTE

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1. INTRODUCTION

1.1. WHY IS COMPUTED TOMOGRAPHY IMPORTANT?

Computed tomography (CT) scanners create cross-sectional images of high radiographic contrast. This is particularly important for diagnosis involving soft tissue (that is, organs not including lung or bone) as the contrast available from CT images is vastly superior to that gained from projection radiography. Therefore, this type of imaging is medically very useful and increasingly the technique of choice for a growing number of examinations. On the other hand, the dose to the patient may be significantly higher than with alternative imaging modalities. This is of particular importance if the examination involves a pregnant patient or child. The cause of excessively high patient dose can usually be attributed to poor optimization of scanner radiographic protocols, but can also be due to poor equipment condition.

CT scanners are under continual technical development, resulting in increasing clinical application [1] which in turn highlights the need for continual professional education. The increasing complexity of scanner operation and application requires careful monitoring by the medical physicist in conjunction with the radiologist and radiographer to ensure that appropriate examination conditions exist and that procedures are optimized for diagnostic quality and patient dose. To achieve this, it is essential to promote and facilitate the implementation of a quality assurance (QA) programme. This includes appropriate training of radiographers and radiologists, use of well-designed equipment that is in proper operating condition, suitable examination protocols and adequate viewing conditions for image interpretation. The involvement of a medical physicist is a key element in the QA process.

It should be further noted that CT scanners are being increasingly utilized by radiotherapy departments for image acquisition for treatment planning purposes [2, 3], in addition to the traditional roles of patient diagnosis and cancer staging, placing further important demands on scanner performance requirements and QA processes. CT scanners are usually found in diagnostic radiology departments and might be accessed by the therapy department by arrangement. However, once the need for CT based treatment planning increases, as is the case when a significant number of treatment units are utilized in a department, and most importantly, when the percentage of therapy patients receiving curative rather than palliative treatment increases, the use of a dedicated CT simulation unit situated within the therapy department becomes a high priority.

1.2. PURPOSE OF THIS PUBLICATION

There currently exist a small number of established quality assurance (QA) publications, including several on acceptance and quality control (QC) testing for CT [4–8]. These publications form a comprehensive available resource in this area. The IAEA recognizes the different resources and needs of Member States. This publication has been compiled in the light of existing publications and has incorporated the principal components of the existing programmes in a harmonized manner to create a useful handbook for the broad range of Member States. It has been developed with the philosophy that CT imaging must be of the highest quality in order to fulfil the diagnostic tasks expected of it. This publication addresses topics such as the special requirements for scanners used for radiotherapy treatment planning and how to ensure adequate performance in shared diagnostic and radiotherapy scanner utilization and in radiotherapy-only use. In some areas, resources, both technological and human, are limited and therefore this publication was developed with the concept of practical application in mind.

2. CT TECHNOLOGY

2.1. BASIC CT TECHNOLOGY

CT is a mature diagnostic modality that is still undergoing rapid development. The basic principles of CT imaging can be found in many texts [9]. The following is a brief introduction.

CT is a radiographic process that produces a photon attenuation map of the patient based on the variable attenuation of a beam of X rays as it passes through a patient. In order to obtain a cross-sectional image, the beam is restricted to form a thin fan across the patient (in the x–y direction) of between 0.5 mm and 10 mm in thickness, in the case of single slice scanners, to produce a single imaged slice in the axial (z) direction. Many hundreds of attenuation profiles are created in each revolution of the X ray tube around the patient. These profiles are then reconstructed to form the required transverse image (Fig. 1).

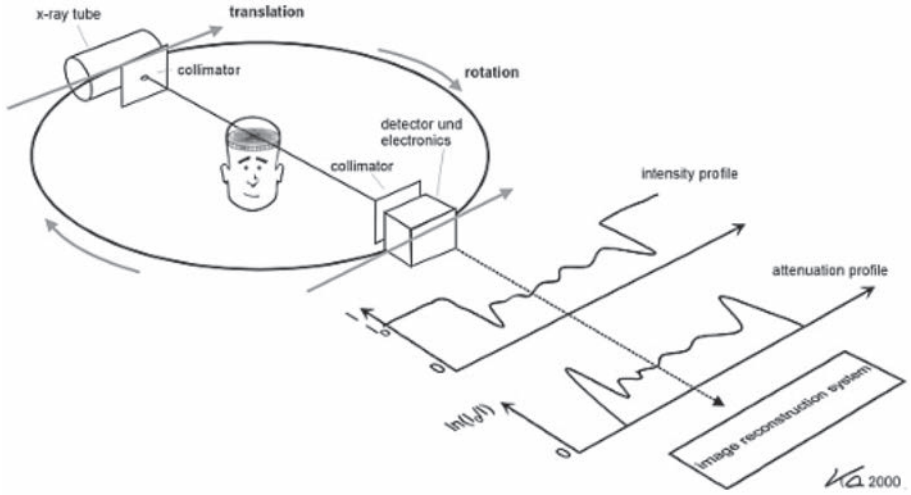


FIG. 1. Simplified diagram of the creation of an attenuation profile in a CT scan. Note that modern scanners use a fan beam to acquire the attenuation profile in one exposure. (Reproduced from Ref. [9] with permission.)

2.1.1. CT gantry: Tube, collimator, filters and detector

The large X ray tube located within the gantry (Fig. 2) operates at between 80 kV and 140 kV. This tube can generate over 109 photons per mm^2 per second at 75 cm from the tube focus for typical CT radiographic settings of tube voltage (120 kV) and current (300 mA[1]).

The X ray tube is typically operated at high voltage and high tube current values for long periods of time, which requires the rapid dissipation of heat to avoid tube failure. The tube cooling system is designed to deal with this. However, it is essential that the ambient temperature around the scanner or heat exchanger be controlled by effective air conditioning to allow optimal operation.

The X ray beam, after leaving the tube, passes through filter material to remove low energy photons. Typically, specially shaped filters are then applied to compensate for attenuation differences in a patient's head or body. It is essential to use the correct filter for the correct body part. The slice width collimator, positioned at the filter exit, determines the width of the X ray beam. In modern scanners, multiple slices (currently up to 320) are acquired simultaneously. These scanners are known as multidetector, multislice or multirow CT scanners. The

[1] Some older CT units may not operate above 175 mA.

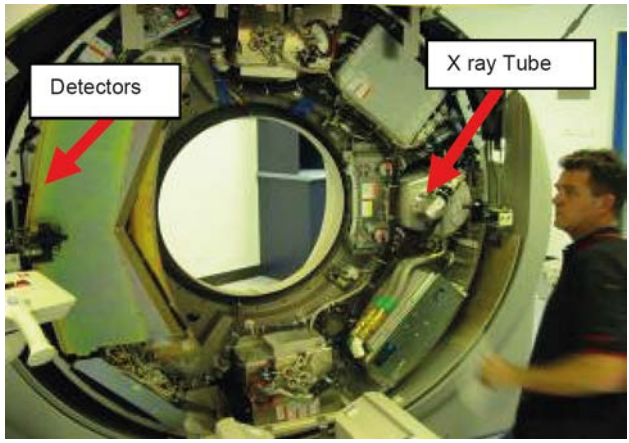


FIG. 2. CT scanner with gantry cover removed. Note the X ray tube on the right hand side with collimator and filters facing towards the scan aperture. Detectors are on the left hand side.

width of the beams for these acquisitions is the product of the individual slice width and the number of slices acquired simultaneously.

The X ray detector element is typically an ionization chamber using high pressure xenon or a scintillation detector. Early scanners used scintillation detectors such as sodium iodide (NaI) or cadmium tungstate ($CdWO_4$); later high pressure xenon generally replaced these early materials and in later years scintillator doped ceramics have been used, such as gadolinium oxysulphide (Gd_2O_2S) or yttrium gadolinium oxide (YGdO). Important specifications for such detector elements, and factors in their development, include a high dynamic range, high quantum absorption efficiency and a fast temporal response with low afterglow.

For a single slice axial scanner, the detector unit will have over 700 elements arranged along an arc to intersect the exit beam of the tomographic plane. This is known as 3rd generation scan geometry (see Figs 3 and 4) and is the basic design for modern CT scanners. In multidetector CT scanners, the detector typically has additional adjoining arcs, or rows, of detector elements. Such multirow detectors may have up to 64 rows, allowing a total acquisition width[2] of 32–40 mm (measured at the isocentre). This type of acquisition can produce slice thicknesses varying from 0.5 mm to 10 mm. With such a detector, the acquisition time is reduced and the occurrence of motion artefacts is considerably reduced.

[2] The cone beam scanner is now available with 320 detectors with a dimension of 160 mm.

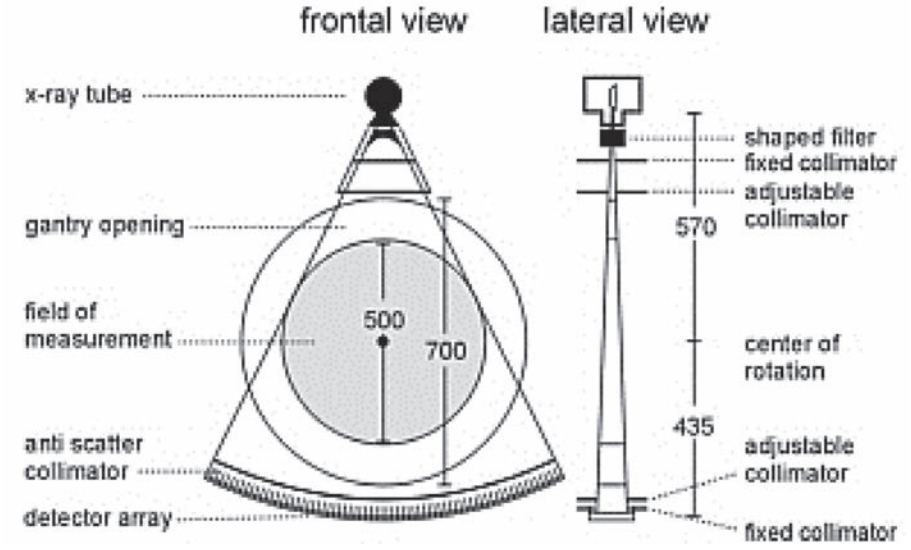


FIG. 3. Schematic representation of the scanning geometry and important components of the CT measurement system in frontal view (x–y plane) and in lateral view (y–z plane). (Adapted from Ref. [9] with permission.)

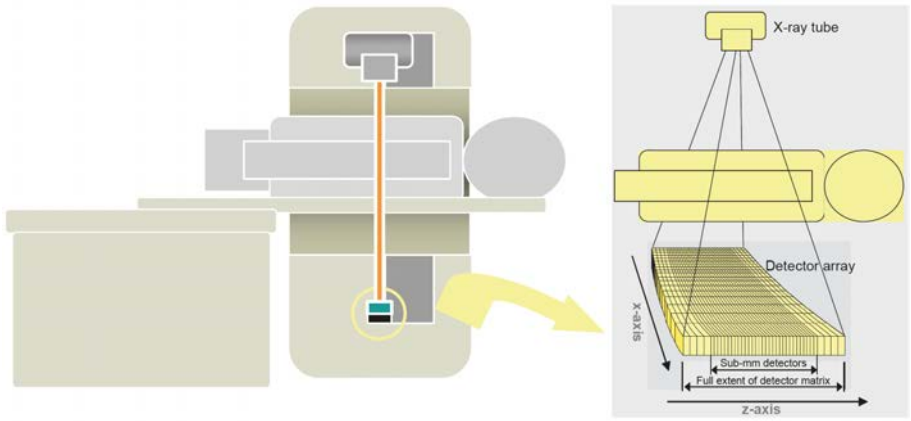


FIG. 4. Schematic representation of the scanning geometry in the lateral view (y–z plane) for a multidetector scanner, demonstrating the multiple detector rows along the z axis.

2.1.2. Image reconstruction

Typically, the reconstruction of an axial image uses projection profiles acquired from a 360° rotation of the tube and detector around the patient.

However, reconstruction is possible with projections of as little as 180º of rotation, while in helical (spiral) CT scanners, variable reconstruction angles are used. The reconstruction is primarily achieved by a filtered back projection method (Fig. 5) that allows almost real time reconstruction, although iterative reconstruction methods are also increasingly being considered now.

2.1.3. Scanning procedures

The simplest image acquisition is the scan projection radiograph (SPR), also known as a Scoutview, Scanogram, Surview or Topogram, depending on the manufacturer. This is taken in order to plan the CT acquisition with the X ray tube and detector moving in one plane relative to the patient (in fact, it is the patient that moves) without any rotation. The SPR (see Fig. 6) differs from a normal projection radiograph since it is divergent in only one direction, whereas the normal projection radiography is divergent in two directions.

The axial slice scan involves acquiring a collection of attenuation profiles around a patient who is stationary on the scan table. This ensures that all the profiles are in the one plane and allows rapid reconstruction computation. The

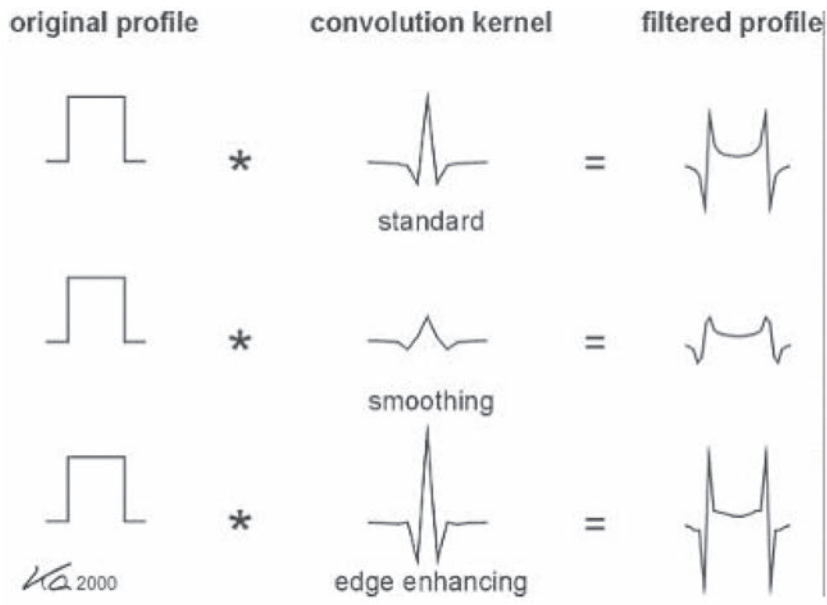


FIG. 5. Image characteristics can be influenced by the choice of convolution kernel, whereby increasing spatial resolution or edge enhancement also means increasing image noise (reproduced from Ref. [9] with permission).

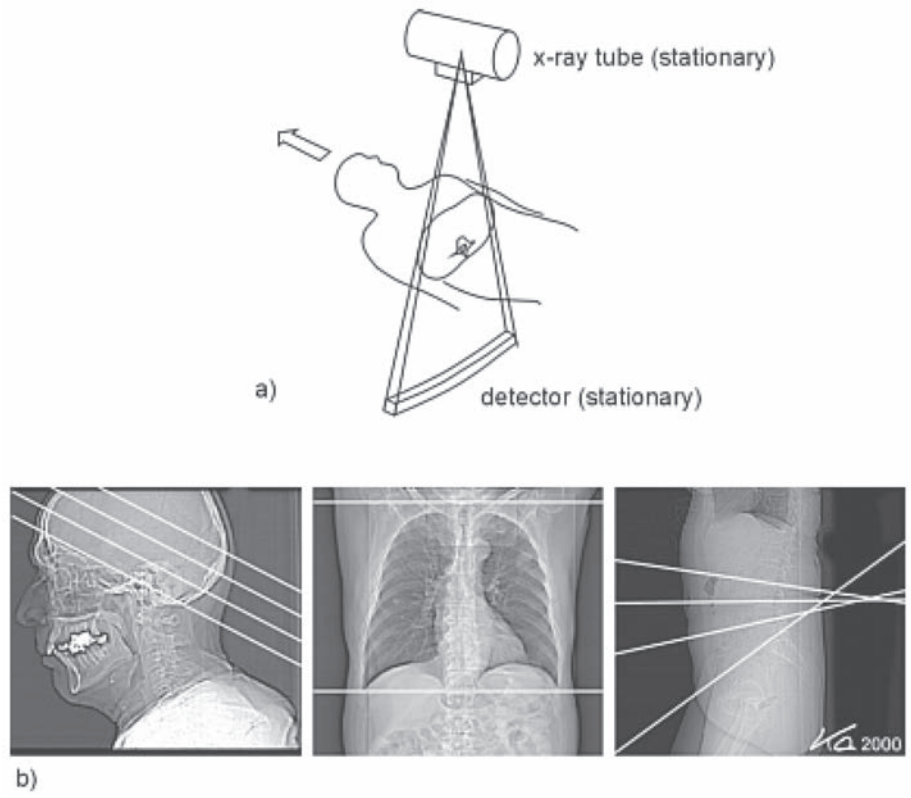


FIG. 6. SPR is taken with low dose and low spatial resolution by transporting the patient through the field of measurement with the X ray tube in a fixed position (a). The projection direction (here AP) is arbitrary in principle. The SPRs allow the selection of the position and gantry tilt for a single slice or for complete scan regions as shown schematically (b) (reproduced from Ref. [9] with permission).

couch is then moved to allow the acquisition of a new slice on a different anatomical region.

Helical, or spiral, scanning is achieved when the couch is moved at the same time as the scan profiles are acquired. This removes some artefacts associated with respiratory motion and allows the scan time to be reduced greatly. However, the reconstruction is complicated as the profiles are no longer in the same plane and may be interpolated (so-called z interpolation) to a pseudo-planar state before reconstruction. Maximum benefit from helical CT acquisition is achieved with multidetector CT (MDCT) acquisition. Here, a process known as z filtering is implemented. These algorithms have different characteristics to axial reconstruction algorithms. One notable feature is the ability to alter the imaged

slice width to any thickness equal to, or greater than, the nominal width of the detector elements used in the original acquisition. This leads to the possibility of generating many sets of images from the one acquisition and can greatly increase the image storage requirements and data management practices of a department.

The key parameter describing helical CT acquisition is pitch, defined as the ratio of the couch advance during a 360º rotation and the width of the total X ray collimation (i.e. the product of the detector width and the number of detector rows). For single slice scanners in particular, a large pitch implies faster acquisition with reduced dose, but at the cost of reduced resolution in the z axis, while low pitch has slower acquisition and increased dose but with better z axis resolution.

2.1.4. Image creation with MDCT scanners

As mentioned above for MDCT scanners, the use of reconstruction algorithms, such as z filtering, allows the reconstruction of CT images at virtually any thickness that is equal to, or greater than, the acquisition slice width of a particular detector as defined at the time of acquisition. This leads to the common practice of scanning the patient using small or 'thin' detector slice widths and then reconstructing the acquired data set firstly for a displayed slice width the same or similar to that of acquisition (so-called thin slices) and also at a thicker slice width (so-called 'thick' slices). Thin slices may display more image noise than thick slices. However, the thin images do not suffer from partial volume effects that reduce the contrast of small dense objects within the patient.

The management of image data generation for an MDCT needs to be carefully considered (see Section 4.1), especially when the increment distance between the start of each reconstructed slice can be selected after acquisition, as well as the possibility of using a different reconstruction kernel for additional image data sets. This offers the possibility of creating very large quantities of image data, with the use and storage of these data becoming a key issue within a clinical setting.

2.1.5. Automatic exposure control

Recently, it has been possible to control automatically the tube voltage (kV), tube current (mA) and exposure time (s) to suit individual patient sizes, from patient to patient, along a specific patient or through different angles at any particular cross-sectional position. This is often described as a form of automatic exposure control (AEC) [9–11] similar in concept to that used in regular radiography. There are a number of ways that this is achieved, including the use of X ray tube current modulation that responds to measured patient attenuation

conditions [12, 13]. Some scanners use the SPR view to determine exposure conditions, and many scanners require operator input to function. The use of automatic functions should be fully understood and their dose implications investigated before being put into general use.

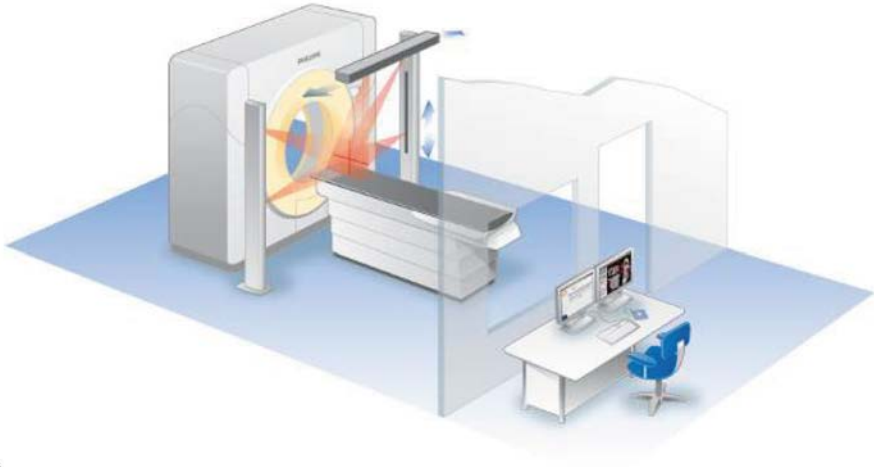
2.2. CT SCANNERS IN RADIOTHERAPY TREATMENT PLANNING

Shortly after the introduction of clinical CT scanners in the early 1970s, it was realized that this imaging modality has much to offer in a radiation oncology setting. CT images provide volumetric information not only about tumour (target) volumes but also about normal (critical) structures. The use of CT images in radiotherapy treatment planning has improved dose delivery to target volumes while reducing the dose to critical organs. CT images can be converted to provide relative electron density information for heterogeneity based dose calculations. However, CT image contrast is based on relative X ray attenuation and so, in some cases, CT soft tissue contrast is not sufficient. This limitation can be overcome by using CT images in conjunction with magnetic resonance studies for treatment planning. PET images can be used to add physiological information.

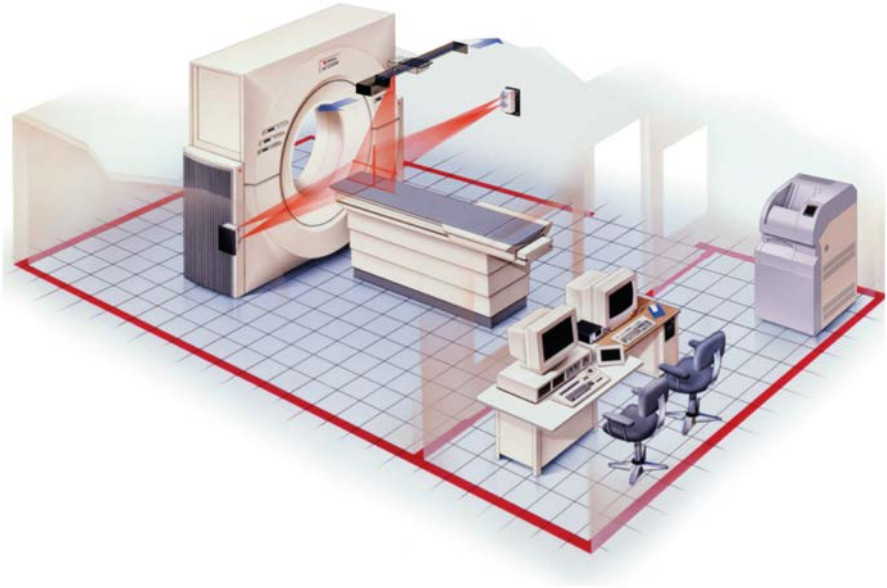
Tatcher [14] proposed treatment simulation with CT scanners. This short article described the feasibility of CT simulation and indicated the potential economic and treatment planning benefits. In 1983, Goitein and Abrams [15, 16] further described multidimensional treatment planning based on CT images. Sherouse et al. [17, 18] went on to describe a CT image based virtual simulation process which they referred to as a "software analog to a conventional simulation". This series of publications described software tools and addressed technical issues that affect today's CT simulation process. The manuscripts pointed out the need not only for fast computers and specialized software, but also for improved patient immobilization and set-up reproducibility.

CT simulators have matured to a point where they are one of the cornerstones of modern radiation oncology facilities. Today's systems incorporate specially designed large bore CT scanners, MDCT scanners, high quality laser positioning systems and sophisticated virtual simulation packages. Figure 7 shows a modern CT scanner equipped with a flat couch top and an external patient positioning and marking laser system.

Additional virtual simulation software features and functions along with increased efficiency and flexibility have enabled CT simulators to replace conventional simulators in many facilities. This trend seems to be further fuelled by the increased demand for imaging studies for conformal three dimensional



(a)



(b)

FIG. 7. CT simulator equipped with a flat couch top, external patient positioning/marking system and movable lasers (a) and fixed wall lasers (b) and specialized CT simulation software. (Courtesy: Philips Medical Systems.)

(3-D) and intensity modulated radiation therapy treatment planning where conventional simulators are of limited value [19].

The implementation of CT simulation and a treatment planning process varies greatly between radiation oncology departments. This diversity is in part driven by significant technical differences between simulation and treatment planning systems offered by different manufacturers and by the technology available at various facilities. Furthermore, the difference in implementation is affected by whether the scanner is used for diagnostic and radiotherapy imaging or for radiotherapy imaging alone. This publication addresses QA concerns for various available technologies and clinical settings.

3. PERFORMANCE REQUIREMENTS IN CT

3.1. SCAN AND PATIENT POSITIONING ACCURACY

3.1.1. CT alignment lights, SPR accuracy

All scanners have patient positioning lights which identify the tomographic plane. These are often referred to as the 'internal lights'. There are often other lights that identify the sagittal and coronal planes, as well as a further set of lights identifying a transaxial plane, set at a fixed distance from the tomographic plane. These are usually called the 'external' lights. Once a patient or phantom is aligned to the external lights, the tomographic plane, as defined by the internal lights, can usually be reached by a single button press.

Testing the quality of the scanner involves establishing a level of certainty in the position of the scanner lights and their accuracy against requirements.

The lights are either tungsten or laser. (Where laser lights are used, safety aspects need to be considered. Generally, the lights are classified as safe for use where the blink reflex is used as the body's natural protection. Care is needed for unconscious or anaesthetized patients).

In clinical practice, the lights are usually used as a rough guide for setting up the patient. The patient is first aligned on the scanner couch using the scanner tomographic plane lights, in order to establish which region of the body to scan, to centre the patient to the isocentre for best image quality and sometimes to avoid irradiating the eyes in brain scanning.

Precise preparation for cross-sectional CT scanning is taken from the SPR, and this is used to define the beginning and the ends of the required scanned volume.

When carrying out a number of tests on a scanner, it can be helpful to perform the light alignment checks first, in order that the lights can be used to assist in setting up the phantoms.

Clinically accurate light positioning is required in diagnostic imaging for biopsy location and for avoiding radiosensitive organs such as the eyes in brain scanning.

3.1.2. External CT positioning lasers for radiotherapy positioning

When a scanner is used for radiotherapy treatment planning purposes, the accuracy required for an external positioning marker is such that a set of laser lights is usually fixed to the walls and ceiling. These are then adjusted and set independent of those in the gantry housing. (These latter lights can be prone to accidental misalignment during a scanner service.) However, they can be tested in the same way as those that are integral to the scanner.

The external patient positioning/marking lasers consist of three sets of lasers: axial, coronal and sagittal. The lasers can be fixed or movable (Fig. 7). It is common for the sagittal laser to be movable in either direction. This feature is needed to enable marks to be placed on the patient's skin away from the midline and, since the CT couch tops do not move in a lateral direction, it is desirable for the sagittal laser to be movable. The other two wall lasers can be movable as well, but this feature is far less critical since the couch moves in these two directions.

External patient positioning/marking lasers for radiotherapy applications are used to establish a relationship between marks on the patient's skin (which are used for positioning in radiotherapy treatment) and anatomy on the CT images. It is often necessary for the accuracy of this relationship to be within 2 mm. Therefore, the relationship between the external lasers and the CT tomographic plane should be accurate to within 2 mm. With correct alignment and verification procedures, this accuracy should be easily achievable.

Other than quantitative CT number accuracy, geometric imaging accuracy, and the need for a flat couch top, the alignment of external lasers with the CT tomographic plane is probably the most significant difference between diagnostic and radiotherapy CT performance specifications. Inaccurate alignment of external lasers and the CT tomographic plane can be a significant source of systematic errors in radiotherapy treatments.

3.2. IMAGE QUALITY

3.2.1. CT number

The CT number (H_s) of a sample of material s is defined by the expression:

$$H_s = K \frac{\mu_s(E) - \mu_w(E)}{\mu_w(E)} \quad (1)$$

where $\mu_s(E)$ and $\mu_w(E)$ are the linear attenuation coefficients at the energy of the X ray beam for water and the scanned sample, respectively and K is a constant, which has a value of 1000 if the CT value scale is in Hounsfield units.

The Hounsfield unit scale is the accepted scale on all modern CT units. The exceptions were for very early scanners and for a very few modern scanners where the K factor is halved in the application of certain high resolution convolution kernels.

From Eq. (1), the CT number for water is zero and, since the attenuation is negligible for air, the CT number (Hounsfield unit) for air is –1000.

The attenuation process for CT is dominated by Compton interactions for soft tissue, with some photoelectric interactions for materials of higher atomic number (Z). Compton interactions are independent of atomic number, proportional to electron density and inversely proportional to the energy, E. The photoelectric effect is approximately proportional to $(Z/E)^3$.

In radiotherapy treatment planning, in order to compute the treatment dose distribution from a CT image, the relationship between relative electron density and CT number needs to be established, since the Compton effect is the predominant X ray interaction at radiotherapy energies.

3.2.2. Image noise

When a uniform material is imaged on a CT scanner, examination of the CT values for individual pixels in a localized area shows that the CT numbers are not all the same, but fluctuate around a mean value (Fig. 8). This random variation is known as image noise and is due primarily to the statistical nature of X ray production and interaction with matter. It is also known as quantum noise.

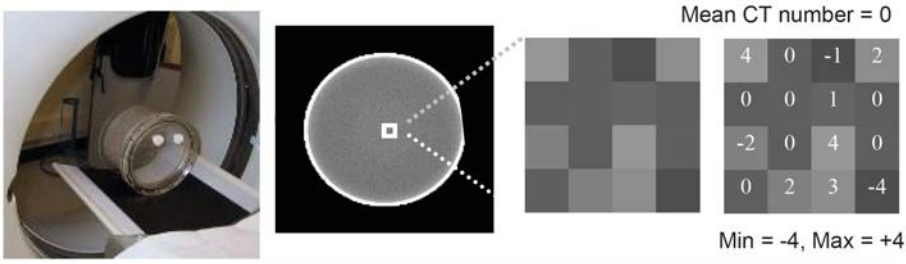


FIG. 8. Variation in grey level from a water phantom (CT number).

In addition, other sources of noise may include structured noise or artefacts and electronic noise[3].

In this publication the **definition of noise as the standard deviation of Hounsfield numbers (σ)** within a region of interest (ROI) is adopted [7]. However, in order to make a direct comparison between CT scanners with different contrast scales, a normalized standard deviation, *S*, is required:

$$S = \frac{\sigma_{water}}{CT_{scale}} \times 100\% \tag{2}$$

where σ_{water} is the standard deviation of the pixel values within an ROI and

$$CT_{scale} = CT_{water} - CT_{air} \tag{3}$$

where CT_{water} and CT_{air} are the CT values obtained for water and air, respectively.

Since, by definition, water has a value of zero and air a value of –1000 when measured in Hounsfield units, then CT_{scale} has a value of 1000. Therefore, the normalized standard deviation becomes a very simple calculation. Some manufacturers, however, halve the CT number range for certain high resolution modes, which can be confusing when no indication is given to the user and needs to be considered in the normalized standard deviation calculation.

The percentage normalized standard deviation is identical to The American Association of Physicists in Medicine expression, σ_{uw}% [21], providing that the CT_{scale} is linear with respect to the linear attenuation coefficient.

[3] Electronic noise is found to be minimal for modern scanners [20], although it can become more significant when scanning at very low tube currents.

3.2.3. Contrast to noise ratio (CNR)

CT scanning utilizes a large photon flux in acquisition in order to achieve low noise images. However, this results in higher patient doses. These images allow the identification of low contrast structures, reflecting very small differences in photon attenuation in the tissue due to composition or density differences. While the image noise in a uniform material is usually a good indicator of the ability to visualize small contrasts in diagnostic images, a more versatile measure is that of CNR. To measure CNR, the contrast of two objects is determined by the difference of the mean CT numbers within selected ROIs and is divided by the average noise for these two ROIs:

$$\mathrm{CNR} = \frac{\mathrm{CT}_1 - \mathrm{CT}_2}{(\sigma_1 + \sigma_2)/2} \tag{4}$$

This parameter is useful when optimizing a CT examination protocol for a particular contrast situation e.g. tissue density contrast, iodine contrast and air tissue contrast.

3.2.4. Uniformity of image noise and CT number

A scan of a water filled phantom should give a CT image with similar pixel values, and similar amounts of noise, across the whole field of view. However, in practice, scans of uniform phantoms often show gradual variations of CT number and noise values across the image.

These variations may be particularly noticeable when the uniform phantom is surrounded by a high contrast material, such as cortical bone substitute material. Variation is also noticeable at the extremities of large phantoms, particularly when an exceptionally large phantom is used to investigate the extended fields of view used on the large bore scanners that are sold for radiotherapy use.

If a phantom, or patient, is not centred at the isocentre, a more pronounced variation of CT number and image noise is also likely to be observed.

The uniformity of CT number is of importance when the scanner is used for quantitative assessment of CT values, particularly for radiotherapy.

3.2.5. Image artefacts

Artefacts are features on the image which do not represent true tissue structure. At best they are a nuisance to the radiologist trying to interpret the image; at worst they are misleading and give rise to false or missed diagnoses.

There are a number of causes of image artefacts. To avoid them, as much as possible, it is essential to have a good installation process, give attention to electronic components and servicing, and conduct regular calibration. At installation, it is essential that room preparation be carefully considered. The room must be stable, level and able to withstand the weight of the scanner. During assembly, the alignment of components is critical and scans must be carefully examined during the acceptance and early clinical phase to check for any indication in phantom images. Electronic component failure or poor board connections can give rise to artefacts. Regular calibration by the servicing engineer and daily calibration by the radiographer will avoid certain artefacts occurring. Visual artefacts may appear due to the quality of the image monitor. The image display monitor, therefore, needs to be included in the whole QA/QC programme.

Image artefacts in CT are generally separated into four descriptive categories; streaking, shading, rings and aliasing. They generally have a variety of causes [22].

Streaking artefacts tend to be caused by inconsistencies in neighbouring projections, either by a high attenuating region such as a metallic implant, or by patient movement. These can also be caused by scanner movement or misalignment of the equipment.

Shading artefacts tend to be due to inappropriate beam hardening corrections, giving a gradual change in accurate CT numbers across a phantom or patient. It is normal practice for calibrations to be carried out by the installation engineer for every scanning factor or combination that will be used in clinical protocols (e.g. voltage (kV), current (mA), slice thickness), to ensure that the correct beam hardening factors are implemented.

Ring artefacts are very common and are generated by detectors that have differing sensitivities relative to each other. The easiest way to avoid most of these is to ensure that the ‘air calibrations’ are run whenever required by the scanner manufacturer; this is usually once a day, or sometimes more frequently or when flagged by the system.

Aliasing artefacts generally appear with high resolution algorithms and merely show the mismatch of the physical aperture to the mathematical filter applied to the projections.

Some examples of artefacts are shown in Figs 9–12.

3.2.6. Spatial resolution

Spatial resolution is the ability of the CT system to create an image of an object without loss of spatial information or 'blurring'. It usually refers to high contrast objects and is defined in the tomographic plane, although 3-D scan reconstruction allows spatial resolution to be considered more generally.[4] The determination of spatial resolution can be made through the use of a high contrast test object through visual inspection or computation to compute the modulation transfer function (MTF) [23, 24], or with the use of an appropriate test object to

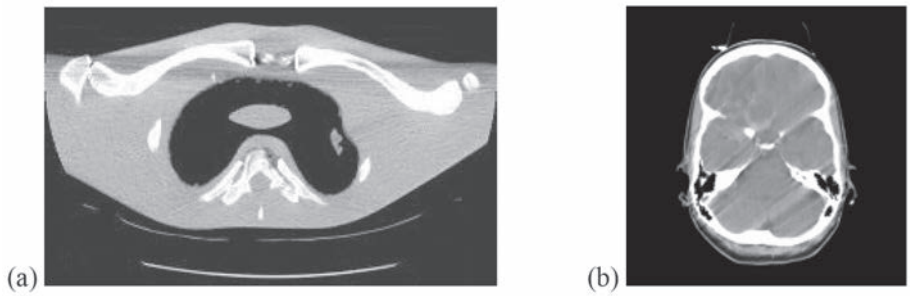


FIG. 9. Examples of streaking artefacts caused by (a) highly attenuated projections through the lateral view of the thorax and (b) patient movement.

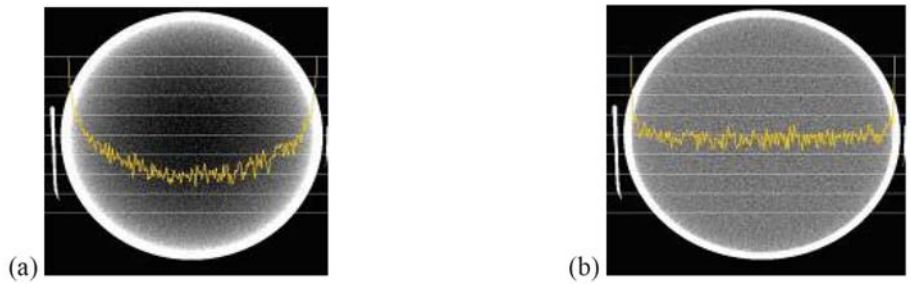


FIG. 10. Example of (a) beam hardening artefact through a water filled phantom and (b) after beam hardening correction applied.

[4] Resolution in the axial direction is often termed z axis resolution and is determined by the imaged slice width. Modern MDCT scanners should be capable of achieving isotropic resolution where the z axis resolution is equal to, or approaches, the tomographic plane resolution. This is essential for good quality multiplanar and 3-D reconstructions.

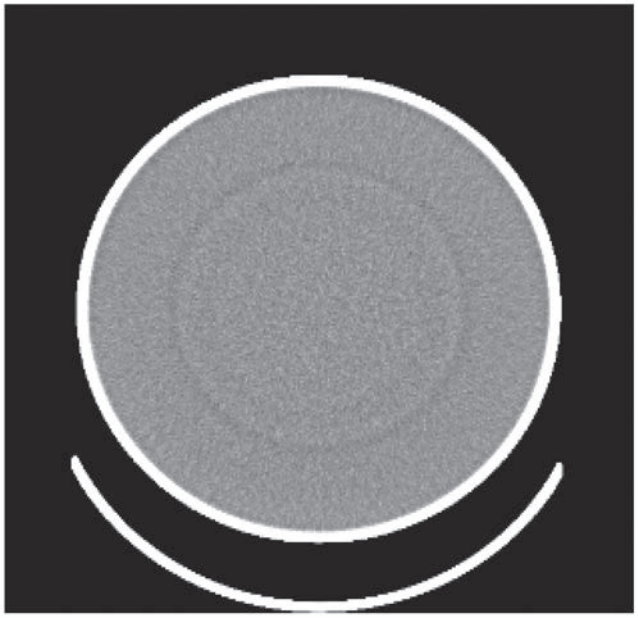


FIG. 11. Example of ring artefacts in a water filled phantom.

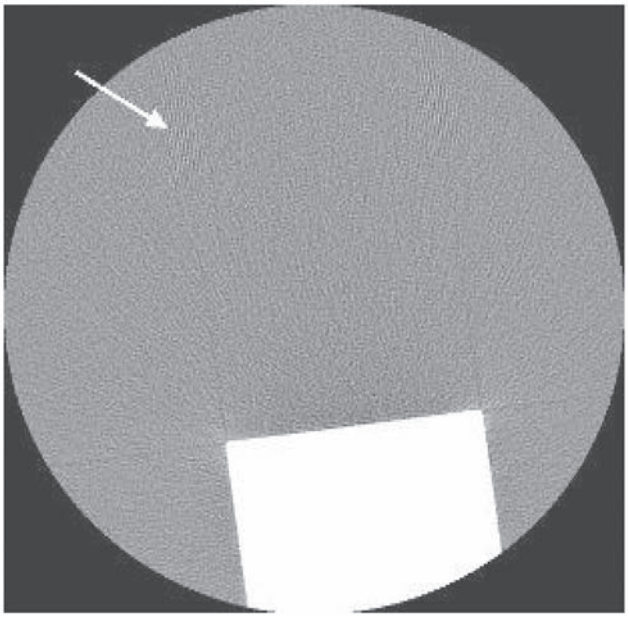


FIG. 12. Example of aliasing artefact.

compute the MTF. The MTF is commonly derived from the image of a bead or wire to give a point spread function (or of an edge to give an edge spread function). Standard methods are then used to compute the MTF from either the point spread function or the edge spread function. Comparisons can then be made of the frequencies at which the MTF curve falls to the 50% and 10% levels. If a high contrast test object is used, the resulting quantity is line pairs per millimetre (lp/cm), and if MTF is determined, the quantity is cycles per centimetre (c/cm).

Calculating the MTF is critically dependent on the ability to access the numerical data and the scanner, as well as having a suitable analysis program [4]. This has become less of a problem in recent years with the advent of the DICOM image transfer standard and the availability of functional software on personal computers. Only one or two manufacturers calculate the MTF as a standard facility on their QC software packages.

3.2.7. Partial volume effects

The partial volume effect, often mentioned in relation to the artefact that it creates, is a consequence of the finite thickness of a CT slice or can be considered as a type of blurring in the z axis. This effect is prominent when viewing objects that are only partially within the image slice, or that are smaller than the image slice thickness. For example, a small blood vessel filled with iodine will have a reduced contrast in a thick slice compared with a thin slice, owing to the overlying tissue of the thick slice 'averaging' the density over the full voxel.

Consequently, the use of thin slices reduces this effect considerably, although it comes at the cost of either higher noise or higher patient dose. The spatial resolution in a CT image is currently limited to approximately 0.5 mm, which is inferior to most other radiological procedures but better than that possible with radioisotope imaging. CT scanners can also acquire slices of thin tissue thickness (minimum 0.5 mm), which allows very precise cross-sectional delineation of structures without interference from partial volume effects.

3.3. CT DOSE

Dosimetry for CT has recently been specified by the International Commission on Radiation Units and Measurements [25] with a complementary code of practice recently published by the IAEA [26]. In these publications, the use of a dose index is formulated that can be a powerful tool in the optimization of CT examinations. The use of these indicators to obtain organ and tissue information is discussed briefly in Section 3.3.3.

3.3.1. Standard CT dosimetry

The CT air kerma index, $C_{a,100}$, measured free-in-air for a single rotation of a CT scanner is the quotient of the integral of the air kerma (K(z)) along a line parallel to the axis of rotation of the scanner over a length of 100 mm and the total collimated beam width, NT, where N is the number of data slices or tomographic sections and T is the nominal detector width. The integration range is positioned symmetrically about the volume scanned, thus

$$C_{a,100} = \frac{1}{NT}\int_{-50}^{+50} K(z)\,dz \tag{5}$$

The CT air kerma index is also measured inside PMMA head and body phantoms and is defined similarly to $C_{a,100}$ (Eq. (1)). The notation $C_{PMMA,100}$ is used as shown in Fig. 13.

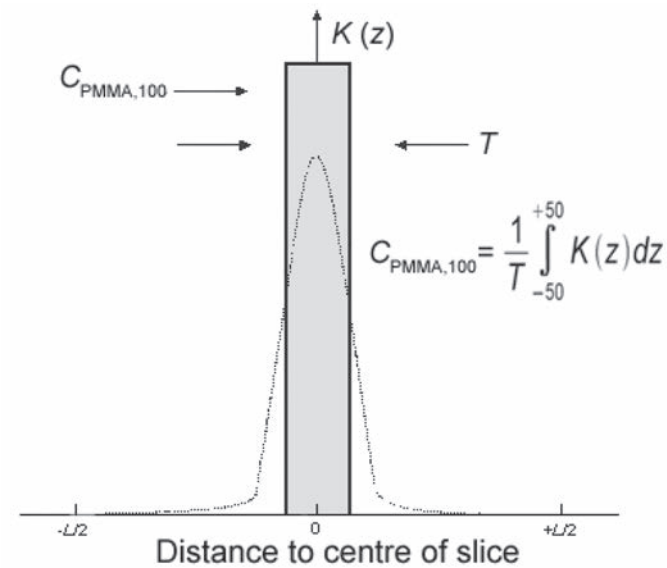


FIG. 13. Profile of air kerma, K(z) measured in the standard CT dosimetry phantom along an axis parallel to the CT scanner axis (z) for a single CT slice of nominal width T mm. The CT air kerma index, $C_{PMMA,100}$, is obtained by integrating the air kerma over a length (L) of 100 mm (reproduced from Ref. [26]).

The weighted CT air kerma index, C_w, combines values of $C_{PMMA,100}$ measured at the centre and periphery of a standard computed tomography dosimetry phantom [27–30]. It is given by

$$C_w = \frac{1}{3}\left(C_{PMMA,100,c} + 2\ C_{PMMA,100,p}\right) \tag{6}$$

The quantity $C_{PMMA,100,c}$ is measured at the centre of the standard CT dosimetry phantom and $C_{PMMA,100,p}$ is the average of the values measured at four positions around the periphery of the same phantom. A weighted "computed tomography dose index" was first introduced by Leitz et al. [31] and is used in an IEC report [30] and the European CT dosimetry protocol [28] (see Table 1 for a comparison of the units).

The subscript *n* is used to denote when the value of $C_{a,100}$ or C_w has been normalized [28] to unit tube current–exposure time product, P_{It}, thus

$$_nC_w = \frac{C_w}{P_{It}} :\ _nC_{a,100} = \frac{C_{a,100}}{P_{It}} \tag{7}$$

A further quantity, C_{VOL}, takes into account the helical pitch or axial scan spacing [30, 32], thus

$$C_{VOL} = C_w \frac{NT}{l} = \frac{C_w}{p};\quad _nC_{VOL} = \frac{C_{VOL}}{P_{It}} \tag{8}$$

where

N is the number of simultaneously acquired tomographic slices;
T is the nominal slice thickness;
l is the distance moved by the patient couch per helical rotation or between consecutive scans for a series of axial scans;
P_{It} is the tube loading for a single axial scan.

The quantity

$$p = \frac{l}{NT} \tag{9}$$

is known as the CT pitch factor (or pitch) for helical scanning.

TABLE 1. COMPARISON OF IAEA AND IEC DOSIMETRY TERMINOLOGY USED IN CT

Quantity	IAEA	IEC
Measured free-in-air:		
CT air kerma index	$C_{a,100} = \frac{1}{NT}\int_{-50}^{+50} K(z)\,dz$	$CTDI_{air} = 1/NT \int_{-\infty}^{+\infty} K_a(z)\,dz$
Measured in standard phantom:		
Weighted CT air kerma index	$C_w = \frac{1}{3}\left(C_{PMMA,100,c} + 2\ C_{PMMA,100,p}\right)$	$CTDI_w = 1/3\ CTDI_{100,c} + 2/3\ CTDI_{100,p}$
Normalized weighted CT air kerma index	${}_nC_w$	${}_nCTDI_w$
Volume CT air kerma index	C_{VOL}	$CTDI_{VOL}$
CT air kerma–length product	$P_{KL,CT} = \sum_j {}_nC_{VOL_j} l_j P_{lt_j}$	$DLP = CTDI_{VOL}\ L$

The CT air kerma–length product, determined for the standard CT dosimetry phantom and a complete CT examination, $P_{KL,CT}$, is calculated using

$$P_{KL,CT} = \sum_{j} {}_{n} C_{VOL_j} \, l_j P_{It_j} \tag{10}$$

where the index j represents each serial or helical scan sequence forming part of the examination, l_j is the distance moved by the patient couch between or during consecutive scanner rotations and P_{It_j} is the total tube loading for scan sequence j. This quantity is analogous to the 'dose length product' introduced in EC guidelines [28] and used in the IEC CT safety and performance standard [33].

3.3.2. Understanding the concepts of CT dosimetry indices

The CT air kerma index (or CTDI) is calculated from a measurement made with a pencil ion chamber, irradiated by a single rotation of the X ray beam, at the central position along the chamber's length. The integration length in the definition of C (or CTDI) matches the length of the pencil ion chamber (100 mm).

The interpretation of this measurement is that it gives an equivalent value to the z average dose at the central slice position, as though a 100 mm length had been scanned with contiguous X ray slices.

The weighted value (C_w) looks at the CT air kerma value in a phantom, and takes into account the variation in dose over the cross-section of the phantom. So it represents an averaged x–y plane (scan plane) dose, averaged along the z axis at the central slice position, as though a 100 mm length had been scanned with contiguous X ray slices.

The volume CT air kerma index (C_{VOL}) goes further, considering the CT air kerma value in a phantom, but in this instance taking into account non-contiguous irradiations along the z axis, i.e. pitch. It therefore represents the average of the dose in the x–y plane and the z average dose at the central slice position as though a 100 mm length had been scanned at a given pitch.

3.3.3. Dosimetry for CT scanners with large collimation beam widths

The early development of the CTDI started when the slice width of a CT scanner was typically 10 mm or less. It was originally developed for head scans and the integration length was selected to be similar to the length of a head examination. It was subsequently changed to 100 mm with the development of the 100 mm pencil ion chamber. The contribution of radiation falling outside the active length of the pencil chamber was considered negligible for beam widths up

to about 20 mm. However, with the advent of MDCT scanners with collimation beam widths of up to 40 mm, there was concern that there would be more scatter not included in this measurement and calculation. However, it has since become apparent that beam widths up to about 80 mm are equivalent in the proportion of radiation that is excluded [34, 35].

In addition, for all these beam widths there is a significant amount of radiation not included in the measurement of $C_{\mathrm{PMMA},100}$ and this leads to a systematic difference in dose compared with a long integration length. This underestimate is of the order of 30–40% for a body phantom and 10–20% for a head phantom [34, 35].

The maximum dose values are reached at the equivalent irradiation lengths of 300 mm for the body phantom and 160 mm for the head phantom.

In concept, the CT air kerma index (or the CTDI in other terminology) represents the average dose in the central slice of a 100 mm scanned volume. The CT air kerma index does not represent the average dose in the central slice region for a longer scanned volume.

The current use of cone beam scanner technology with collimation beam widths of 160 mm, for example, clearly demonstrates the deficiencies of the standard CT dosimetry methodology.

The IEC has proposed a solution to this problem [36], one which has been adopted and expanded on by the IAEA [37] as an interim solution pending the emergence of an alternative dosimetric model. This methodology gives two interpretations of the definition of $\mathrm{CTDI}_{\mathrm{w}}$ (C_{w}) depending on the beam width. For beams of less than 40 mm, the conventional definition is given. For a beam width greater than 40 mm, C_{w} can be written as:

$$C_{\mathrm{w},NT} = C_{\mathrm{w,Ref}} \times \left(\frac{C_{\mathrm{a},100,NT}}{C_{\mathrm{a},100,\mathrm{Ref}}} \right) \tag{11}$$

where $C_{\mathrm{w,NT}}$ is the weighted CT air kerma index for a beam width of NT mm (if NT > 40 mm), $C_{\mathrm{w,Ref}}$ is the weighted CT air kerma index for a reference beam width of 20 mm (or closest possible below 20 mm), and similarly, $C_{\mathrm{a},100,\mathrm{NT}}$ is the CT air kerma index measured free-in-air with a 100 mm integration length chamber for a beam width of NT mm and $C_{\mathrm{a},100,\mathrm{Ref}}$ is a similar quantity at the reference beam width. The methodology on measurement techniques is given in Ref. [37].

Others have advocated the use of pencil chambers with 300 mm of active length. These detectors also require phantoms of up to 350–450 mm in length. The use of small (point) detectors, along with a new dosimetry formalism, is also currently being actively pursued [35, 38, 39]. However, recent work [40] has

brought into focus the large dose variations found at, and near, the surface of phantoms undergoing CT scanning on MDCT scanners, which may limit the use of small detectors for general dosimetry unless a tightly overlapping pitch is used for the measurement.

3.3.4. Calculation of organ doses from CT scanners

It must be remembered that measures such as C_{VOL} are indicators of dose only. They are measured within a standard sized PMMA phantom, and represent the dose as though a 100 mm length of the phantom had been scanned. Although they are not measures of the dose to organs or tissue received by the patient, or even a population of patients, they do, however, have a use in comparison of doses delivered by different scanners and protocols.

While organ doses can be measured for specific patient categories with the use of thermoluminescent dosimeters and appropriate anthropomorphic phantoms [41], the use of Monte Carlo modelling is a more usual approach. To date, a number of centres [42, 43] have used mathematical models to simulate typical patients and determine conversion factors that allow organ dose to be calculated from a reference dosimetry index, such as $C_{a,100}$ or C_w for a given set of scanner factors, patient position variables and scan length parameters. This process has been described in the literature [1] and has been applied to a variety of patient models [44–46] and includes paediatric dose estimation. A number of calculation engines are available, as listed in Appendix IV. As mentioned above, the viability of these conversion factors is currently problematic for some MDCT scanners owing to the wide beam widths, variability of peripheral dose patterns and use of modulating tube current under the control of automatic exposure systems.

3.3.5. Dose calculations for CT scanners using some form of AEC

Dose calculations are necessarily more complicated when AEC systems are in place. Dose distribution within a slice will change with rotational aspects of AEC systems and will change along the patient with the z axis modulated tube current. Detailed analysis can be undertaken by investigating the average tube current for each slice, which is often given. However, this can be time consuming. Also, the information on the rotational variation of the tube current is often not available or accessible.

Depending on the requirement, some pragmatic approaches can be taken. For example, either the average tube current can be used or, since the maximum and minimum current values are often given, best case and worst-case calculations can be undertaken. This is usually sufficient for most purposes.

3.3.6. Dose parameters and DICOM headers

A DICOM standard supplement was issued in 2007 for the reporting of dose parameters in CT [47].

This requires a report summary to be given for the whole patient examination and the accumulated dose applied. The patient information, the patient study information and the general equipment information is stored within the general part of the structured report.

For each CT acquisition (irradiation event), it is required that the $CTDI_{VOL}$ and dose–length product be stated, along with the corresponding $CTDI_{air}$.

Also, the total dose–length product of a patient examination is given and the accumulated irradiation effective dose in millisieverts may also be given. This is optional and, if given, the method of evaluation needs to be stated.

The supplement requires that examination details be given, including: the type of examination, the patient region examined, the parameters of dose evaluation, the dose of a single event and the information relating to the acquisition sequence (sequential, helical or fluoroscopy). Work on refining the DICOM header for CT and associated structured reports on dose is ongoing.

4. CONSIDERATIONS FOR THE SELECTION OF A CT UNIT

4.1. FACTORS TO CONSIDER IN CHOOSING A CT SYSTEM

While CT has been a mature imaging modality for a long time, technological developments in this field continue to be rapid and broad based. Current and past commercial offerings range from single slice to multidetector scanners with a multitude of X ray tubes, generators, detector types, reconstruction hardware and software, gantry sizes, filming/archiving options and accessory offerings. No single scanner is ideal for all clinical settings and requirements. Selection of an inadequate scanner can result in unnecessary upfront and reccurring expenses, inability to provide adequate support and maintenance of the scanner and, most importantly, an inability to provide optimal service to patients. Selection of an ideal scanner for an individual facility can be a daunting task and involves consideration of factors such as clinical needs and technical parameters, operational setting, and staff and cost considerations.

4.1.1. Clinical needs

Consideration of an individual institution's clinical needs should be the primary concern in all CT scanner selection processes. The commercially available equipment is designed to meet diverse clinical needs, with some software and hardware applications being highly specialized with respect to specific medical procedures. Evaluation of clinical needs includes consideration of:

(a) *Patient workload*: The numbers of patients examined on a CT scanner in a clinical practice varies considerably. Often, new scanners are required in response to an increased demand for services. Clearly, the need for rapid scans and rapid image reconstruction will be partially driven by daily patient workload. Less busy clinics may be able to function quite well with slower image acquisition and reconstruction options. An exception here is scanners which will be used for emergency/trauma and cardiac scanning, where speed may be a critical requirement. Image storage capacity and the size of the image database will also be driven by the number of scanned patients.

(b) *Patient population*: CT scanners are offered with a variety of X ray generator and X ray tube configurations. Some of these configurations offer the capability for generation and maintenance of high tube currents (mA) for long periods of time. These systems are designed to accommodate populations where the average patient is larger than the standard 70 kg patient or where long scan volumes or prolonged scanning is needed. Larger X ray generators and X ray tubes may not be needed in regions where average patient size is less and where long scan volumes and prolonged scanning are not utilized [48].

(c) *Clinical applications*: All manufacturers offer scanner hardware and software features which are designed for specific clinical applications (cardiology, neurology, angiography, thoracic imaging, virtual endoscopy, orthopaedics, trauma, image guidance of interventional procedures, oncology, etc.) [49]. A key element to the selection of the uses to which the CT scanner is to be put is the effect the services of the scanner will have on existing and future services and the general reduction in hospitalization due to the non-invasive nature of CT procedures. All involved clinical departments should be consulted when selecting a CT scanner to ensure that their individual needs are met. This should be performed as early as possible, as delays may result in wasted efforts or inability to evaluate desired features adquately. Examples of the complexity of requirements range from 2-D reconstructions that are available from the scanner console

to 3-D constructions and software packages that are often carried out on an ancillary computer workstation. The range of computer software packages is extensive and usually optional. While inclusion of unnecessary options typically results only in increased cost and complexity of operation, exclusion of necessary features can result in suboptimal patient care and inability to perform certain procedures. Special radiation oncology considerations are addressed in Section 4.2.

(d) *Technical parameters*: The technical parameters given consideration when selecting a CT scanner are substantial and include the total scan time and scan length, scanner image quality and radiation dose issues [50–54]. Expert advice should be sought from a knowledgeable and independent source, such as a medical physicist specializing in diagnostic radiology.

4.1.2. Operational setting

The design of scanner installations in individual institutions, their integration with other health care facilities, an understanding of the patient and information workflow and a proper regard for maintenance, safety and quality all affect CT scanner utility and long term operation.

(a) *Scanner installation and environmental conditions*: The installation process should ideally be under the direction of a project manager. Many support departments in a facility need to be considered and coordinated, including information technology, radiation safety, picture archive and communication systems (PACS) and radiology information systems (RIS) managers. The scanner location must be carefully considered with regard to the ease of patient transfer to the scanner. CT systems require stable and controlled environmental conditions, particularly in relation to the power supply[5], a dust free environment, air conditioning, appropriate lighting, appropriate reporting areas and adequate radiation shielding [55, 56]. Ancillary equipment such as computers and cooling equipment may also be required and their position determined by the requirements of the manufacturer. Other ancillary equipment such as contrast pumps, fluoroscopy systems, emergency cardiac resuscitation systems and life support systems are usually found in the scan room. Other items, such as CT scanner workstations [57] and PACS workstations (see below), hard copy devices and histology stations are located outside the scan room.

[5] In some cases, it may be necessary to upgrade the power supply or to use a power conditioner before installation.

(b) *Availability of support, consumables and maintenance*: A system should be chosen that can be adequately supported at the location of the institution. The performance of a CT system can only be realized consistently if the system is maintained properly and software is kept current by timely updates. While a manufacturer may provide service in a particular area, it may not always have enough adequately trained staff to support unique and advanced models. Routine service for a well utilized scanner can be quite frequent (every 2–6 months). Such maintenance is usually covered by a contract which may or may not include the cost of parts and repairs. The response time to a breakdown should also be included in the contract. Consumable costs should also be considered and these would include contrast agents, saline, cannulae, etc. Storage media may also be a consumable. If not mentioned in a maintenance agreement the X ray tube may also be considered a consumable. If maintenance is to be carried out 'in house', then the purchase agreement should include adequate training for staff.

(c) *Patient workflow*: The installation of a new CT scanner can be seen as an opportunity to change and increase patient throughput [58, 59]. Key elements involved in these changes include: assigning a scan protocol at the acceptance of the referral, scheduling of patients, patient preparation, scan set-up (which may come from the RIS or PACS) and image reconstruction.

(d) *Local environment for image reading and information workflow*: The use of soft copy reporting with appropriate display systems is strongly encouraged as part of an integrated information flow system. In fact, much of the functionality of an MDCT comes from the capability to review many images rapidly (see below). If circumstances make it essential that film images be made for diagnosis, it must be considered that there are significant difficulties with film processing to be overcome [60], as well as the consumable cost of film materials.

Modern MDCT scanners generate numerous images and special attention needs to be given to the configuration of the information flow, as outlined in Fig. 14. Immediate questions that need to be addressed include the location of the reporting areas, the identification of which images will be reported on, and what data will be saved and where.

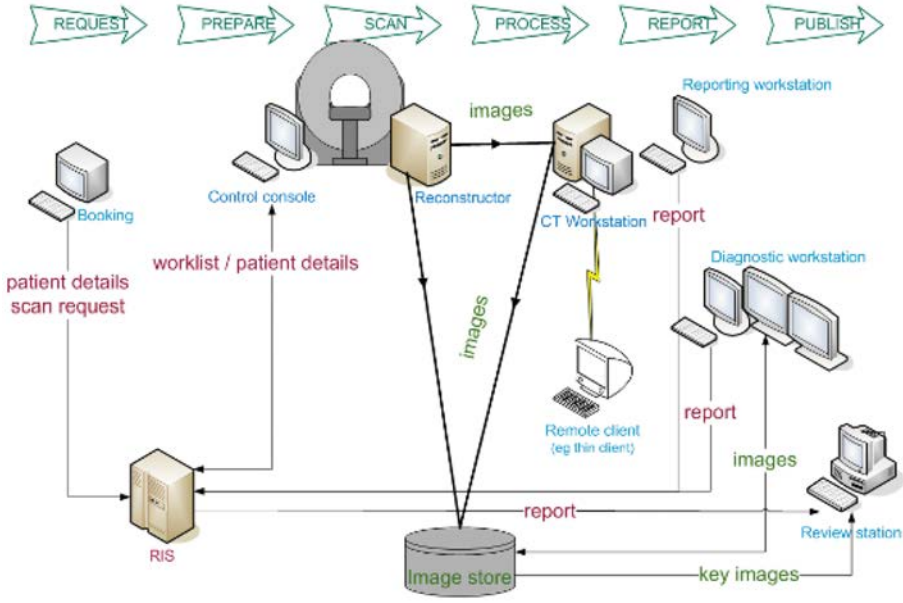


FIG. 14. An example of a simple option for information workflow (reproduced from Ref. [50] with permission).

The display system plays a major role in the overall performance of a CT scanner, both in terms of the ease of image interpretation and the image quality presented to the radiologist. For all diagnostic interpretation (reporting), the viewing conditions must be optimal, with low ambient lighting levels and no surfaces that allow light reflection.

For optimal information flow, it is recommended that reporting be done on a PACS or CT scanner workstation with integrated RIS capability [61]. The management of the generated CT data and the integration with the RIS and PACS systems are critical. A CT scanner can produce up to 12 GB of image data every day for storage. A policy for both short term and long term storage is needed [62] and the use of the PACS system (if available) is an important consideration. Alternatively, local backup storage may be considered. If images are to be sent to external clients, this may require a facility to burn data discs with automatic labelling if possible.

(e) *Integration of the CT scanner with other systems*: As mentioned above, the CT scanner will typically be integrated with other information systems within the department and hospital, such as the PACS, RIS and hospital information system and possibly with external facilities. Most CT scanners have the required capability for system integration on the console, as well as at specialist workstations. Image data exchange between systems has been standardized through compliance with DICOM standards. While the majority of scanners manufactured after 2000 should have suitable DICOM compliance, older models often do not have full DICOM capability and interfacing with some other products is not possible. Patient information, including the final report, is managed by HL7 systems [63]. An overarching framework is provided by Integrating the Healthcare Enterprise (IHE), which is a global initiative designed to advance the state of data integration in health care across all hospital systems and promote the coordinated use of established information standards, such as DICOM and HL7, through common functional profiles [64]. Specialist knowledge is required to ensure that data exchanges can be achieved between a CT installation and existing hospital systems.

Confidentiality and data security are important issues, particularly in a digital environment. It is the responsibility of all health professionals to ensure that identifiable patient data are kept confidential.

(f) *QC, safety and dose monitoring*: QA is covered elsewhere. Safety includes infection control within the scanner facility. Patient dose should be the responsibility of the medical physicist and can be inferred from phantom measurement and particularly from patient dose audits. Increasingly, the DICOM header is used to store important scan data that can include parameters used for patient dose (C_{VOL} ($CTDI_{VOL}$), P_{KL} (DLP) etc.). This feature needs to be considered if automatic patient dose assessment is to be contemplated.

4.1.3. Staff considerations

It is also essential to consider the staffing requirements for a CT scanner and the need for specialist training to read CT images and to operate and maintain the CT equipment. Well-trained and experienced personnel (radiologist, radiographer, medical physicist) are required. This will include training on the specific scanner, which is usually provided by the manufacturer as part of the purchase package. This training is usually conducted in three phases: prior to installation on a similar system located elsewhere, at installation and then a few

weeks after clinical services have begun. Staff with training in IT are essential [65]. For in-house servicing, training of the service engineers is also essential.

4.1.4. Cost considerations

A cost effectiveness analysis is essential before the purchase of a CT scanner. Cost does not necessarily correlate with equipment quality and clinical utility, therefore equipment selection purely on a cost basis should be avoided. Cost can be classified as capital costs, recurrent costs and hidden costs. A full estimate of total costs over the expected life of the scanner should be performed and included in the purchasing decision.

(a) *Capital costs*: These are costs incurred only at the time of purchase, although they may be paid over a number of years. Key components in these costs are the CT scanner, consisting of gantry, couch, console and computers; CT scanner workstations and installation costs. Other important costs include ancillary equipment, accessories, additional clinical application software, additional hardware and licences for remote access and staff training. The cost of disposal of existing equipment should also be considered.
(b) *Recurrent costs*: These costs include scanner service and maintenance, replacement X ray tube (if not included in the maintenance agreement), contrast injector pump and consumables, staff costs and energy costs.
(c) *Hidden costs*: These are sometimes overlooked in the project analysis. The following list of possible additional costs should be considered. These include the possible upgrading of the power supply for the scanner, including specialist requirements for power cables; building works, including radiation shielding and load bearing considerations; system integration with existing information systems (noted above); networking costs; PACS storage; increased cost of patient movement; additional patient preparation and waiting facilities; increased reporting facilities; desktop computer upgrades; scanner software upgrades and licences if not covered in maintenance contracts.

4.2. SPECIAL CONSIDERATIONS FOR RADIOTHERAPY CT SELECTION

In the late 1990s, the imaging equipment manufacturers began designing major devices (CT, magnetic resonance and PET scanners) specifically for radiation therapy or with radiation therapy needs in mind. This paradigm change

resulted in a variety of imaging devices available for radiation therapy simulation, ranging from conventional single slice CT scanners to large bore multidetector scanners [66] with specialized radiotherapy software.

As discussed in Sections 2.2 and 3.1.2, the CT scanner used for radiotherapy should include a laser–patient positioning/marking system, a flat couch top and possibly virtual simulation/3-D treatment planning software (Fig. 7(b)) and 4-D capacity. Considerations for radiotherapy CT scanner features should include: bore (gantry opening) size, tube heat capacity, couch top, patient marking lasers, CT simulation software, contouring and localization of structures, image processing and display, and simulator geometry.

(a) *Bore size*: Conventional CT scanners typically have a 70 cm bore. For the vast majority of procedures, a 70 cm bore size which typically allows objects with a largest dimension of 50 cm is sufficient. However, one of the requirements in the treatment of several cancer sites (breast, lung, vulva, for example) is for extremities to be positioned away from the torso. Another case is that of head and neck patients scanned in immobilization devices. When acquiring a CT scan with a patient in such treatment positions, extremities often cannot fit through a conventional 70 cm diameter scanner bore opening. In such situations, patient positioning needs to be modified to acquire the scan. This can result in a less than optimal treatment position (the patient may be less comfortable and therefore the daily set-up reproducibility may be compromised) or in a mismatch between the imaging and treatment positions. For treatment planning purposes, it is desirable to have the full extent of the patient's skin on the CT image. Lateral patient separation can often be larger than 50 cm. Large bore CT scanners (>70 cm diameter bores) were specifically designed with radiation therapy needs in mind and have increased the available scan field of view (SFOV). However, at extended SFOV, the validity of quantitative CT values comes into question. CT numbers for some scanners are accurate only for smaller SFOVs and the values towards the periphery of large SFOV images are not reliable. This can be a concern for some dose calculation algorithms because inaccurate CT numbers can lead to dose calculation errors. The impact of CT number accuracy for increased SFOV images on dose calculation accuracy should be evaluated during scanner commissioning [67].

(b) *Tube heat capacity*: Compared with diagnostic scanning, radiotherapy image acquisition often requires longer scan volumes, with thinner slices (preferably 3 mm), which should be acquired in the shortest possible time to minimize spatial deformations due to patient motion. Acquiring a large number of images in a short time is often beyond the heat storage capacity

of single slice CT scanner X ray tubes. Often, fewer images are taken, slice thickness is increased, the image quality is decreased (reduced mAs), or scan pitch is increased to reduce the amount of heat produced during the scan and to allow for the entire scan to be acquired in a single acquisition. Greater X ray tube heat storage capacity can minimize the need for scan parameter compromises.

(c) *CT couch top*: The couch tops used for patient support in radiation therapy during both imaging and treatment should facilitate easy, efficient, reproducible and accurate patient positioning. It is not only important that a couch top improve patient positioning on a single device (i.e. treatment machine), but also that the repositioning of a patient from one imaging or treatment device to another be considered. A great improvement in this process is effected when all couch tops involved in patient simulation and treatment have a common design. They do not have to be identical, but they should have the same dimensions (primarily width), flex and sag under patient weight, and they should allow registration (indexing) of patient immobilization devices to the couch top. Figure 15 illustrates this concept. The CT scanner couch top has the same width as the linear accelerator used for patient treatment and both allow registration of the patient immobilization system to the treatment couch. The capability to register the immobilization device and the patient to a treatment couch can improve

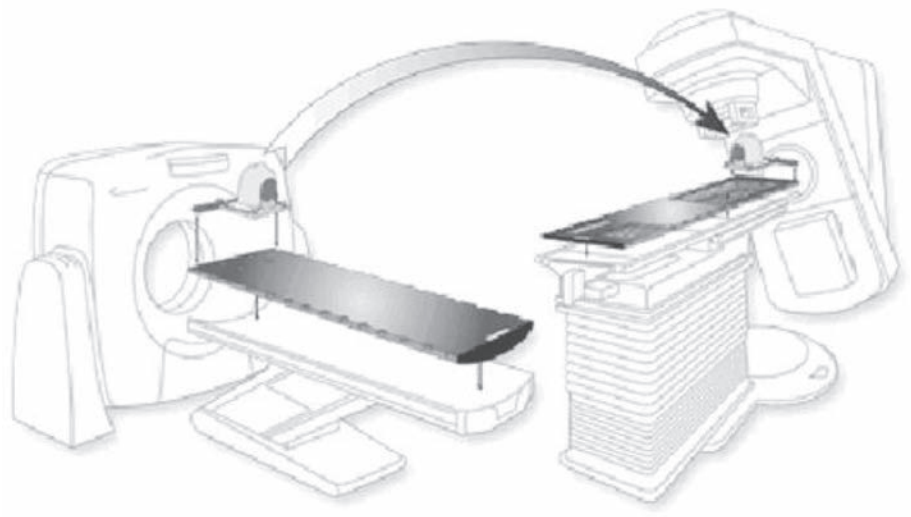


FIG. 15. Transfer of the patient from CT scanner to the treatment machine can be improved by ensuring compatibility between the CT scanner and linear accelerator couch tops (courtesy: CIVCO Medical, Inc.).

immobilization, set-up reproducibility, accuracy and efficiency. The patient is always positioned in the same place on the treatment machine and patient daily set-up can be facilitated using the treatment couch positions.

(d) *Patient marking lasers*: A laser system is necessary to provide reference marks on patient skin or on the immobilization device. Figure 7 shows a laser system for a CT scanner used for radiotherapy imaging. Preferably, at least the overhead (sagittal) laser should be movable. However, this is not imperative and the alignment accuracy and the stability of the lasers are much more important. It is desirable for the lasers to have easy alignment adjustability and to have positional stability with time. Laser projection should also be at least 50 cm long at the tomographic plane.

(e) *CT simulation software (option)*: CT (virtual) simulation software is a package that may be available with some CT scanners and is used as an interface between the CT scanner and the radiotherapy treatment planning software (Fig. 16). This software is intended to simplify CT scanning for radiotherapy planning and also to increase the efficiency of this process. As with all software programs, user-friendly, fast and well-functioning CT (virtual) simulation software with useful features and tools will be a determining factor for the success of a virtual simulation program. Commercially available programs far surpass in-house written software and are the most efficient approach to virtual simulation. Several features are very important when considering virtual simulation/3-D treatment planning software; in some instances, the virtual simulation software is built into the radiation treatment planning computer system.

(f) *Contouring and localization of structures*: Contouring and localization of structures is often cited as one of the most time consuming tasks in the treatment planning process. The virtual simulation software should allow fast user-friendly contouring with the help of semi-automatic or automatic

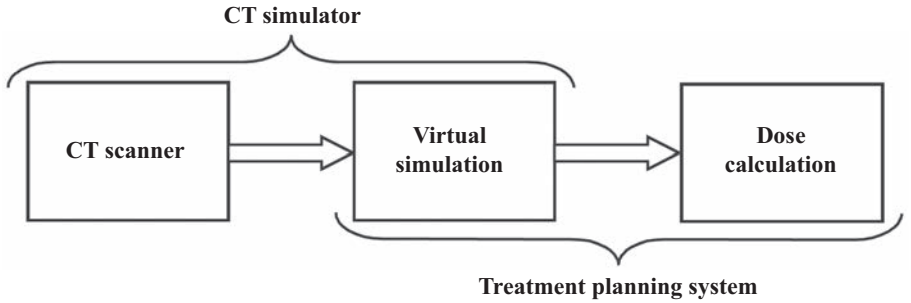


FIG. 16. Simplified diagram showing relationship of CT simulation software, CT scanner and radiotherapy treatment planning (adapted from Ref. [2]).

contouring tools. An array of editing tools (erase, rotate, translate, stretch, undo) should be available. The capability to add margins in three dimensions and to draw treatment portals automatically around target volumes should be available. An emphasis should be placed on functionality and efficiency.

(g) *Image processing and display*: The virtual simulation workstation must be capable of processing large volumetric sets of images and displaying them in different views as quickly as possible (near real time image manipulation and display are desired). The quality of reconstructed images is just as important as the quality of the original study set. The reconstructed images can be either digitally reconstructed radiographs [17, 18] from a 3-D data set or multiplanar reconstructions and these are used for target volume definition and treatment verification having a direct impact on the accuracy of patient treatments.

(h) *Simulator geometry*: A prerequisite of CT simulation software is the capability to mimic the functions of a conventional simulator and of a medical linear accelerator. The software has to be able to show gantry, couch, collimator and jaw motion, source–surface distance changes, beam divergence, etc. The software should facilitate design of treatment portals with blocks and multileaf collimators.

5. BASIC PRINCIPLES OF QA IN CT

5.1. QA ACTIVITIES

A QA programme in diagnostic radiology, as defined by the World Health Organization [68], is an organized effort by the staff operating a facility to ensure that the diagnostic images produced are of a sufficiently high quality that they consistently provide adequate diagnostic information at the lowest possible cost and with the least possible exposure of the patient to radiation. Registrants and licensees must establish a comprehensive QA programme for medical diagnosis with the participation of appropriate medical physicists, taking into account the principles established by the WHO [68].

QA programmes for medical exposures should include:

(a) Measurement of the physical parameters of the radiation generators and imaging devices at the time of commissioning and periodically thereafter.

(b) Verification of the appropriate physical and clinical factors used in patient diagnosis (or treatment).
(c) Written records of relevant procedures and results. This includes a manual that defines clear lines of responsibility, outlines the individual QC tests performed, stipulates the test frequencies, aids staff training, facilitates audit of a service and helps to keep information within the service.
(d) Verification of the appropriate calibration and conditions of operation of dosimetry and monitoring equipment.
(e) Optimization of clinical protocols and equipment operation to achieve the aims of QA as stated above.
(f) Regular and independent quality audit reviews of the QA programme.

QA programmes are designed to ensure that the radiology equipment and staff procedures can yield the desired information. They include:

(a) Administrative procedures or management actions designed to verify that:
— The QC tests are performed properly and according to a planned timetable;
— The results of these tests are evaluated promptly and accurately;
— The necessary corrective measures are taken in response to these results;
— The assignment of responsibility for QA actions is made;
— Standards of quality for equipment in the facility are established;
— Adequate training is provided;
— The appropriate equipment for each examination is selected, including the writing of adequate equipment specifications.
(b) Acceptance testing and commissioning (see Fig. 17):
— Acceptance tests are those performed to verify that the purchase specifications have been met by the vendor. These are often performed by the company installing the equipment under the supervision of the medical physicist [69] or, alternatively, performed independently by the medical physicist. Commissioning tests are those undertaken at the time the equipment is put into service and are used to establish baseline levels of performance, including measurements which may be helpful in optimization of protocols (Section 6). These are performed by the medical physicist. To a large extent, acceptance and commissioning tests overlap. This publication primarily describes the tests that form a comprehensive ongoing QC programme for CT, but it is recognized that it is necessary to ensure that the equipment, as delivered, conforms to specified standards and that appropriate initial baseline values are established and used to ensure the maintenance of the quality of the equipment throughout its working life. These acceptance and

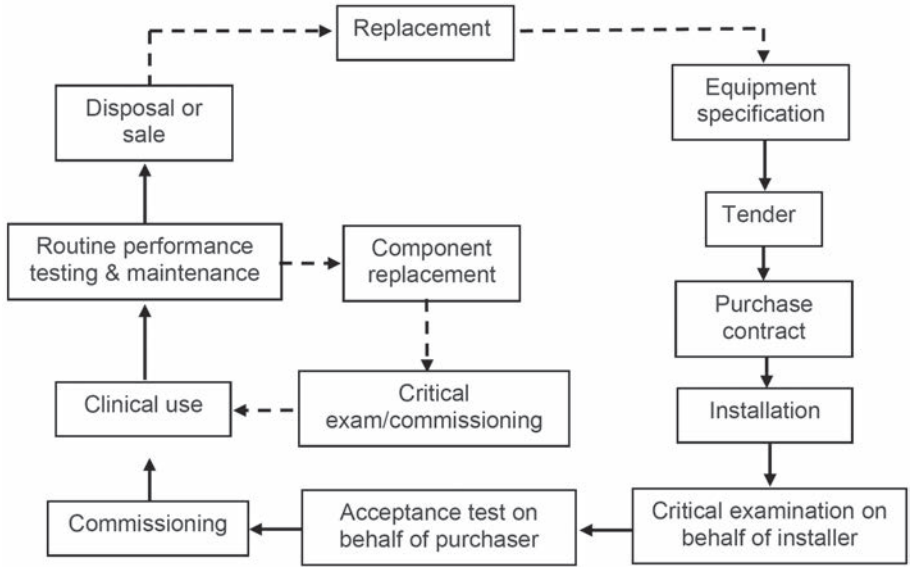


FIG. 17. Life cycle of a piece of equipment.

commissioning tests are included in this publication and are indicated as such. During acceptance testing, a qualified person should check the electrical and mechanical safety of any new installation.

(c) QC tests (also classified as either constancy or status tests by the IEC) are used to test the components of the radiological system and to verify that the equipment is operating satisfactorily.

(d) Verification of QC equipment and material.

(e) Follow-up of any corrective actions proposed:

—It is important that routine QC testing be properly performed in the CT facility and that results be documented thoroughly and carefully. It is equally important that problems and potential problems be clearly documented and communicated to the facility in a timely manner and that the medical physicist be assured that the receiving party has received and understood the supplied information. This is especially the case when safety concerns are raised.

—The reporting structure in the facility should be understood by the medical physicist, who should ideally report problems to an individual who is empowered to call in service personnel and, if necessary, who can ensure that the equipment is not used until the problems are corrected. The medical physicist may be asked to explain the problems to service personnel and to share test results with them. The medical physicist and

the representative from the facility should work together to ensure that the problems have been appropriately corrected.

(f) Education and training of the staff, including radiologist, radiographer and medical physicist. Each must meet a minimum level of competency.

(g) Continuing education. Each team member must undertake sufficient continuing education to ensure that they are up-to-date on new techniques and that they are 'refreshed' relative to their basic knowledge, e.g. radiation safety.

(h) Experience. To ensure proficiency, the radiologist must read a sufficient number of cases, the radiographer must undertake a minimum number of cases a year and the medical physicist must conduct a sufficient number of CT unit surveys under American College of Radiology guidelines [70].

5.2. ROLES AND RESPONSIBILITIES

5.2.1. The licensee or registrant

The licensee or registrant has specific responsibilities to ensure that all regulatory and/or licensing requirements are met. Further, the licensee or registrant must ensure that all radiologists, radiographers, medical physicists and other personnel who work at the facility are appropriately qualified and trained and meet all continuing education and experience requirements.

It is the responsibility of the licensee or registrant to ensure that a QA programme is in place that encompasses all aspects of the imaging process. The specific tasks within that programme may be delegated to appropriate staff who already have expertise in carrying out those tasks. Notwithstanding the above delegation of authority, it remains the ultimate responsibility of the licensee or registrant that the elements of the QA programme are fulfilled.

A radiologist should be identified by the facility to have the specific responsibility of ensuring that all required QA activities are performed.

5.2.2. The radiologist

Although it is recognized that the radiologist will delegate many of the following tasks, they still have responsibility for the following tasks:

(a) Ensuring that medical physicists and radiographers have adequate training and continuous education courses in CT;

(b) Ensuring that all equipment is appropriately maintained;

(c) Motivating, supervising and managing all aspects related to the QA programme in the area of CT;
(d) Providing an orientation programme for radiographers based on a carefully established procedures manual;
(e) Designating a single radiographer to be the primary QC radiographer to perform the prescribed QC tests and oversee those that have been delegated to other individuals;
(f) Ensuring availability of the equipment and the necessary materials for the implementation of the QC tests;
(g) Arranging staffing and scheduling so that adequate time is available to carry out the QC tests and to record and interpret the results;
(h) Ensuring that a medical physicist is available to oversee the equipment related QC programme and to perform the medical physicist's tests;
(i) Reviewing the radiographer's test results at least every 3 months, or more frequently if consistency has not yet been achieved, and reviewing the medical physicist's test results annually, or more frequently as needed;
(j) Designating an individual to oversee the radiation protection programme for employees, patients and other individuals in the surrounding area;
(k) Ensuring that records of employee qualifications, mammography technique and procedures, infection control procedures, QC, safety and protection are properly maintained and updated in the CT QC procedures manual;
(l) Providing feedback continually, both positive and negative, to radiographers on image quality and QC procedures.

5.2.3. The radiographer

The responsibilities of the radiographer include:

(a) Ensuring that the appropriate protocol and technique factors are used for the requested examination.
(b) Ensuring that the QC tests are performed, interpreted and recorded appropriately. This is best achieved when one radiographer assumes overall responsibility for QC matters and is able to train others to assist in QC activities.
(c) Recording imaging problems.
(d) Undertaking additional continuous education courses.

5.2.4. The medical physicist

The medical physicist is a person trained in medical physics and certified as a medical physicist according to the applicable programme in the State, if such a

programme exists. The medical physicist should be specialized in diagnostic radiology.

The responsibilities of the medical physicist include:

(a) Advising the facility on CT image quality and on radiation protection of the patient, staff and members of the public.
(b) Advising the facility on acquisition, installation and shielding for CT.
(c) Conducting tests to ensure the safety and proper performance of equipment used in CT. These include acceptance, commissioning and routine QC tests.
(d) Advising the radiologist and radiographer on optimization (see Section 6).
(e) Providing oversight and advice to the radiographer who carries out the radiographer's component of the QC programme.

6. OPTIMIZATION OF CLINICAL PRACTICE

6.1. INTRODUCTION

6.1.1. General principles of radiation protection

All potential diagnostic and interventional radiology exposures must be subject to the principles of justification and optimization which are common to all practices dealing with exposures to ionizing radiation [71]. This may be stated as follows [72]:

> "The principal aim of medical exposures is to do more good than harm to the patient, subsidiary account being taken of the radiation detriment from the exposure of the radiological staff and of other individuals."

For patients undergoing medical diagnosis or treatment, there are two levels of justification. Firstly, the practice involving exposure to radiation must be justified in principle through the endorsement of relevant professional societies, as matters of effective medical practice will be central to this judgement [73, 74]. Secondly, each procedure should be subject to a further, case-by-case justification by both the referrer, who is responsible for the management of the individual patient and who determines that the exposure is necessary for diagnostic purposes, and the radiologist or other practitioner who may direct the radiological procedure [75].

6.1.2. The need for optimization in CT

Technical and clinical developments in CT have not, in general, led to a reduction in patient dose per examination. However, the growth in CT examinations and the corresponding increase in dose to the population is well documented[6] [81]. This has led to increased concerns about the magnitude of these doses and the potential risks that these imply. The latest mortality data on the Japanese atomic bomb survivors [82] are consistent with there being a risk of cancer induction at dose levels typical for CT examinations. This is particularly true of CT examinations of paediatric patients, who may also be at greater risk from stochastic effects than the general population. Further, repetitive CT examinations (e.g. multiphase contrast studies) have the potential to give absorbed doses to tissues that may approach or even exceed the threshold for deterministic effects.

6.1.3. General principles of optimization

Once clinically justified, each examination should be conducted so that the dose to the patient is the lowest necessary to achieve the clinical aim. The optimization process necessarily requires a balance between patient dose and image quality along with other clinical considerations, including the use of a contrast agent. Dose reductions must not be achieved without regard to any loss of diagnostic quality in the image that may accompany the dose reduction, as previously noted by the IAEA [74]:

> "The objective of the diagnostic radiology process as a whole is to obtain the requested diagnostic information with the minimum patient exposure within prevailing resource limitations."

The requirement for image quality should be tailored to the clinical problem and lower levels may be acceptable in some circumstances. Certainly, the size and shape of the patient will influence the level of dose required [83]. Accordingly, the equipment operators should minimize patient dose while

[6] Further, the growth of CT is so fast that around 75 million CT examinations are performed annually in the United States of America out of a total population of around 300 million people. The National Council on Radiation Protection in the USA has estimated [76] that the contribution of medical radiation to the population collective dose has increased from 15% in 1980 to 53% in 2006, with CT accounting for the major share of this increase at approximately 1.5 mSv per capita per year. Similar increases have been noted in the United Kingdom [77] and this trend is common to most developed countries [78–80].

maintaining acceptable image quality for the diagnostic information required. It must be recognized that optimization is a multidisciplinary task involving the radiologist, radiographer and medical physicist.

Dose surveys for CT procedures indicate wide variations in patient dose and indicate the need for optimization with the use of established diagnostic reference levels (DRLs) to aid in this process.

6.1.4. General recommendations for good practice in CT

Recommendations concerning achievable standards of good practice in CT have been developed by the European Commission in the form of quality criteria [84, 85]. These provide an operational framework for radiological protection initiatives in which technical parameters for image quality are considered in relation to patient dose. Diagnostic and dose requirements for CT are specified in terms of the quality criteria considered necessary to produce images of standard quality for a particular anatomical region. The subjective image criteria include those criteria that relate to the visualization or critical reproduction of anatomical features. Criteria concerning patient dose are given in terms of DRL associated with the examination technique used for standard sized patients. Quality criteria have been developed for most CT examinations, together with examples of technique parameters influencing the dose.

6.2. FACTORS AFFECTING IMAGE QUALITY AND DOSE

6.2.1. Introduction

A large number of factors in a CT examination can affect the dose and image quality, collectively described by the image noise and spatial resolution. These factors can be generally categorized as (i) radiographic protocol or scan parameter, (ii) equipment, (iii) image reading condition and (iv) patient related factors. Patient related factors can usually be controlled through adjustments in the scan parameters, therefore leaving three main groups of factors. Of these, the equipment factors are often set by the scanner configuration at manufacture and are beyond the control of the operator.

Controllable factors affecting patient dose in CT include:

- Radiographic protocol or scan parameters;
- kV and mAs for manual operation;
- Pitch;
- Reconstruction filter;

- Scan length and number of scan series;
- Patient size variation, usually requiring changes in examination protocol;
- AEC (correct dose modulation techniques);
- Collimation selection including MDCT considerations of over-beaming;
- Helical acquisition considerations of over-ranging;
- Scan mode (axial, helical or MDCT);
- Use of external filters for body part shielding.

6.2.2. Scan parameters

The relationship between scan parameters, image quality and dose is complex. To some extent this depends on whether the scanner is single slice or multislice and whether scanning is axial ('step and shoot') or helical. Table 2

TABLE 2. EFFECT OF SCAN PARAMETERS ON IMAGE QUALITY AND DOSE[a]

Parameter	Noise	Spatial resolution		Dose
		Imaged slice width	Tomographic plane	
Current (mA)	✓			✓
Rotation time[b,c]	✓		✓	✓
Voltage (kV)	✓			✓
Focal spot selection[c,d]	(✓)	✓	✓	(✓)
Nominal imaged slice width	✓	✓		
Pitch	✓	✓		✓
Total X ray beam collimation	(✓)[e]	(✓)[e]		✓
Detector group width	✓	✓		
Scan volume				✓
x–y reconstruction kernel	✓		✓	
z axis interpolation algorithm	✓	✓		

[a] When an individual parameter is varied while all other scan factors are kept constant.
[b] If the sampling frequency is dependent on rotation time, then spatial resolution will be affected.
[c] The focal spot size has a direct impact on spatial resolution, through penumbral effects.
[d] Small amount of change to noise and dose with focal spot size.
[e] Single slice, dual slice.

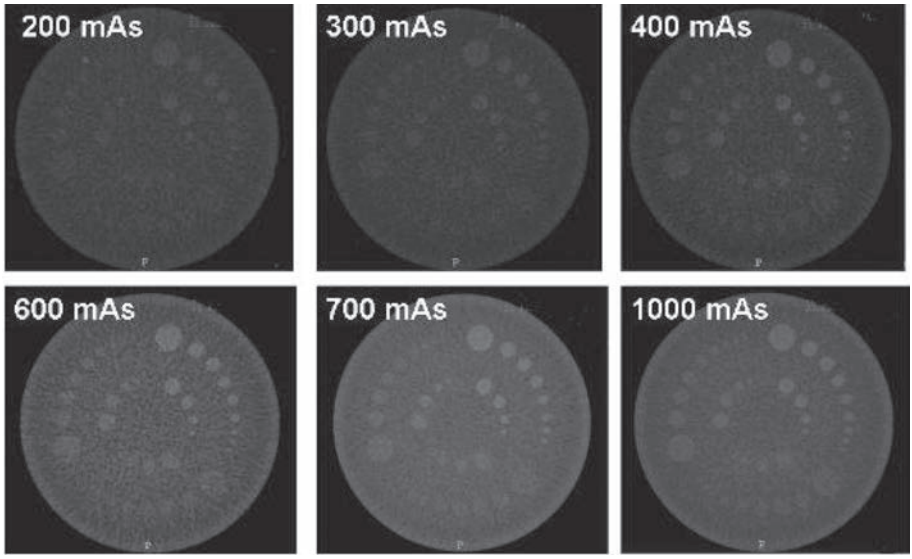


FIG. 18. The effect of mAs on noise and the consequent effect of low contrast visibility.

gives a summary of the effect of changing a scan parameter on image quality and dose, while keeping other parameters constant.

Noise, dose and mAs

The image noise is expected to follow the relationship

$$\sigma^2 \propto \frac{1}{N} \tag{12}$$

where N equals the number of photons contributing to the image.

In practical terms this means that image noise will vary inversely as the square root of the mAs as illustrated in Fig. 18. For example, the doubling of the mAs will result in an increase in the dose of 100% combined with a decrease in the image noise by a more modest 41%.

Dose, axial resolution and pitch

The dose will also vary inversely with the scan pitch[7]. It is, therefore, good practice to select the highest pitch value and the lowest mAs consistent with the required clinical diagnosis. It is usual to select a pitch in the range of one to two.[8] However, some CT scanners have the selection of mAs and pitch tied together so that the ratio of mAs to pitch remains constant when the pitch is altered. Under these circumstances, changing the pitch (alone) has minimal impact on patient dose and pitch values of less than one may be preferred and safely used. The use of helical scanning, implied by the use of pitch, means that over-ranging will occur, which increases dose. This effect increases with pitch and most significantly with detector array size (see MDCT optimization discussed in Section 6.2.4). One disadvantage of using high pitch values with a single slice scanner is that the imaged slice width (axial resolution) can increase significantly. This is generally not the case with MDCT scanners using the z axis interpolation algorithm, except for reconstructions at the thinnest slice widths possible for the detector selected.

Noise, dose and kV

Until recently, it was assumed that the optimal tube energy for CT scanner operation was 120 kV, with increases up to 140 kV for larger patients. Nowadays, it is acknowledged that when intrinsically high contrast objects are required for diagnosis, larger noise values can be tolerated without significant loss of diagnostic ability. This is seen, for example, when calcium or iodine contrast agent is to be visualized against soft tissue.

Recent computer simulations and phantom measurements of CNR and CNR divided by the square root of dose [86] demonstrate that if the contrast is formed by objects of similar composition but of differing density, then the optimum tube potential is indeed about 120 kV. However, when high atomic number materials are used, particularly in the case of iodine, there are significant dose advantages in the use of lower kV values. Consequently, CT scanners should be calibrated over the full range of tube voltage values and the use of lower kV values should be investigated for the use of iodine by the optimization team.

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[7] See the Glossary for the definition of pitch. Note that for some MDCT scanners, manufacturers define pitch differently to this by multiplying the 'actual pitch' by the number of detectors used in the acquisition. In order to determine dose values, the glossary definition of pitch should be used.

[8] Only in exceptional circumstances should a pitch below unity be used with older single and dual slice scanners, since this is analogous to overlapping scanning in axial mode.

6.2.3. Optimization with AEC

The use of mA modulation technology [1, 13, 87], usually referred to as AEC or automatic dose control, is designed to: (i) reduce patient dose compared with the use of a fixed mA technique and/or (ii) improve image noise by reducing 'photon starvation' associated with regions of large variation in attenuation, such as the shoulders. Typically, AEC techniques utilize the SPR views used to determine attenuation profiles and then modify the mA using either the z axis (longitudinal) dose modulation, x–y axis (angular) dose modulation or both of these simultaneously within a scan period.

The use of AEC can lead to significant dose reductions, for example, Hundt et al. [88] have shown in a sample of patients undergoing abdominal and chest examinations that the introduction of X ray tube current modulation technology alone achieved reductions in dose of close to 30%. Note, however, that the indiscriminate use of AEC does not guarantee dose reductions and that the use of AEC should be based on the required clinical image quality. Heggie [89] has shown that for one MDCT scanner equipped with AEC without optimization, the doses were typically 35% higher than those for the same procedure performed on a single slice scanner. After optimization, dose reductions of between 14% and 58% were achieved so that for most procedures the doses were comparable with the original single slice scanner results.

The decision about acceptable image quality is usually made on the basis of achieving an acceptable level of image noise and ultimately should be made by an experienced radiologist. Once this noise level has been established and adopted into pre-programmed protocols it is important for radiographers to adhere to these protocols unless clinical indications dictate otherwise.

6.2.4. Optimizing MDCT protocols

MDCT scanners offer a number of clinical advantages, but because of a combination of unique design characteristics and superior scanning speed they are particularly prone to delivering high patient doses [1, 78, 80, 84, 85, 90] unless technical factors are carefully optimized by the user. Radiographers should be mindful that the default protocols provided by manufacturers of MDCT scanners can often be modified to achieve substantial dose reductions without loss of diagnostic integrity by 'tailoring' the technical parameters used in an examination [89–93] as illustrated in Fig. 19. The following points may be useful in MDCT optimization:

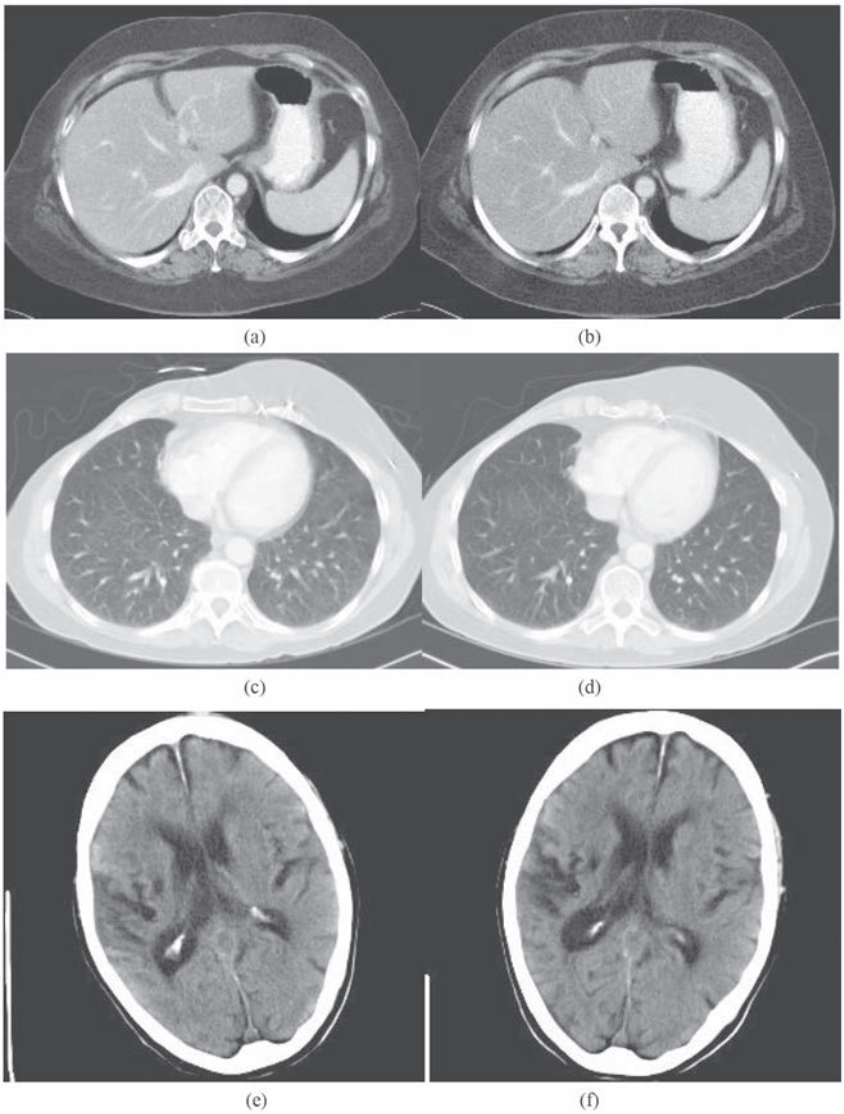

FIG. 19. Scans of three patients taken before and after optimization using the same reconstruction algorithm, voltage (kV) and slice width, but with different exposures. Slices have been selected to be matched anatomically to the extent feasible: (a) abdomen–pelvis scan with $CTDI_{VOL}$ = 11.3 mGy, DLP = 561 mGy·cm; (b) abdomen–pelvis scan with $CTDI_{VOL}$ = 6.8 mGy, DLP = 349 mGy·cm; (c) chest scan with $CTDI_{VOL}$ = 6.4 mGy, DLP = 214 mGy·cm; (d) chest scan with $CTDI_{VOL}$ = 5.1 mGy, DLP = 171 mGy·cm; (e) brain scan with $CTDI_{VOL}$ = 70 mGy, DLP = 960 mGy·cm; (f) brain scan with $CTDI_{VOL}$ = 40 mGy, DLP = 580 mGy·cm. Whilst the optimized images (b), (d) and (f) have more noise than images (a), (c) and (e), the former still remain diagnostic.

(a) Optimal scanning block size:
   — *Concern for unnecessary dose to sensitive tissues*: Restrict the scanned volume to the minimum necessary for diagnosis and scan in one large block rather than in multiple smaller contiguous blocks. This will minimize the impact of over-ranging. This is of particular concern for the lower abdomen/pelvis, to avoid irradiating the male gonads, and for the head, to avoid the lens of the eye.
   — *Concern to tailor the beam parameters to tissue region requirements*: Image noise may be tolerated more readily in one part of the body (e.g. the chest) than in another. Consequently, for these body parts the use of separate scan blocks might outweigh any dose penalty from over-ranging.
   — *Head and body filters*: Many scanners are designed to operate with different beam filtration for the head and abdomen, and as the filtration cannot be altered in mid-scan, two scan blocks are needed to utilize the optimum filtration conditions[9].

(b) Optimal collimation setting requires use of the widest beam collimation consistent with clinical requirements (e.g. 16 mm × 1.5 mm rather than 16 mm × 0.75 mm). This will reduce the effect of over-beaming. However, care is needed as it is usually possible to reconstruct a new set of images at a wider image slice thickness without having to make another X ray acquisition. However, it is not possible to reconstruct a thinner imaged slice width, except with a new X ray acquisition.

(c) Use of axial, as opposed to helical, techniques for routine head scans, unless clinical indications suggest otherwise. As noted previously, over-ranging will inevitably lead to irradiation of the lens of the eye if the helical mode is employed.

(d) Use of anatomy dependent, attenuation based methods of X ray tube current modulation (AEC) with an appropriately selected reference mAs or noise index. For optimum performance with this technology, the SPR must be acquired over the full length of the patient that is of clinical relevance using the same kV that will subsequently be utilized for the volume acquisition.

(e) If AEC is not available, it is recommended that technique charts be posted for the common CT examinations to assist in maintaining proper image quality.

[9] For example, the radiation output per mAs is typically 25% and may be as much as 40% higher in head as opposed to body mode [91].

(f) The use of multiphase examinations should be minimized. This is a fundamental decision relating to clinical justification and optimization.

It should be noted that even when optimized, scan protocols applicable to one scanner should not be transferred to another scanner without consideration of the differences in X ray tube filtration and geometry. Even scanners from the same manufacturer may differ significantly in X ray beam filtration and detector type (efficiency) and geometry.

6.2.5. Use of body part shielding

Attempts at shielding relatively radiosensitive tissues such as the breast and eyes have been discussed in the literature. During CT scanning of the chest and upper abdomen, substantial breast dose reductions have been demonstrated without compromising diagnostic image quality, by using breast shields made of high Z materials, such as lead or bismuth, raised above the surface of the chest [94–96]. Similarly, shields may also be useful in minimizing the dose to the lens of the eye during head CT examination [97], although other investigators have shown that appropriate tilting of the gantry offers better dose reduction possibilities.

It should be noted that with the use of tube current modulation technology (Section 6.2.3), body part shielding is not recommended.

6.3. THE OPTIMIZATION PROCESS

To optimize a CT examination, a process involving 14 steps can be followed:

Initial preparation

(1) Establish agreement for an optimization process with the radiology department, including a schedule of achievable targets.
(2) Determine the priority for examinations to be optimized for a particular modality in conjunction with clinicians and radiographers, considering such factors as examination risk and frequency.
(3) Check QA status of equipment used for the procedure.
(4) Establish clinically appropriate image quality requirements in collaboration with clinicians.

Dose, image quality and clinical acceptability assessment

(5) Determine patient doses (preferably from a patient audit or possibly from phantom based measurements (see Section 9.6)).
(6) Determine image quality (preferably with the assistance of radiologists; the assessment should include a measure of noise and resolution).
(7) Review scan protocol, examining the purpose of the examination and the adequacy of technical factors to account for patient size, with special consideration being given to any paediatric protocols.

Review of current status of procedure

(8) Compare examination dose with appropriate benchmarks, such as the DRL, as available [98].
(9) Compare examination image quality with appropriate benchmarks, if available.
(10) In conjunction with the radiologist and radiographer, review examination related data including:
—Radiographic protocol;
—Equipment configuration;
—Image reading conditions.
(11) Investigate the effect on image quality and dose of varying the parameters in the above list.

Intervention

(12) Recommend changes to the radiographic protocol, equipment configuration and/or viewing conditions, on the basis of the review of the procedure (see above).

Verify effect of optimization process

(13) After an agreed period of clinical introduction, repeat the dose and image quality analysis to determine the effectiveness of the optimization intervention.
(14) Record the results of the optimization procedure in such a way that they are accessible to all interested parties, particularly radiographers and clinicians.

6.4. CLINICAL EXAMPLES OF OPTIMIZATION

6.4.1. Chest CT

Image noise is a key indicator of image quality for CT and is related to the attenuation of the X ray beam as it traverses the body, just as in plain radiography. The lower attenuation of the chest compared with the abdomen or pelvis results in lower image noise for the same mAs. More importantly, since subject contrast in the chest is relatively high, a modest increase in noise will have minimal impact on the diagnostic content of the images. Therefore, when compared with an abdomen or pelvis examination, a chest examination involves a significantly lower radiation dose to achieve an acceptable image quality.

Many studies [99–105] have shown that, depending upon the clinical question being asked, significant dose reductions are possible in chest CT. The use of lower kV settings (for example, 80 kV compared with the commonly used 120 kV) has also been advocated for CT pulmonary embolism angiography, where a reduced radiation dose combined with increased image contrast has been documented [106]. Kalra et al. [107] have also demonstrated that the application of specialized noise reduction filters was successful in reducing image noise for low radiation dose chest CT images, allowing for some compromise in image sharpness.

6.4.2. CT perfusion

CT perfusion studies (continuous or repeated scanning to follow the time course of the injected contrast agent) should be avoided if possible. If they are undertaken, it is necessary to ensure that an established, optimized protocol is followed. In particular, the scan duration should be limited to the absolute minimum required, as hair loss has been known to occur in patients undergoing this procedure [108].

6.4.3. Other emerging CT procedures

There are other novel CT applications ranging from CT fluoroscopy to functional and 4-D CT. The latter can use a gating signal to select, retrospectively, CT projections belonging to a particular phase of, for example, the breathing or cardiac cycle. In order to have a sufficient number of projections in all phases, these scans typically use significant oversampling, for example, by using a very small pitch in helical CT. The resulting scan times are usually many times longer than those recorded in conventional CT scanning and could, therefore, also result in much larger doses received by the patient if the tube current is not reduced.

Investigations [109] into the combined effect of lowering the kilovoltage setting (80 kV) and using an AEC technique (ECG pulsed tube current modulation) for coronary CT demonstrated that radiation dose exposure can be reduced by up to 88% for slim patients without impairing image quality. The operator, however, should also be aware that AEC may not always be available with these modern CT techniques.

6.5. OPTIMIZATION FOR CHILDREN[10]

Prior to 2001, the vast majority of CT imaging of children was conducted using the same or similar techniques as those used for adult imaging. In 2001, several articles [82, 110, 111] received considerable media attention by pointing out that this approach was not necessary and resulted in estimated radiation doses to the smallest children as much as three times those given to an adult. Since then, considerable work has been published in the literature on protocols to reduce the dose to children undergoing CT examination [112–123]. However, many of these protocols are scanner specific and not transferable to other CT units.

The instructions below provide guidance in either developing CT protocols for children or verifying that current protocols are appropriate. It should be stressed that this publication is only intended as a guide and the technique parameters provided are only suggestions that primarily apply to the use of manual techniques. The interpreting radiologist, in consultation with a medical physicist, must evaluate any changes to the practice's techniques that reduce radiation dose so that the adequate diagnostic information is available.

Technique reduction factors were developed from radiation measurements obtained with ionization chambers and anatomical paediatric CT phantoms [124]. These phantoms range in size from infant to large adult and consist of tissue equivalent plastic that has a CT number of '0'. The abdomen phantoms have a tissue equivalent spine; the thorax phantoms have a tissue equivalent spine and lung tissues and the head phantoms have a tissue equivalent skull bone.

In order to use these reduction factors, the radiologist should first work with the CT radiographer to gain familiarity with techniques used for both adults and children. It must then be verified that the adult technique factors do not deliver estimated radiation doses *larger* than those recommended by national or international DRL values[11]. No universal CT technique can be used with all

[10] This material has been adopted with permission from the Image Gently Alliance.

[11] Values have also been established by the American College of Radiology's (ACR) CT Accreditation Program [125, 126].

vendors' CT equipment for the adult patient. Differences in CT scanner design (e.g. bow tie filters, focal spot–detector distance, detector efficiency) make it impossible to estimate patient radiation dose based on technique factors alone. Consequently, a medical physicist specializing in diagnostic radiology must measure the radiation output from the CT scanner in order to estimate the dose and assist in establishing appropriate adult abdomen and head techniques.

The adult abdomen and head techniques will become the baseline techniques. Using these baselines, Tables 3 and 4 allow estimation of the appropriate reductions in mAs for children based on their age or PA thickness. Ideally, the PA thickness of the paediatric patient should be measured with calipers. If the child's PA thickness cannot be obtained, Tables 3 and 4 list an 'average' age that corresponds to the thicknesses in the first column. The resulting techniques should provide a radiation dose that is approximately *equal to, or less than, the estimated adult CT dose for the same procedure*. (Again, the facility medical physicist should assist with this.)

These instructions assume that all technique factors (other than tube current and/or gantry rotation time) remain fixed as techniques are adjusted for paediatric patients. Although several authors have advocated changing other parameters (i.e. kV) in order to reduce dose [127–131], these instructions will not apply if parameters other than mAs are changed. It is necessary to work closely with the medical physicist in these situations to ensure that image quality is maintained while the desired dose reductions are achieved.

The data in Tables 3 and 4 list reduction factors for the mAs provided that manual techniques are used. If the AEC features of the CT scanner are being used for imaging, the AEC system should automatically reduce techniques for children provided the adult baseline is set up properly. In this case, part A of the procedure should be followed to verify that the dose estimate from the baseline technique does not exceed recommended adult values. The baseline mAs values should be entered into the spreadsheet provided to obtain the correct paediatric techniques for the scanner. The proper functioning of the AEC system on the CT scanner can be verified by comparing the mAs values listed on the CT images with the appropriate value listed in Tables 3 and 4. If the mAs values listed on the CT images are less than or equal to the corresponding value in Tables 3 and 4, the paediatric radiation doses are less than or equal to the estimated adult radiation doses.

Reducing patient dose in CT increases the quantum mottle or background noise in the images. Since increased quantum mottle affects low contrast image quality more than high contrast image quality, dose reductions for low contrast images may be limited. For example, soft tissue differentiation (low contrast) requires lower noise in the image than studies of bony detail or lung parenchyma

TABLE 3. mAs REDUCTION FACTORS FOR THE PAEDIATRIC ABDOMEN AND THORAX

Room #: _____ **CT Unit:** _____ **Date:** _____

Abdomen baseline:	kV	mA	Time (s)	Pitch abdomen	Pitch thorax
	fill in	fill in	fill in	fill in	fill in
		Abdomen		Thorax	
PA thickness (cm)	Approx. age	mAs reduction factor (RF)	Estimated mAs = BL x RF	mAs reduction factor (RF)	Estimated mAs = BL x RF
9	newborn	0.43		0.42	
12	1 yr	0.51		0.49	
14	5 yr	0.59		0.57	
16	10 yr	0.66		0.64	
19	15 yr	0.76		0.73	
22	small adult	0.90		0.82	
25	medium adult	1.00	fill in	0.91	
31	large adult	1.27		1.16	

TABLE 4. mAs REDUCTION FACTORS FOR THE PAEDIATRIC HEAD

Room #: _____ **CT Unit:** _____ **Date:** _____

Head baseline:	kV	mA	Time (s)	Pitch	Filter
	fill in	fill in	fill in	fill in	fill in

PA thickness (cm)	Approx. age	Head mAs reduction factor (RF)	Head Estimated mAs = BL x RF
12	newborn	0.74	
16	2 yr	0.86	
17	6 yr	0.93	
19	medium adult	1.00	fill in

(high contrast). These instructions should provide adequate image quality for the paediatric soft tissue studies since the dose will be similar to adult techniques and the noise level should not change. It may be possible to reduce doses to a greater degree for high contrast studies.

Image Gently Alliance procedure for optimization of child exposures

Establishment of baseline techniques for an adult head and abdomen CT

(1) The medical physicist should determine the C_{VOL} for an adult body phantom and an adult head phantom using the standard 320 mm and 160 mm diameter body and head CT dosimetry phantoms, respectively (Appendix I).
(2) If the measured C_{VOL} of the adult abdomen or head phantoms exceed 25 mGy and 75 mGy, respectively, the medical physicist and clinical staff should work to reduce either the tube current (mA) or rotation time (s) to lower the doses.
(3) Record the final tube potential (kV), tube current (mA), rotation time (s), pitch and bow tie filter settings in Tables 3 and 4 as the baseline techniques for the adult abdomen and head.

Determination of the appropriate mAs for a paediatric thorax, abdomen and head CT

(1) Multiply the baseline (abdomen or head) tube current (mAs) by the indicated reduction factor to determine the appropriate paediatric tube current (mAs) and enter this in the table for all patient PA thicknesses/ages, or use an Excel spreadsheet[12] to perform these calculations automatically.
(2) The other parameters in the protocol (kV, pitch, bow tie filter) must remain the same. It is important to verify with the CT manufacturer that the bow tie filter in the scanner does not change if the FOV is reduced for paediatric patients.
(3) If the pitches of the thorax and abdomen scans are different, the medical physicist should calculate the correct thorax baseline from the abdomen baseline. Alternatively, the Excel spreadsheet automatically performs this correction if the different pitch values are entered.

[12] An Excel spread sheet file is available with this publication or from http://www.pedrad.org/associations/5364/ig/index.cfm?page=598

(4) When examining paediatric patients, find the mAs reduction factor from the completed tables that corresponds to the applicable PA thickness/age.
(5) The mAs ratios in Tables 3 and 4 assume that the kV used for a paediatric examination is the same as that used to determine the baseline mAs for either the head or the abdomen. If a reduced kV is selected for paediatric examinations, the suggested mAs ratios in these tables do not apply.

7. OUTLINE OF PERFORMANCE TESTS

These tests are intended to verify the performance of the CT scanner. They include tests for CT acceptance and commissioning as well as tests of the operational stability of the equipment or equipment elements used to acquire, process and display the CT images.

The tests have been classified into two types, essential and desirable, according to their importance in influencing image quality and radiation dose. The performance of the first category of tests is considered indispensable; however, it is recommended that the tests in the second category be carried out only if adequate human resources and equipment can be made available.

Test priority

Desirable describes the test procedures that should be performed, if feasible.
Essential refers to tests that *must* be carried out in a facility.

Some of the QC tests need to be performed frequently (daily or monthly). Therefore, it is recommended that these tests be performed by local personnel who are present daily in the installation (technical personnel, normally radiographers). The lower frequency tests have been assigned in the majority of cases to medical physicists and radiologists.

Suggested frequency

Acceptance: These tests are carried out to ensure the scanner delivered is operating in accordance with the manufacturer's specification.
These tests are also carried out at any major software or hardware upgrade.

Commissioning: These tests are carried out to provide a baseline for ongoing QC tests and for future optimization. They may use different phantoms and scan protocols, depending on availability of phantoms and typical scan protocols.

These tests are also carried out at any major software or hardware upgrade.

Annual: QC tests that are not likely to alter within a shorter timescale.

Monthly: QC tests that need more frequent monitoring.

Each test in the QC programme has a specified tolerance level for 'acceptable' and 'achievable' results as applicable. Should the results of a test fall outside the specified tolerance, the test should usually be repeated to confirm the result before action is taken. In some cases, only the 'acceptable' level has been defined.

Performance standards

Acceptable indicates that performance must be within certain tolerances, and if it is not, the equipment should not be used.

Achievable indicates the level of performance that should be attained under favourable circumstances, which is the level at which a facility should work, if feasible.

A facility should strive to ensure that equipment operates at the achievable level of performance (as specified in appropriate tables when defined), as this will produce the highest image quality and the most appropriate dose performance. It is recognized, however, that limited resources and other factors may occasionally prevent the achievable levels from being obtained.

Suitable minimum specifications for test equipment and phantoms are provided in Appendix I.

Table 5 lists all the tests that need to be carried out by radiographers and medical physicists for diagnostic installations (with additional tests for therapy applications also included).

TABLE 5. SUMMARY OF CT PERFORMANCE TESTS

Test number	Test name	Test personnel[a]	Diagnostic facility	Radiotherapy facility
8.2	CT alignment lights	R	✓	✓
8.3	SPR accuracy	R	✓	
8.4	CT number, image noise, image uniformity and image artefacts	R	✓	✓
8.5	Image display and printing	R	✓	✓
10.2	External CT positioning lasers	R		✓
10.3	Couch top alignment and positional accuracy	R		✓
10.4	CT number of multiple materials	R		✓
9.2	Visual inspection and review of programme	P	✓	✓
9.3	CT alignment lights	P	✓	✓
9.4	SPR accuracy	P	✓	✓
9.5	kV and half-value layer (HVL)	P	✓	✓
9.6	Radiation dose	P	✓	✓
9.7	CT number accuracy, image noise, image uniformity and image artefacts	P	✓	✓
9.8	Image display and printing	P	✓	✓
9.9	Imaged slice width	P	✓	✓
9.10	X ray beam width	P	✓	✓
9.11	Spatial resolution (MTF or modulation)	P	✓	✓
10.6	External CT positioning lasers	P		✓
10.7	Couch top alignment and index accuracy	P		✓
10.8	Gantry tilt	P		✓
10.9	CT number accuracy	P		✓

[a] R: radiographer; P: physicist.

8. RADIOGRAPHER'S TESTS

This section can be considered as a stand-alone section for radiographers and may, therefore, contain material that is duplicated elsewhere in this publication.

8.1. INTRODUCTION

A brief description of the methodology to be used when performing the radiographer's QC tests for diagnostic examinations is provided in Table 6. A listing of essential additional radiographer's tests is given in Table 9 for those facilities also providing CT images for radiation therapy treatment planning.

The order in which tests are performed does not necessarily follow the order in which they appear in this publication. The preferred order will depend on various factors relating to the CT facility as well as the evaluator's preferences, always having in mind that there exist tests whose results affect the execution of others.

8.2. CT ALIGNMENT LIGHTS

8.2.1. Introduction

The CT alignment light accuracy test shows the congruence of scan localization light and X ray field tomographic plane (Fig. 20). Patient anatomy to be scanned is often defined by scan alignment lights, therefore congruence of scan localization light and tomographic plane is a prerequisite for the proper definition of anatomy. This test is particularly important if the scanner is used for biopsy purposes.

Alignment lights on the scanner may be located within the gantry or 'internal' at the tomographic plane, outside or 'external' to the gantry at a reference distance from the tomographic plane, or both. If both internal and external alignment lights are supplied, and are independently aligned, both should be tested as detailed below.

Lights used for referencing the patient position for radiotherapy tend to be wall and ceiling mounted, these lights have tighter accuracy and testing of these is detailed in Section 10.

TABLE 6. RADIOGRAPHER'S QC TESTS (DIAGNOSTIC RADIOLOGY)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
8.2	CT alignment lights	E	Monthly	±5 mm	±1 mm
8.3	SPR accuracy	E	Monthly	±2 mm	±1 mm
8.4	CT number, image noise, image uniformity and image artefacts	E	Monthly	CT number ±5 HU[b] Noise ±25% BB[c] Uniformity ±10 HU Minimal artefacts	CT number ±4 HU Noise ±10% BB Uniformity ±4 HU Minimal artefacts
8.5	Image display and printing[d]	E	Monthly	See text	

[a] E: essential, basic requirement.
[b] HU: Hounsfield unit.
[c] % BB of baseline value.
[d] Circumstances may require more frequent testing of some printers and monitors.

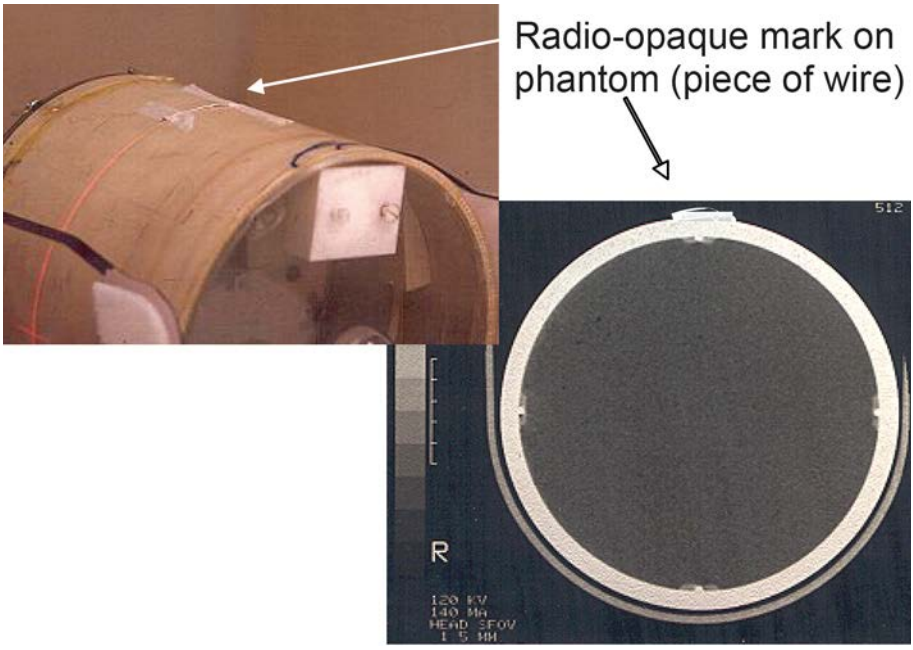

FIG. 20. Light alignment with scan plane: Imaging method.

8.2.2. Scope

Objective: To ensure that the internal and external alignment lights are properly aligned with the tomographic plane and the couch.
References: [2, 4, 7, 8].
Frequency: Monthly.

8.2.3. Equipment, materials and instrumentation

The test device should consist of a thin absorber, e.g. a wire of about 1 mm diameter. This can be used in conjunction with a standard CT phantom (see Section I.1.1).

The test device shown in Section I.2.2, or similar, may also be used, particularly if a scanner is used for radiotherapy treatment planning applications.

8.2.4. Scan protocol

A thin slice width should be used, preferably 1 mm (or the thinnest available width) in axial mode.

8.2.5. Methodology

(1) Centre the test device in the external light field parallel to the tomographic plane.
(2) Move the test device into the tomographic plane and check that the test device is aligned with the internal field light.
(3) Scan the test object with narrow axial (sequential) slices over the range of ±3 mm about the centre of the light field.

Note 1: Alternatively, an X ray film may be positioned at the isocentre and the centre of the patient positioning light marked.

Note 2: If the CT scanner provides automatic patient positioning light accuracy evaluation, based on another method, this may be used instead after it has been validated.

8.2.6. Analysis

The alignment is acceptable if the full length of the test device is visible with high contrast in the correct image. If not, determine, by comparing the images from the series, the extent of misalignment of the lasers.

8.2.7. Interpretation of results

Tolerances:

Test quantity	Acceptable	Achievable
CT alignment lights	±5 mm	±1 mm

8.2.8. Recommendations and corrective actions

(1) If the internal and the external light fields are not aligned, determine which is aligned with the tomographic plane and contact the service representative for realignment.

(2) If X ray film is used and if the measured alignment error from holes in the processed film to the midpoint of the radiation field exceeds ±5.0 mm, then run detailed tests or contact the service representative.
(3) Document the steps taken to repair the CT equipment in order to meet the standards.

All corrective actions must be documented and records of each corrective action, repair and service maintained.

8.3. SPR ACCURACY

8.3.1. Introduction

The SPR (manufacturer terminology includes: 'scoutview', 'scanogram', 'topogram', 'surview', 'pilot', etc.) is used by the radiographer to prescribe the start and finish of a CT acquisition series, as well as for measuring distances, for example in pelvimetry[13]. In addition, the SPR permits the examination to be planned and controlled accurately and provides a record of the location of images.

8.3.2. Scope

Objective: To ensure that the SPR image accurately indicates the patient position.
Reference: [4].
Frequency: Monthly.

8.3.3. Equipment, materials and instrumentation

SPR accuracy test tool (Section I.2.1).

8.3.4. Scan protocol

A 1 mm slice (or the thinnest available) width axial acquisition at each end of the scan sequence or a 1 mm reconstructed (or the thinnest available) display width.

[13] Note that SPR images result from an X ray beam divergent in only one dimension. Reliable distance information can only be given along the z axis (along the couch).

8.3.5. Methodology

(1) Place the SPR accuracy test tool along the long axis of the couch.
(2) Create an SPR image, making sure that the markers at each end of the tool are scanned.
(3) Locate the markers on the SPR image and programme 1 mm or the thinnest available slice scans directly over these markers.
(4) Initiate axial or spiral scan run.

8.3.6. Analysis

The two acquired CT slices of the markers, based on the SPR image, should clearly display each marker.

8.3.7. Interpretation of results

Tolerances:

Test quantity	Acceptable	Achievable
SPR accuracy	±2 mm	±1 mm

8.3.8. Recommendations and corrective actions

Measurements exceeding the suggested tolerances indicate the need for equipment repair by a qualified service engineer.

8.4. CT NUMBER ACCURACY, IMAGE NOISE, UNIFORMITY AND IMAGE ARTEFACTS

8.4.1. Introduction

Much information can be gained on the quality of the scanner's imaging capability from a simple scan of a water filled test object (or phantom containing a uniform material), including the measurement of CT number, image noise and uniformity, and a visual assessment of artefacts.

FIG. 21. A non-commercial head phantom with simulated bone exterior.

8.4.2. Scope

Objectives: To ensure:

(a) CT number accuracy;
(b) Noise levels of CT images being within the tolerances with respect to their reference values;
(c) Uniformity of CT number and image noise across the FOV (i.e. proper system calibration);
(d) Minimal image artefacts.

References: [4, 7, 67].

Frequency: Monthly.

8.4.3. Equipment, materials and instrumentation

Phantom as described in Section I.1.1. (The phantom used may be the manufacturer's phantom, a commercial phantom, or a simple phantom made using plastic pipe with end caps (Fig. 21).)

8.4.4. Scan protocol

As specified by the physicist at commissioning.

It should be noted that if the scanner has a software driven QC procedure, the pre-programmed settings should be used.

8.4.5. Methodology

(1) Centre the phantom in the tomographic plane.[14]
(2) Select scan relevant protocol.
(3) Scan phantom in axial mode.
(4) Repeat for additional scan protocols as specified above.

8.4.6. Analysis

(1) Measurement of CT number accuracy, image noise and uniformity, and check for absence of image artefacts are performed on the same phantom images. For MDCT scanners, all quantitative measurements are performed using a centrally located image slice.
(2) CT number and image noise are measured in a centrally placed circular ROI of appropriate diameter as specified at commissioning.
(3) ROI size is critical. The ROI diameter should be specified by the physicist at commissioning. If using CT implemented QC software, the specified ROI diameter must be used.
(4) The CT number is the measured ROI mean value and the noise is the ROI standard deviation.
(5) Uniformity is measured as the absolute difference of CT numbers between a centrally placed ROI with each of four ROIs placed on the edge. Each of these four values should be compared with the given tolerance (see Section 9.7 for ROI placement definitions).
(6) All images acquired during the noise test should be visually inspected for image artefacts. Thus, for MDCT scanners, this means examining all image slices in the acquisition, not just the central image slice.

[14] Head sized phantom to be preferably placed in the head rest. Body sized phantom to be placed on a uniform part of the table avoiding the head rest fixing structures. Manufacturer's phantoms may be attached using supplied attachment devices.

8.4.7. Interpretation of results

Test quantity	Acceptable	Achievable
CT number	±5 from baseline value[a]	±4
Image noise	±25% of the baseline value[a]	±10% of the baseline
Uniformity	±10	±4
Artefacts	No artefacts that have the potential to compromise diagnostic confidence	No visible artefacts

[a] The value as specified by the physicist at commissioning.

8.4.8. Recommendations and corrective actions

Increased image noise can result from several factors, including:

- Selection of inappropriately low kV or mAs, or both;
- Malfunction of scanner hardware and/or software;
- Artefacts inside the ROI being used for measurement of standard deviation.

If the measured values are out of tolerance or the images exhibit artefacts, or both, user calibrations of the scanner should be performed and the measurements repeated. If the problem persists, corrective actions should be initiated.

8.5. IMAGE DISPLAY AND PRINTING

8.5.1. Introduction

The correct set-up of display monitors and film printing devices is essential to achieving a good diagnostic outcome.

8.5.2. Scope

Objective: To confirm that the image displays are reproducing all of the grey scale information in the image accurately and that the printed images are a faithful reproduction of the image display.

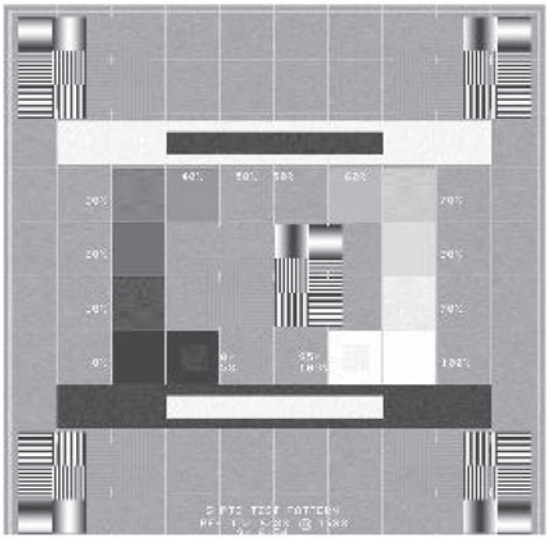

FIG. 22. SMPTE test pattern. Note the 5% and 95% patches inset within the 0% and 100% patches, respectively. These should be visible on the image displays and on the film images. Note that the 0% patch usually cannot be visualized on a printed (reflective) image such as this figure, but should be visible on CT films and displays.

References: [132, 133].

Frequency: This test should be carried out weekly, whenever service is performed on the image displays or printer, and whenever degradation in the image quality is suspected.

8.5.3. Equipment, materials and instrumentation

SMPTE test pattern (Fig. 22) or AAPM TG18-QC test pattern.[15]

8.5.4. Scan protocol

Not applicable.

[15] The appropriate test pattern must be installed and/or verified by the medical physicist during the commissioning of the scanner (see Section 9.5).

8.5.5. Methodology

(1) Display the SMPTE or AAPM test pattern on the image display.[16]
(2) Set the window width and level at the specified values. (The medical physicist can provide the appropriate values as per Section 9.8.5.)
(3) With the room lights set at the illumination level normally used, view the images on the image display.
(4) Print the test pattern.
(5) View the film images on the view box. Use the view box normally used for viewing clinical films and ensure that the room lighting conditions are the same as those used during clinical viewing of film images.

8.5.6. Analysis

Look at each image carefully. Determine if it is possible to see brightness or density differences between adjacent steps of the step wedge. Observe the visibility of the 5% and 95% inset patches.

8.5.7. Interpretation of results

Test quantity	Acceptable	Achievable
Visibility of step wedge patches, and of the 5% and 95% inset patches	The brightness or density differences between adjacent steps of the step wedge should be visible. Both the 5% and 95% inset patches must be visible.	

8.5.8. Recommendations and corrective actions

(1) Each of the steps of the step wedge should be visible and appear different in brightness or density from the adjacent steps. The 5% and 95% inset patches must be visible.

[16] This test must be carried out for each image display associated with the CT scanner, including the operator's console, radiologist's workstation, etc., to ensure that the displays are producing images of similar quality.

(2) Loss of visibility can be caused by changes in the image displays or printer, or by changes in room illumination levels or view box luminance. As the illumination levels increase, it becomes more difficult to discern subtle differences, especially for the darker steps and the 5% inset patch. Confirm that the room illumination levels are at those normally used.

(3) If all steps and inset patches are still not visible, corrective action must be initiated. Contact the medical physicist for recommendations regarding corrective action.

9. MEDICAL PHYSICIST'S TESTS[17]

9.1. INTRODUCTION

Table 7 lists the tests to be performed by the medical physicist. The order in which tests are performed does not necessarily have to be the same as they appear in this publication. The preferred order will depend on various factors relating to the CT facility and the medical physicist's preferences. It should always be considered that some test results are required before other tests can be performed effectively. All annual tests would also be carried out at acceptance or after having changes made to equipment, including software updates, as detailed in the tests themselves. Additionally, at commissioning, the baseline values for radiographer's tests would be established. This requirement is indicated in the test descriptions themselves and is also indicated in the suggested frequency column of Table 7. In most MDCT units, many of the tests below are prescribed by the manufacturer, using the provided suitable phantoms, and are automated for rapid execution. This may not be the case with older scanners.

[17] It is considered that the measurement of kV and HVL is not necessary for CT scanners for either QA or patient safety reasons.

TABLE 7. MEDICAL PHYSICIST'S TESTS (DIAGNOSTIC RADIOLOGY)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
9.2	Visual inspection and review of programme	E	Acceptance and annually	All items must meet local standards	
9.3	CT alignment lights	E	Annually	±5 mm	±1 mm
9.4	SPR accuracy	E	Annually	±2 mm	±1 mm
9.5	kV and HVL[b]	E	Acceptance and after changes	kV ±5% nominal HVL ≥ specified by radiation protection regulations and within the tolerances specified by the manufacturer	kV ±2% nominal HVL ≥ specified by radiation protection regulations and within the tolerances specified by the manufacturer
9.6	Radiation dose	E	Annually	±20% of manufacturer's specifications ±20% of baseline	±10% of baseline
9.7	Image noise	E	Commissioning and annually	±25% of baseline	±10% of baseline
9.7	Image uniformity	E	Commissioning and annually	±10 HU for head and body	±4 HU for head and body
9.7	CT number accuracy	E	Commissioning and annually	±5 HU (water) ±10 HU (other material) compared with baseline values	±4 HU

TABLE 7. MEDICAL PHYSICIST'S TESTS (DIAGNOSTIC RADIOLOGY) (cont.)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
9.8	Image display and printing	E	Commissioning and annually	See text	
9.9	Imaged slice width	E	Annually	+0.5 mm for <1 mm ±50% for 1–2 mm; ±1 mm above 2 mm	
9.10	X ray beam width	D	Acceptance and after changes	Within manufacturer's specifications	
9.11	High contrast spatial resolution (MTF or modulation)	D	Acceptance and after changes	Within manufacturer's specifications	

[a] E: essential, basic requirement; D: desirable.
[b] It is considered that the measurement of kV and HVL is not necessary for CT scanners for either QA or patient safety reasons.

9.2. VISUAL INSPECTION AND PROGRAMME REVIEW

9.2.1. Introduction

The visual inspection of a CT facility can achieve much without the use of specialized test equipment. *Warning*: (i) do not test the emergency off button except in the presence of the supplier engineer, (ii) interlocks on doors should be configured in such a way that a person entering the scanner room will not automatically shut down the X ray output of the CT scanner.

9.2.2. Scope

Objective:

(a) To ensure that the CT scanner and adjacent areas are safe and have the necessary radiation protection supplies, such as lead aprons and portable barriers.
(b) To ensure that the CT QC tests are being carried out at the appropriate intervals and that corrective action is taken, when necessary.
(c) To provide assistance and feedback relative to radiation safety and QC to the CT radiologists and radiographers.

Frequency: This evaluation is to be carried out at acceptance testing and annually thereafter.

9.2.3. Equipment, materials and instrumentation

(1) Visual inspection and programme review checklist. (A sample form is included in the Annex. This should be modified by the medical physicist to include other items of importance at the specific facility.)
(2) QC charts and logs.
(3) Written safety procedures.

9.2.4. Scan protocols

Not applicable.

9.2.5. Procedure

(1) Evaluate the CT scanner, scanner room, radiologist's reporting room, operators' area and computer room using the visual inspection checklist.

(2) Review the CT QC charts and logs with the CT radiographers. Anomalies in the data, missing data, or lack of corrective action or follow-up should be discussed.
(3) Review the scan protocols and written safety procedures and discuss them with the CT radiologists and radiographers.

9.2.6. Analysis

Not applicable.

9.2.7. Interpretation of results

Test quantity	Acceptable	Achievable
Visual inspection checklist	All items on the checklist should be present and functional.	
QC charts and logs	All tests are carried out as scheduled. Corrective action is requested and QC tests are performed after corrective action to ensure the system is functioning properly.	
Written safety procedures	All staff must be aware of the written safety procedures and follow them, as appropriate.	

9.2.8. Recommendations and corrective actions

(1) Each of these evaluations is an essential part of the facility's QA programme. If test procedures are missing or equipment is not functioning properly, then appropriate action, including repair or replacement, as the case demands, must be undertaken.
(2) The CT radiographers are responsible for carrying out the routine QC tests. If the results fall outside control limits or do not meet the stated criteria, the responsible physicist should be notified.
(3) All staff must be aware of, and follow, the written safety procedures.
(4) The medical physicist should discuss the results of the programme review with the responsible radiologist. Emphasis should be placed on areas requiring corrective action. The radiologist and the medical physicist should discuss deficiencies with the radiographers and stress the importance of these issues for the safety of both patients and staff.

9.3. CT ALIGNMENT LIGHTS

9.3.1. Introduction

The CT alignment lights accuracy test shows the congruence of the scan localization light and the X ray field tomographic plane. Patient anatomy to be scanned is often defined by scan alignment lights; therefore, congruence of the scan localization light and the tomographic plane is a prerequisite for proper definition of the anatomy. This test is especially important if the scanner is to be used for biopsy purposes.

Alignment lights on the scanner may be located within the gantry or 'internal' at the tomographic plane, outside or 'external' to the gantry at a reference distance from the tomographic plane, or both. If both internal and external alignment lights are supplied and independently aligned, both should be tested as detailed below.

Positioning lasers used for referencing the patient position for radiotherapy tend to be wall and ceiling mounted. These have tighter tolerances and testing of these is detailed in Section 10.

9.3.2. Scope

Objective: To confirm that the internal and external alignment lights are properly aligned with CT gantry and couch.
References: [4, 7, 8].
Frequency: Monthly.

9.3.3. Equipment, materials and instrumentation

The test device consists of a thin absorber, e.g. a wire with a diameter of about 1 mm. This can be used in conjunction with a standard CT phantom, such as that detailed in Section I.1.1.

The test device shown in Section I.2.2, or a similar one, may also be used, particularly if the scanner is used for radiotherapy treatment planning applications.

9.3.4. Scan protocol

A thin slice width should be used, preferably 1 mm (or the thinnest available) in axial mode.

9.3.5. Methodology

(1) Centre the test device in the external light field parallel to the tomographic plane.
(2) Move the test device into the tomographic plane and check that the test device is aligned with the internal field light.
(3) Scan the test object with narrow axial slices over the range of ±3 mm about the centre of the light field.
(4) Examine the images and make a hard copy, if appropriate.

Note 1. Alternatively, an X ray film may be positioned at the isocentre and the centre of the patient positioning light marked.

Note 2. If the CT scanner provides automatic patient positioning light accuracy evaluation, based on another method, this may be used instead after it has been validated.

9.3.6. Analysis

The alignment is acceptable if the full length of the test device is visible with high contrast in the correct image. If not, determine, by comparing the images from the series, the extent of misalignment of the lasers.

9.3.7. Interpretation of results

Coincidence between the pin pricks and the X ray beam exposures indicates alignment between the internal lights and the tomographic plane.

Tolerances:

Test quantity	Acceptable	Achievable
CT alignment lights	±5 mm	±1 mm

9.3.8. Recommendations and corrective actions

(1) If the internal and the external light fields are not aligned, determine which, if either, is aligned with the tomographic plane and contact the service representative for realignment.

(2) If the film is used and if the measured alignment error from holes in the processed film to the midpoint of the radiation field is greater than ±5.0 mm, then run detailed tests or contact the service representative.
(3) Document the steps taken to repair the CT equipment to meet standards.

All corrective actions must be documented and records of each corrective action, repair and service carried out maintained.

9.4. SPR ACCURACY

9.4.1. Introduction

The SPR (manufacturer terminology includes 'scoutview', 'scanogram', 'topogram', 'surview', 'pilot', etc.) is used by the radiographer to prescribe the start and finish of a CT acquisition series, as well as for measuring distances, for example, in pelvimetry.[18] In addition, the SPR permits the examination to be planned and controlled accurately, and provides a record of the location of the images.

9.4.2. Scope

Objective: To ensure that the SPR image accurately indicates the patient position.
Reference: [4].
Frequency: Monthly.

9.4.3. Equipment, materials and instrumentation

SPR accuracy test tool (Section I.2.1).

9.4.4. Scan protocol

A 1 mm (or the thinnest available) slice width axial acquisition at each end of the scan sequence or a 1 mm (or the thinnest available) reconstructed display width.

[18] Note that SPR images result from an X ray beam divergent in only one dimension. Reliable distance information can only be given along the z axis (along the couch).

9.4.5. Methodology

(1) Place the SPR accuracy test tool along the long axis of the couch.
(2) Create an SPR image making sure that the markers at each end of the tool are scanned.
(3) Locate the markers on the SPR image and enter 1 mm or minimum slice scans directly over these markers.
(4) Initiate axial or spiral scan run.

9.4.6. Analysis

The two CT slices acquired of the markers based on the SPR image should be centred over each marker.

9.4.7. Interpretation of results

Tolerances:

Test quantity	Acceptable	Achievable
SPR accuracy	±2 mm	±1 mm

9.4.8. Recommendations and corrective actions

Measurements exceeding the suggested tolerances indicate a need for equipment repair by a qualified service engineer.

9.5. X RAY GENERATOR (kV ACCURACY AND HVL)

It is considered that the measurement of kV and HVL is not necessary for CT scanners for either QA or patient safety reasons. If the patient dose, as indicated by the CTDI, is excessively high, one may investigate the kV and HVL to determine the potential causes of the high dose. If the kV is suspected of being outside tolerance, or that sufficient filtration in not present, then measurements of scanner output with the CTDI may be compared with those published by independent organizations (such as ImPACT http://www.impactscan.org/) or with manufacturer's specification data.

FIG. 23. Measurement of dose in a standard CTDI PMMA body phantom (left) and in air (right).

9.6. RADIATION DOSE

9.6.1. Introduction

Measurement of CT dose is essential to ensure the constancy of the CT scanner X ray output and performance and, most importantly, to assist in optimization of procedures to obtain the necessary image quality at a known and acceptable radiation dose to the patient (Fig. 23).

9.6.2. Scope

Objectives:

At acceptance:

(a) To compare the determined C_{VOL} ($CTDI_{VOL}$) with the manufacturer's specifications;

(b) To compare the C_{air} ($CTDI_{air}$) values for all X ray beam collimations with the manufacturer's specifications (if available).

At commissioning testing:

(a) To determine the C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) values for selected protocols representing the main examinations clinically performed at the facility;

(b) To measure the C_{air} ($CTDI_{air}$) values for all X ray beam collimations and kVs, if not already completed at acceptance;

(c) To confirm, for selected scan protocols, the accuracy of the displayed C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) values on the scanner console, if applicable.

At annual tests:

(a) Review and update the C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) values for selected protocols representing the main examinations clinically performed at the facility;
(b) Measure a small set of C_{air} ($CTDI_{air}$) values to confirm output consistency;
(c) Review, for selected scan parameters, the accuracy of the displayed C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) values on the scanner console, if applicable.

References: [4, 26, 37, 134].

Frequency: Acceptance, commissioning and annual testing. In addition, tests have to be performed after a system service which might impact on dose, including, but not limited to, replacement of an X ray tube or filtration component.

9.6.3. Equipment, materials and instrumentation

(1) Suitable dosimeter and CT dose phantoms (see Section I.3.1).
(2) Chamber stand for free-in-air measurement and devices to stabilize and secure the phantom.

9.6.4. Scan protocol

At acceptance, as specified by the purchase specifications.

At commissioning testing, the specified scans should be acquired at a minimum using one typical kV, mA, scan time setting, CT mode (head or body, axial or helical) and typical abdomen/soft tissue reconstruction kernel. The total beam width applied should be that typically used with patients, and matrix size should also be noted. More extensive testing is recommended to give the physicist an understanding of the system noise performance. Ideally, this might include:

(a) All clinically used kV settings to be tested on relevant phantom sizes, with appropriate radiation filter settings;
(b) Measurement in axial and helical modes.

Such data will serve as a platform for the more detailed studies needed when optimizing scan protocols.

At annual tests, as specified during commissioning.

9.6.5. Methodology

(a) Measurement of C_{VOL} ($CTDI_{VOL}$)
   (1) Place the head phantom on the head rest.
   (2) Move the phantom into the tomographic plane so that the tomographic plane bisects the length of the phantom.
   (3) Centre in the FOV vertically and horizontally using CT lights.
   (4) Place the ion chamber in the centre hole of the phantom and use PMMA plugs to fill unused holes.
   (5) Ensure the phantom is not tilted or twisted. Alignment to within 5° is acceptable.
   (6) Perform the SPR, both anterior–posterior and laterally, to ensure the phantom is centred vertically and horizontally. Position accuracy to within ±1 cm is acceptable.
   (7) Verify the centring within the body of the phantom by taking a single axial slice.
   (8) Select the scan protocol. WARNING: Some scanners may over-rotate unless the rotation is restricted to 360°.
   (9) Record the measured dosimeter reading.
   (10) Note the displayed $CTDI_{VOL}$ from the acquisition workstation monitor for that particular scan protocol, if available.
   (11) Reposition the ion chamber in one of the peripheral holes in the phantom.
   (12) Repeat steps (8)–(11) until the dose in all four peripheral positions has been measured.
   (13) Place the body phantom on the couch, avoiding the extreme end of the patient couch.
   (14) Repeat steps (2)–(12).

(b) Measurement of C_{air} ($CTDI_{air}$)
   (1) Place the ion chamber so that it overhangs the end of the scanner couch.
   (2) Move the chamber into the tomographic plane so that the tomographic plane bisects the length of the sensitive volume of the ion chamber.
   (3) Ensure the ion chamber is not tilted or twisted. Alignment to within 5° is acceptable.

(4) Perform the SPR, both anterior–posterior and laterally, to ensure the ion chamber is centred vertically and horizontally. Position accuracy to within ±1 cm is acceptable.
(5) Verify centring by taking a single axial slice.
(6) Select approximately 120 kV and 100 mAs with the largest X ray beam collimation and scan the ion chamber in axial mode.
(7) Record the measured dosimeter reading.
(8) Repeat steps (6) and (7) for all other X ray beam collimations.
(9) Repeat the measurement using the lowest and highest kV at the reference X ray beam collimation width[19].

(c) Accuracy of the C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) values displayed on the scanner console
(1) Compare the measured C_{VOL} ($CTDI_{VOL}$) and $P_{KL,CT}$ (DLP) data with the values displayed on the CT console for the same acquisition parameters.

9.6.6. Analysis

(a) Measurements for comparison with manufacturer's specifications
(1) Calculate the C_{VOL} ($CTDI_{VOL}$) values from the dosimeter readings for the central and peripheral measurements using Eq. (5) (Section 3.3.1).
(2) Repeat for other collimations and phantoms as required.
(3) Calculate the $P_{KL,CT}$ (DLP) from the C_{VOL} ($CTDI_{VOL}$) values (note that for a single axial rotation, the C_w and C_{VOL} values are identical).
(4) Compare with manufacturer's specifications.

(b) Measurement of C_{air} ($CTDI_{air}$) for all beam collimations
(1) Calculate the C_{air} values from the dosimeter readings for all X ray beam collimations and kV used.
(2) Compare these values with results from acceptance or previous constancy testing.
(3) Plot the graph of C_{air} versus X ray beam collimation for constant kV values.
(4) Normalize the C_{air} series of values using the reference C_{air} value.

[19] A reference collimation beam width of 20 mm is recommended or, if not possible, the largest possible collimation width below 20 mm.

9.6.7. Interpretation of results and tolerances

Test quantity	Acceptable	Achievable
C_{VOL}	(a) <±20% between manufacturer's and measured (b) <±20% between displayed and measured	
C_{air}	For each X ray beam collimation at acceptance: (a) <±20% compared with manufacturer's specifications and measured (b) annually <±20% compared with baseline	

9.6.8. Recommendations and corrective actions

(1) Should any of the values be outside the tolerances, service is required.
(2) In the case of CT scanners with total beam widths in excess of 40 mm, follow appropriate advice [37].
(3) Check with clinical staff on the use of any non-standard combinations of scan fields of view and filter combinations. It is important to check the C_{VOL} for these configurations and compare to that displayed.
(4) Similarly, it is important to verify if the C_{VOL} measurements displayed for paediatric use are based on a 32 cm, 16 cm or other diameter dose phantom.

9.7. CT NUMBER ACCURACY, IMAGE NOISE, UNIFORMITY AND IMAGE ARTEFACTS

9.7.1. Introduction

Much information on the quality of the scanner's imaging capability can be gained from a simple scan of a water filled test object (or phantom containing a uniform material). From this, measurement of CT number, image noise and uniformity, combined with a visual assessment of artefacts, can be made.

9.7.2. Scope

Objectives:

At acceptance, to ensure that:

(a) CT number values comply with the manufacturer's specifications for defined acquisition parameters and phantoms;

(b) Noise values comply with the manufacturer's specifications for defined reconstruction settings;
(c) CT numbers in a homogeneous medium are uniform;
(d) Artefacts are not visible.

At commissioning testing, to ensure that:

(a) CT number values are recorded over a range of clinically relevant X ray tube energies and phantom sizes;
(b) CT homogeneity is recorded over a range of clinically relevant X ray tube energies and phantom sizes;
(c) Noise values and uniformity baseline values are set for radiographer's QC testing (see Section 8.4).

At annual tests, to ensure that image noise and uniformity remain within the specified tolerance limits of baseline measurements.

References: [4, 7].

Frequency: Acceptance, commissioning and annual testing. In addition, after service which might impact on noise, including, but not limited to, replacement of an X ray tube, system calibration, software changes or upgrades, etc.

9.7.3. Equipment, materials and instrumentation

At acceptance, the test phantom must be as specified by the purchase specifications. This phantom is usually supplied by the manufacturer.

At commissioning testing, the test phantom as described in Section I.1.1.

At annual tests, the test phantom as described in Section I.1.1, including both head and body phantoms.

9.7.4. Scan protocols

At acceptance, as specified by the purchase specifications.

At commissioning testing, the specified scans should be acquired at a minimum using one typical kV, mA and scan time setting, CT mode (head or body, axial or helical) and typical abdomen/soft tissue reconstruction kernel. The collimation applied, reconstructed slice widths (in MDCT) and matrix size should correspond to those typical of clinical conditions and should also be noted. More extensive testing is recommended to give the physicist an understanding of the system noise performance. Ideally, this might include:

(a) All clinically used kV settings to be tested on relevant phantom sizes;
(b) A range of reconstruction filters for the baseline scan parameters determined for radiographers;
(c) Measurement in axial and spiral mode.

Such data will serve as a platform for the more detailed studies needed when optimizing scan protocols.

The measurement of noise profiles for MDCT units to investigate interslice noise variation could also be useful at commissioning.[20]

The automatic QC routines should not be used by the physicist if the test results and methodology are not clearly indicated.

At annual tests, as specified above in commissioning.

If the scanner has a software driven QC procedure, the pre-programmed settings could be used.

9.7.5. Methodology

(1) Centre the phantom in the tomographic plane.[21,22]
(2) Select the scan relevant protocol.
(3) Scan the phantom in axial mode.
(4) Repeat for additional scan protocols as specified above.

Establishing baseline values for the radiographer's constancy protocol

Since image noise depends on scan and reconstruction parameters, these must be unambiguously defined for measurement and when setting baseline values for regular constancy tests.

The use of axial mode is preferred over helical, to avoid noise values depending on local z interpolation weighting factors and interpolation algorithms.

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[20] In MDCT units, abnormalities in noise profile may indicate detector problems or a misalignment of the collimators with the tube axis. Therefore, it is advisable, if possible, to assess the noise for all detector rows and compare their values to their average and to manufacturer specifications. If provided, manufacturer's protocols should be adopted for this testing.

[21] Head sized phantom to be preferably placed in the head rest. Body sized phantom to be placed on a uniform part of the table avoiding the head rest fixation structures. Manufacturer's phantoms may be attached using supplied attachment devices.

[22] For scanners which have an FOV larger than 50 cm (mostly on large bore scanners used in radiotherapy), it is recommended that the phantom be moved off-centre to verify uniformity in the image periphery. This is especially important for therapy, as the patient is often positioned to the side of the scanner due to treatment immobilization devices.

9.7.6. Analysis

(1) Measurement of CT number accuracy, image noise and uniformity and check of absence of image artefacts are performed on the same phantom images.
(2) CT number and image noise are measured in a centrally placed circular ROI of appropriate diameter.
(3) ROI size is critical. The ROI diameter should be specified by the physicist at commissioning. If using CT implemented QC software, the specified ROI diameter must be used. The following criteria have been recommended by the IEC [7]:
    — For the measurement of CT number, the diameter for the ROI should be approximately 10% of the diameter of the image of the phantom.
    — For the measurements of noise, the diameter for the ROI should be approximately 40% of the diameter of the image of the phantom.
    — The ROI in the centre should not overlap an ROI near the edge of the phantom.
(4) The CT number is the measured ROI mean value, and noise is the ROI standard deviation.
(5) Uniformity is measured as the absolute difference of CT numbers between a centrally placed ROI with each of four ROIs placed on the edge. Each of these four values is to be compared with the given tolerance (see Fig. 24 for ROI placement).
(6) All images acquired during the noise test are to be visually inspected for image artefacts. Thus, for MDCT scanners, this means examining all image slices in the acquisition, not just the central image slice (as is needed only for the ROI measurements).
(7) In MDCT scanner constancy tests, noise levels should also be measured in the outermost slices.

For evaluation of the images, image processing software on the CT may be used or images transferred to a PC or workstation may be evaluated using commercial or public domain image processing software. A DICOM compliant image analysis software with scripting capabilities may be considered.

9.7.7. Interpretation of results and tolerances

At acceptance, the test results must comply with the tolerances specified at the time of purchase and as usually supplied by the manufacturer.

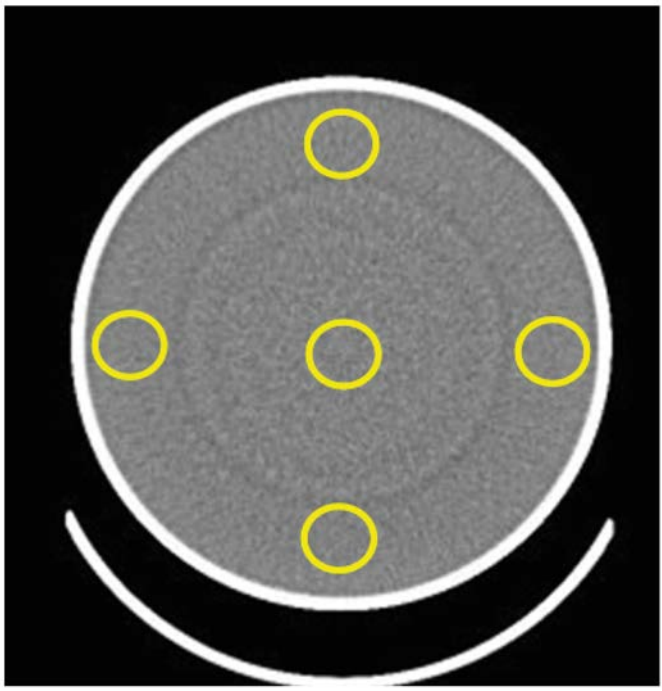

FIG. 24. Placement of ROIs for noise and uniformity measurements. Central ROI used for noise; peripheral and central ROIs for uniformity.

At commissioning testing, baselines will be established and tolerances confirmed.

At annual tests, the tolerances are as below:

Test quantity	Acceptable	Achievable
CT number	±5 from baseline value	±4
Image noise	±25% of the baseline value	±10% of the baseline
Uniformity	±10	±4
Artefacts	No artefacts that have the potential to compromise diagnostic confidence	No visible artefacts

9.7.8. Recommendations and corrective actions

(1) If the noise level is higher than the manufacturer specified, the tube voltage (kV) and mAs used for the test procedure should be checked. If measurements remain outside of tolerance, the system should be recalibrated using a calibration procedure that can be performed by the user. The test should be repeated. In the case of persistent out-of-tolerance results, corrective service is required.
(2) In many cases, increased noise measurement results can be due to the presence of artefacts in the image, especially ring artefacts. Corrective action to minimize the artefacts, for example, performing an air calibration, is necessary before useful noise measurements can be obtained on repeat measurement.
(3) Any problems in the measurement and results for noise profiles for MDCT units should be discussed with the manufacturer's agent.
(4) Testing uniformity will mainly reveal problems with beam hardening correction.

The noise value from successive scans may vary by up to 15%, depending on the size of the ROI. Therefore, a number of scans (e.g. ten) can be taken to obtain a mean noise value with greater precision.

9.8. IMAGE DISPLAY AND PRINTING

9.8.1. Introduction

The quality of the image display and the associated view conditions for the acquisition console, diagnostic workstation and display of printed images as appropriate is critical for the effective diagnosis of CT images.

9.8.2. Scope

Objectives:

At acceptance, to ensure that the image displays are performing to specifications, including:

(a) Geometric distortion and size calibration, luminance response;
(b) Display resolution (low and high contrast), display noise, veiling glare, display chromaticity;
(c) SMPTE or AAPM TG18-QC test pattern.

At commissioning testing, to ensure that room illuminance and display reflections are acceptable.
At annual tests, to ensure:

(a) Size calibration;
(b) Room illuminance and display reflection, luminance response;
(c) Display resolution (low and high contrast), display noise;
(d) SMPTE or AAPM TG18-QC test pattern;
(e) Printer testing, which should include mid-density, density difference, visibility of 5% and 95% patches, high and low contrast resolution and visual appearance, are all acceptable.

References: [132, 133, 135–138].
Online resources:

SMPTE Online Test Pattern Tutorial
http://brighamrad.harvard.edu/research/topics/vispercep/tutorial.html

AAPM TG-18 Test Pattern Information
http://aapm.org/pubs/reports/OR_03.pdf. The test patterns are available at http://aapm.org/pubs/reports/OR_03_Supplemental/

Monitor QC Program for Planar Displays
http://www.planar.com/products/docs/MBU/white_papers/Planar-MonitorQCProgram.pdf

Frequency: Acceptance and commissioning testing, annually, and whenever service is performed on the image displays or printer, and whenever degradation in the image quality is suspected.

9.8.3. Equipment, materials and instrumentation

(1) Photometer to measure both luminance and illuminance, capable of measuring small area (5–10 mm diameter) luminance on displays.
(2) Optical densitometer with a relatively small aperture, e.g. 3 mm.

9.8.4. Scan protocol

Not applicable.

9.8.5. Methodology

(1) Display the SMPTE or AAPM TG18-QC test pattern on the image display. Note that this test must be carried out for each image display associated with the CT scanner including the operator's console and radiologist's workstation to ensure that the displays are producing images of similar appearance and quality.
(2) Set the window width and level at the specified values. Note that the window width should be set to encompass all of the numerical values in the test image. For example, if the test image is a 9-bit image (maximum number 512), then the width should be set at 512. The level should be set to the value dependent on the operating system. For example, in some systems the level must be set at 0 or 1, the lower end of the 512 range. In other systems the level should be set at 256, the middle of range of numerical values in the test image.
(3) With the room lights set at the illumination level normally used, critically view the test image on the display.
(4) Using a calibrated photometer, measure the luminance of the 100%, 90%, 40% and 10% patches. Determine the difference in the luminance levels between the 90% and 40% patches, and between the 40% and 10% patches.
(5) Measure the room illuminance. Note that the luminance and illuminance are measured with different detectors on the photometer. Ensure that the appropriate detector is used.
(6) Print the test pattern on the printer.
(7) Critically view the film images on the view box. Use the view box normally used for viewing clinical films and ensure that the room lighting conditions are the same as those used during clinical viewing of film images.
(8) Measure the view box luminance and the room illuminance.
(9) Measure the optical densities of the 0%, 10%, 40% and 90% film patches using a densitometer.

9.8.6. Analysis

(1) Look at each test pattern image on both the image display and film carefully.
(2) Determine if it is possible to see brightness or density differences between adjacent steps of the step wedge.
(3) Observe the visibility of the 5% and 95% inset patches.
(4) Observe low and high contrast resolution patterns. These should be used for constancy testing.

(5) Refer to Assessment of Display Performance for Medical Imaging Systems (AAPM, 2005) for further information and procedures.

9.8.7. Interpretation of results

Test quantity	Acceptable	Achievable
Visibility of step wedge patches and of the 5% and 95% inset patches	The brightness or density differences between adjacent steps of the step wedge should be visible Both the 5% and 95% inset patches should be visible	
Film densities	Measured film densities should be as noted in Table 8	
Display maximum luminance levels	Primary workstation display[a]: 500 cd/m^2 Secondary display[b]: 200 cd/m^2 (typical values for LCD displays)	Primary workstation display: 600 cd/m^2 Secondary display: 250–300 cd/m^2
Room illuminance levels	15–50 lux	
View box luminance level	A minimum of 1800 cd/m^2	A minimum of 3000 cd/m^2

[a] Primary display refers to the workstation used by the radiologist for primary interpretation.
[b] Secondary display refers to displays used by other physicians for reviewing images.

9.8.8. Recommendations and corrective actions

(1) For both image displays and film images, each of the steps of the step wedge should be visible and appear different in brightness or density from the adjacent steps. The 5% and 95% inset patches must be visible.
(2) Loss of visibility can be caused by changes in the image displays or the printer, or by changes in room illumination levels. As the room illumination levels increase, it becomes more difficult to discern subtle differences, especially for the darker steps and the 5% inset patch. Confirm that the room illumination levels are those normally used.
(3) If all steps and inset patches are still not visible, corrective action must be initiated.

TABLE 8. TYPICAL FILM DENSITIES FOR SMPTE TEST PATTERN IMAGES

Patch	Density	Tolerance (±)
0%	2.45	0.10
10%	2.10	0.10
40%	1.15	0.10
90%	0.30	0.05

(4) Degradation in the maximum luminance level occurs over time with CRT type displays. (This does not occur as dramatically for LCD type displays.) However, the monitors can usually be recalibrated to maintain the maximum luminance level.
(5) The image display luminance differences from step to step are impacted primarily by the display calibration, the look-up table and the room illuminance level, and should remain relatively constant over time. Consequently, the differences noted at acceptance testing should be maintained.
(6) The densities of the test pattern patches are impacted primarily by the type of film, film processor (development time and temperature, chemical activity, etc.) and look-up table. It is essential to have a QC programme for the film printer and processor in order to maintain the appropriate film densities.

Corrective action for displays

(7) If the 5% or 95% patches are not visible on the display, ensure that the room illumination levels do not exceed 15–50 lux.
(8) If the room illumination levels are appropriate, then it will be necessary to readjust the brightness and contrast levels on the display, if these are accessible to the medical physicist. First, set both the brightness and contrast levels to zero. Adjust the brightness control until the background on the display is just visible. Next, adjust the contrast control until the 95% patch is readily visible. Continue to increase the contrast adjustment until distortion is seen in the alphanumerics. When this occurs, decrease the contrast setting slightly until the distortion disappears. If the brightness and contrast controls are independent, the display should now be properly

adjusted. If the controls are not independent, it may be necessary to readjust the brightness control to obtain an appropriately dark background (but with the 5% patch visible) and then readjust the contrast control. If the 5% and 95% patches cannot be displayed at the same time, then it will be necessary to obtain assistance from the service engineer to recalibrate the display.

Corrective action for film printers

(9) Review the QC charts for the film printer. If the printer is a 'wet process' system, i.e. one using liquid developer, fixer and wash, then ensure that the chemicals are fresh and that the processor is in control.
(10) Print a test film from the printer and confirm that the densities obtained on the test film are within the control limits for the printer. If this is not the case, check the expiration date for the film emulsion batch being used. If the film is expired, switch to a new emulsion batch with an appropriate expiration date. If the film is performing well, contact the manufacturer or check for error in the LUT.

9.9. IMAGED SLICE WIDTH

9.9.1. Introduction

This test assures the user that the displayed image represents a specified thickness of tissue. This thickness is no longer directly governed by the acquisition slice width as set by the collimation and detector slice width. In MDCT units, the image slice width is selected in software. However, the thinnest image slice width is limited by the traditional acquisition parameters of collimation and detector width and pitch.

9.9.2. Scope

Objective: To ensure that the reconstructed imaged slice width is similar to that selected on the CT scanner console.
References: [4, 7, 8].
Frequency: Acceptance testing, annually, and as part of the end of warranty testing, and whenever service is performed on the CT system which might have an impact on the image quality, including, but not limited to, replacement of an X ray tube, system calibration, software changes or upgrades, etc.

9.9.3. Equipment, materials and instrumentation

See Section I.1.5.

9.9.4. Scan protocol

At acceptance, as specified by the purchase specifications. This usually includes all collimator settings. In MDCT scanners the maximum number of imaged slice widths should be acquired for each collimator setting [7].

At commissioning testing, it is recommended that:

(a) The clinical range of axial slice widths is measured, as determined by combinations of collimation and detector settings.
(b) Helical slice widths should be determined at a number of acquisition and reconstructed slice thicknesses. It is recommended that the reconstructed image slice width be the same width as the effective detector thickness and that spot checks be made for larger reconstructed slice widths. These should also be investigated under different pitch conditions.
(c) Where possible, scan protocols used for noise, dosimetry and slice width should be standardized and based on the clinical requirements of usage of the CT scanner.

Such data should later be analysed in conjunction with similar dosimetric data to assist in future detailed studies needed when optimizing scan protocols.

At annual tests, as specified above in commissioning.

The number (if any) and the nature of annual testing should be determined during the commissioning of the CT scanner.

9.9.5. Methodology

Axial mode

(1) Align the axial z sensitivity test device so that its axis coincides with the axis of rotation of the CT scanner with the image centred in the FOV.
(2) Perform an SPR to confirm acceptable alignment and to define the tomographic plane (see Fig. 25).
(3) Select a scan protocol as determined above.
(4) Scan the test object as required.
(5) Repeat for all other selected scan protocols.

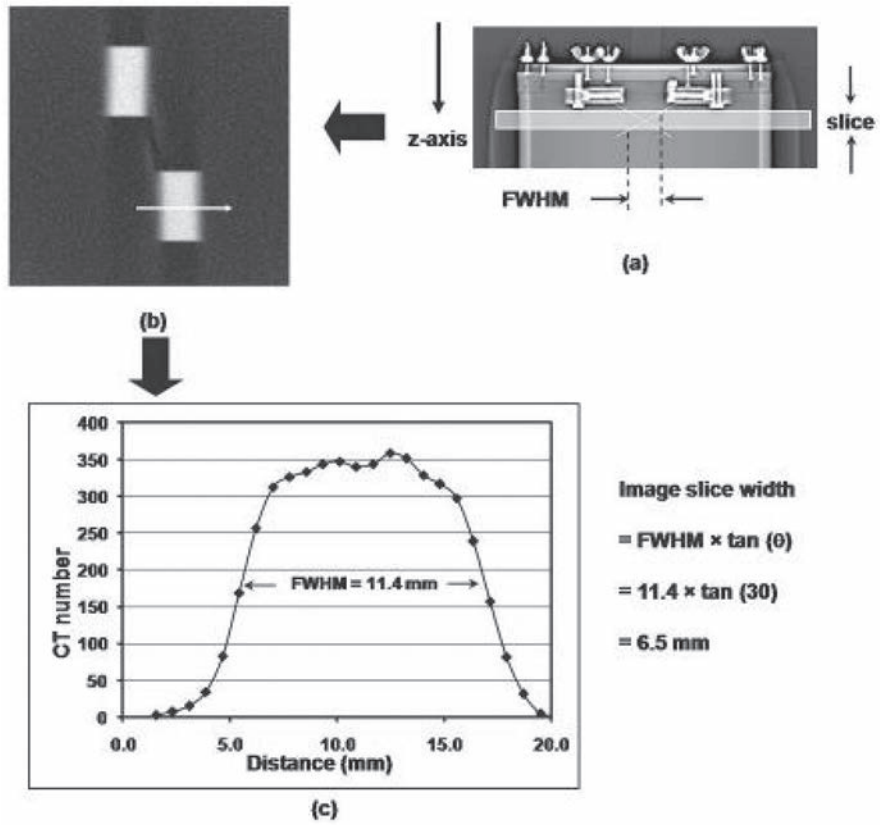

FIG. 25. (a) SPR image of the ramp test object, (b) image through the inclined ramps, (c) profile through image in (b). Nominal slice width is 6 mm.

Helical mode

(1) Place the helical z sensitivity test object on the CT couch or secure it on a stand so that it is centred in the FOV and aligned such that the metal foil insert is parallel to the tomographic plane.
(2) Perform an SPR and define the scanned volume for helical scanning to ensure the metal foil is fully imaged.
(3) Select a scan protocol as defined in the above section.
(4) Scan the test object.
(5) Reconstruct images at intervals of approximately one tenth of the nominal imaged slice width.

(6) Repeat step (5) for other clinically relevant imaged slice widths.
(7) Repeat steps (3)–(6) for other scan protocols.

9.9.6. Analysis

The evaluation for MDCT should be performed at least for both outer tomographic sections and for one representative inner tomographic section.

Axial mode

(1) Plot the profile of CT numbers across the image of the ramp (see Fig. 25(c)).[23]
(2) For MDCT, one scan will result in multiple images. For annual testing, imaged slice widths should be measured at least in the central slice position. For acceptance, imaged slice widths should be measured at least for both outer slices and one representative inner slice [7].
(3) Determine the CT number maximum (CT_{max}) and CT number background (CT_b).
(4) Determine the CT number corresponding to half the maximum height using the following relationship:

$$CT_{half} = (CT_{max} - CT_b)/2 + CT_b$$

(5) Determine the full width half maximum (FWHM) in pixel values by calculating the distance between the two points corresponding to the CT_{half} values. Simple linear interpolation may be used to obtain this distance more accurately.
(6) Convert this to a distance (FWHM) by correcting for the pixel size.[24]
(7) Calculate the imaged slice width from:

$$\text{slice width} = \text{FWHM} \times \tan\theta$$

[23] To decrease the impact of image noise, it may be preferable to average over several rows.

[24] The FWHM option may, in some instances, also be inferred directly using calipers and the line profile tool.

where θ is the angle of inclination of the inclined plane to the tomographic plane and is a property of the phantom.[25]

Helical mode

(1) Place an ROI over the central portion of each reconstructed image corresponding to the position of the metal disc insert and measure the average CT number in it (see Fig. 26).
(2) Plot the measured values of the CT number versus image position (distance in the z axis direction) as in Fig. 27.
(3) Determine CT_{max} and CT_b.
(4) Determine the CT number corresponding to half the maximum height using the following relationship:

$$CT_{half} = (CT_{max} - CT_b) / 2 + CT_b$$

(5) Determine the FWHM by calculating the distance between the two points corresponding to the CT_{half} values. Simple linear interpolation may be used to obtain this distance more accurately. This distance represents the imaged slice width.

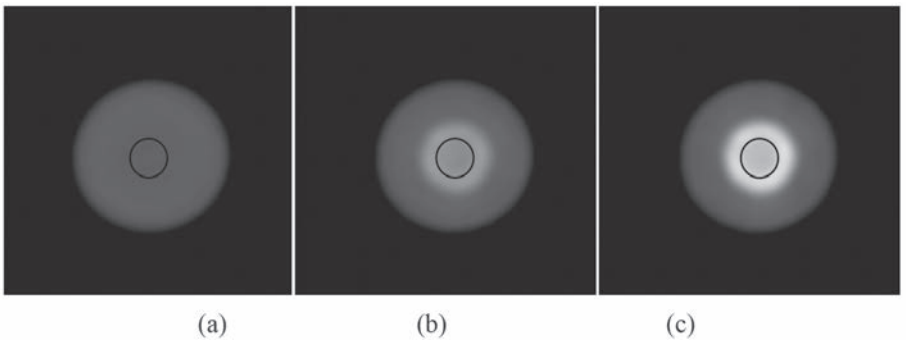

FIG. 26. Images of test object with ROIs shown drawn over metal 'delta function' insert. Insert is (a) not in the tomographic plane, (b) partially in the tomographic plane and (c) centred in the tomographic plane.

[25] The analysis is simplest if the angle, θ, is 45°. For thin slices, however, a smaller angle allows more pixel values to contribute to the profile and ensures a more accurate estimate of the imaged slice width. Where a test object utilizes two inclined planes, the slice width can be calculated from the average value. If the test object is aligned perfectly with respect to the scanner–patient couch axis, the two values should be identical. With any misalignment, the profiles will differ but the errors will cancel each other out.

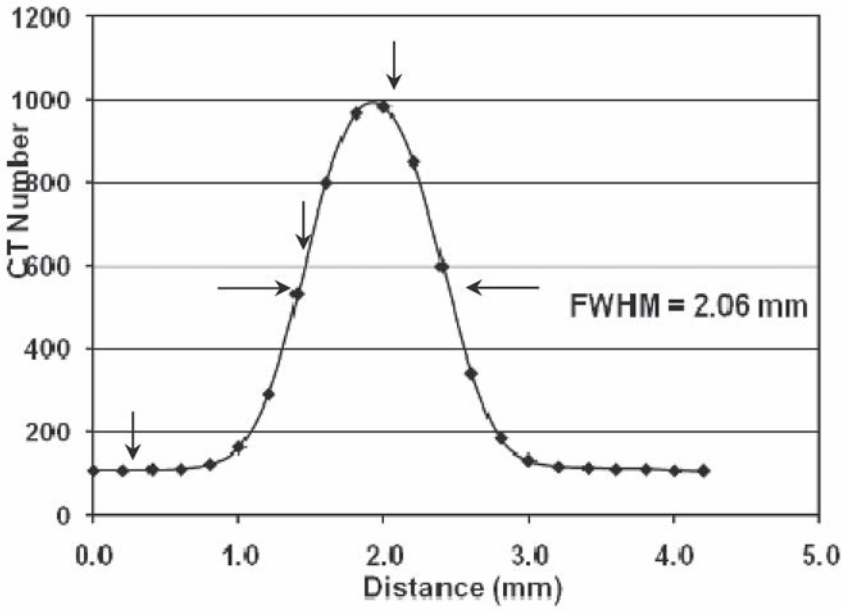

FIG. 27. Plot of pixel value in ROI placed over position of 'delta function' insert as a function of image slice position. The positions of the three images in Fig. 26 are indicated by vertical arrows. Nominal imaged slice width is 2 mm.

9.9.7. Interpretation of results

Nominal slice width (mm)	Acceptable
≤1	<nominal + 0.5 mm
>1 and ≤2	±50%
>2	±1 mm

(1) *At acceptance*, for MDCT assessed in axial mode, the measured imaged slice widths for contiguous images should agree with an acceptable tolerance of 20%.

(2) For extremely thin slices, especially if acquired in axial mode, the measured image thickness may be significantly thicker than the nominal thickness, for example, a nominal 1 mm thick slice may actually be 3 mm thick or more, because of limitations imposed by the thickness of the ramp. Verify the phantom specifications and apply appropriate corrections if needed.

9.9.8. Recommendations and corrective actions

If tolerances are not met, a service engineer should be contacted.

9.10. X RAY BEAM WIDTH

9.10.1. Introduction

The X ray beam width, also known as the radiation profile or irradiation slice width, is a measure of the collimated beam width along the z axis. It is generally defined and measured for axial scanning. In MDCT, the X ray beam width is larger than the total image width (in order to ensure uniformity of exposure to all detectors along the z axis), and this becomes more significant with thinner beam widths. This is known as over-beaming and the physicist should be aware of the extent of this phenomenon.

9.10.2. Scope

Objectives:

(a) To determine the accuracy of collimator settings;
(b) To determine the extent of over-beaming.

References: [4, 5].

Frequency: Acceptance and commissioning testing, and after changes.

9.10.3. Equipment, materials and instrumentation

A special X ray detector is required, either:

(a) A strip of therapy localization film[26], with a film backing plate and adhesive tape; or
(b) A strip of gafchromic film; or
(c) An array of thin thermoluminescent dosimeter chips loaded in a holder (not recommended for thin slice widths (details of this method are given in Ref. [4])).

[26] Dental film, industrial film or Polaroid film can also be used for this test.

9.10.4. Scan protocol

At acceptance, as specified by the purchase specifications. This usually includes all collimator settings.

At commissioning testing, it is recommended that all collimation settings be tested, including combinations of detector setting and slice width.

This is not recommended as an *annual test.*

9.10.5. Methodology

(1) Place film on a flat foam block to minimize scatter on the film.
(2) Raise the couch so that the film surface is on the isocentre of the scanner.
(3) Mark the isocentre of the film.
(4) Scan the film in axial mode for one collimation setting using a set kV and mAs as required to give the film a density that is clearly below the film maximum density.
(5) Repeat for all other selected scan protocols.
(6) Repeat steps (4) and (5), this time using the same kV but half the mAs value.

9.10.6. Analysis

(1) Plot the profile of the density of the exposed film using a film scanning device[27] for each collimation setting from steps (4) and (5) above.
(2) In a similar way, determine the maximum density for each scan made in step (6) above. Note that this density value corresponds to the 50% exposure level for the corresponding film processed above in step (1) and so defines the FWHM value for that film.
(3) Apply the density value for the FWHM to each of the beams plotted above in step (1) to determine the X ray beam width.
(4) One simple method to provide a direct measure of the over-beaming is to divide the nominal collimation beam width by the measured X ray beam width.[28]

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[27] Some transmission scanners used for personal computers can, with care, be used for this task.

[28] The more exactly defined term of geometric efficiency requires that the measured slice profile be converted from a film density profile to an air kerma profile through a suitable conversion process. The same applies to the use of thermoluminescent dosimeters to measure the profile. An added difficulty in this approach is that the conversion function will also vary with changes in the spectral make-up of the profile found outside the central section.

9.10.7. Interpretation of results

(1) *At acceptance*, the X ray beam width should be within manufacturers specifications.
(2) Compare measures of over-beaming for different collimation settings with data available for similar scanners.

9.10.8. Recommendations and corrective actions

(1) If tolerances are not met, a service engineer should be contacted.
(2) Direct measures of over-beaming are useful in the optimization process when determining the trade-off between dose and the advantages of using narrower beam collimation.
(3) An alternative relative indicator of the extent of over-beaming can be gained by comparing in-air pencil chamber dose measurements (C_{air}). This approach is further discussed in Section 9.6.

9.11. SPATIAL RESOLUTION

9.11.1. Introduction

The measurement of spatial resolution is important in acceptance and commissioning of a CT scanner, as resolution is a key parameter that can be optimized, along with image noise and dose. However, it is recognized that scanner resolution, as measured by routine QC tests, has not been seen to vary in the experience of many physicists working in the area. However, the use of specific phantoms and purpose written software does allow the possibility for this to be assessed in a reproducible way.

9.11.2. Scope

Objective: To ensure that the spatial resolution of a reconstructed image complies with manufacture standards.
Reference: [4].
Frequency: Acceptance testing and whenever the CT system is serviced, which might have an impact on the image quality.

9.11.3. Equipment, materials and instrumentation

See Section I.1.4.2, Fig. 33(a) or (b). For MTF evaluation of spatial resolution, software is required. This may be supplied either by the manufacturer or a third party.

9.11.4. Scan protocol

At acceptance, as specified by the purchase specifications.

At commissioning testing, it is recommended that the effect of reconstruction filters on spatial resolution be evaluated.

At annual tests, no annual testing of this parameter is considered necessary.

9.11.5. Methodology

(1) Centre the phantom in the tomographic plane.
(2) Align the test object parallel to the image plane.
(3) Note that this is essential, as any misalignment will cause the objects to become effectively larger (bars, rods, etc.) or more blurred (edge). The bead is the only test object that is not subject to alignment problems; this advantage is offset by a potentially lower signal to noise level.
(4) Select a scan relevant protocol.
(5) Scan the phantom.
(6) Repeat for additional scan protocols as specified above.

9.11.6. Analysis

(1) For resolution patterns (bars, rods, etc.), visual analysis is appropriate.
(2) For MTF evaluation, the image data need to be transferred to the same computer platform as the analysis software.

9.11.7. Interpretation of results

Within manufacturers' specifications.

9.11.8. Recommendations and corrective actions

If tolerances are not met, a service engineer should be contacted.

10. ADDITIONAL TESTS FOR RADIATION THERAPY

This section can be considered as a stand-alone section for radiotherapy testing and, therefore, may contain material that is duplicated elsewhere in this publication.

Image quality is of the utmost concern in both diagnostic and radiotherapy imaging and generally the same image quality is expected in both applications. Image quality in radiotherapy applications is of crucial importance in ensuring that tumour volumes are accurately delineated. The diagnostic image quality tests are equally applicable to radiotherapy applications and should be included in routine QA programmes.

Radiotherapy scanning places demands on the spatial and geometric accuracies of the CT scanner, couch top and external lasers that are often beyond the requirements of diagnostic imaging. Spatial and geometric distortions can lead to systematic radiotherapy patient treatment errors. A number of tests are needed to ensure correct installation and operation. More importantly, the diagnostic and radiotherapy physicists and service administration need to understand the unique needs for each speciality with respect to CT imaging, especially in shared CT settings. For example, Fig. 28 shows the image of a bath of water where the sloping water surface indicates that the whole image is rotated. As the image quality is acceptable, this would not be a problem in diagnostic applications; yet it could lead to potentially unacceptable systematic errors in radiotherapy treatments. Similarly, quantitative CT values, CT gantry and couch top alignment, CT image spatial orientation and integrity, and external laser alignment are all less important considerations in diagnostic applications, but are of crucial consideration in radiotherapy applications.

10.1. INTRODUCTION TO RADIOGRAPHER'S TESTS

A brief description of the methodology to be undertaken when performing the radiographer's QC tests for use in radiotherapy treatment planning is provided in this section (Tables 9 and 10). A listing of routine radiographer's tests needed for diagnostic examinations is given in Table 9 and detailed in Section 8.

The order in which tests are performed does not necessarily follow that in which they appear in this publication. The preferred order will depend on various

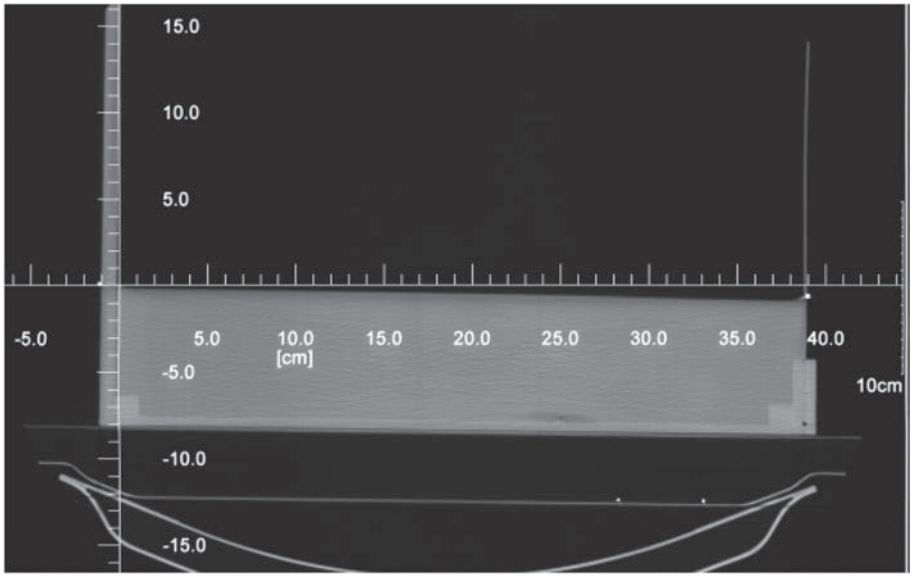

FIG. 28. CT image of a water bath demonstrating a sloping water surface. This indicates that the entire image is rotated.

factors relating to the CT facility, as well as the evaluator's preferences, always bearing in mind that there exist tests whose results affect the execution of others.

10.2. EXTERNAL CT POSITIONING LASERS

10.2.1. Introduction

External CT positioning laser accuracy tests show the congruence of scan localization light and the tomographic plane. Alignment lights on the scanner may be located within the gantry at the tomographic plane, or outside the gantry at a reference distance from the tomographic plane, or both. If both internal and external alignment lights are supplied, and are independently aligned, both should be tested (see Section 8.2). As indicated in Table 9, the tolerance for CT alignment lights is ±5 mm, which is outside acceptable limits for most radiotherapy applications, and if CT lights are used for this application, it is necessary to tighten the tolerance to ±2 mm. On some CT scanners, it may not be possible to achieve a sustainable alignment tolerance for CT lights of ±2 mm. This can be caused by several factors, for example, on some scanners the lights are mounted on the fibreglass cover, which may be vibrating or have poor

TABLE 9. RADIOGRAPHER'S QC TESTS (DIAGNOSTIC RADIOLOGY)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
8.2	CT alignment lights	E	Monthly	±5 mm	±1 mm
8.3	SPR accuracy	E	Monthly	±2 mm	±1 mm
8.4	CT number[b], image noise, image uniformity and image artefact	E	Monthly	CT number ±5 HU[c] Noise ±25% BB[d] Uniformity ±10 HU Minimal artefacts	CT number ±4 HU Noise ±10% BB Uniformity ±4 HU Minimal artefacts
8.5	Image display and printing	E	Monthly[e]	See text	

[a] E: Essential, basic requirement.
[b] At least one value should be checked daily to ensure correct dose calculation values for the treatment planning system.
[c] HU: Hounsfield units.
[d] %BB: Percentage of baseline value.
[e] Circumstances may require more frequent testing for some printers and monitors.

TABLE 10. RADIOGRAPHER'S QC TESTS (ADDITIONAL FOR RADIOTHERAPY APPLICATIONS)

Test number	Test	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
10.2	External CT positioning lasers	E	Daily[b]	±2 mm	±1 mm
10.3	Table top alignment and positional accuracy	E	Daily	±2 mm	±1 mm
10.4	CT number of multiple materials	E	Monthly	CT number ±5 HU[c]	CT number ±4 HU

[a] E: Essential, basic requirement.
[b] Daily, or at least on those days prior to using the system for treatment planning purposes.
[c] HU: Hounsfield units.

positioning reproducibility. The achievable accuracy of CT positioning lights for an individual scanner, and its impact, should be understood and taken into consideration for radiotherapy scanning.

External CT positioning lasers typically offer greater accuracy for patient marking and alignment and need to be regularly tested to a tight tolerance, as described below. As already discussed, wall lasers can be fixed or movable. Operationally, it is not necessary for the wall lasers to be movable, and the same quality and accuracy of treatment can be achieved with both laser arrangements. Movable lasers simplify the radiotherapy and patient marking process. However, if the wall lasers are fixed in the CT scanner room, a reference set of marks can be placed on the patient's skin and later adjusted on the treatment machine prior to the first treatment. In this set-up, the sagittal and coronal lasers are typically aligned with the origin of the tomographic plane and the axial laser is spaced at a specific distance from the tomographic plane. It can be convenient if this spacing is a distance that can be easily used in calculations (i.e. 500 mm or similar). For a movable laser arrangement, the sagittal and coronal lasers can be moved while the axial laser is typically fixed. The origin of sagittal and coronal lasers is still aligned with the tomographic plane origin and the axial laser is positioned some distance away. Alignment and QA for fixed and movable lasers are largely the same, with some additional tests for movable lasers. If internal CT gantry lights can be accurately and reproducibly aligned, then they can be used and tested along with the external CT positioning lasers.

10.2.2. Scope

Objective: To verify that wall mounted CT positioning lasers are properly aligned with the tomographic plane isocentre. The test procedure is the same for fixed and movable lasers.

References: [2, 3].

Frequency: Daily, or at least on those days prior to using the system for treatment planning purposes.

10.2.3. Equipment, materials and instrumentation

CT laser QC test tool [2] or similar test tool, as described in Section I.2.2.

10.2.4. Scan protocol

A thin slice width should be used, preferably 1 mm in the axial mode.

10.2.5. Methodology

(1) Ensure that the flat top is attached to the couch.
(2) Use the CT laser test tool or similar and establish that the internal light field is parallel to the tomographic plane. This involves aligning the markers on the device with the internal gantry laser. A scan is then taken to ensure that the lasers align with the radiation field and a couch position for this scan is recorded.
(3) The device is then aligned with the external CT positioning lasers and the couch position recorded. At this point, the external CT positioning lasers should bisect the markers on the device.
(4) Alternatively, if the internal lights are not of sufficient accuracy, the CT laser test tool can first be scanned and a couch position, which has the alignment between the centre of the test device and tomographic plane, identified. The test device is then aligned with the wall lasers and the longitudinal couch position noted.

10.2.6. Analysis

To ensure the internal gantry lights align with the tomographic plane, the test device holes should appear similar from left to right, as shown in Fig. 29, and should appear as large as possible. If there is any misalignment or rotation, the image of the holes in the test device will not be solid (they will appear dim, as in Fig. 29(b), or will not be symmetrical between the left and right pegs). At this time, the couch position at which the test device is aligned with the tomographic plane should be recorded.

Next, the couch is retracted so that the test device is aligned with the external CT positioning lasers and the couch coordinate is recorded.

10.2.7. Interpretation of results

The external CT positioning lasers should identify the tomographic plane within. The wall lasers should be spaced from the tomographic plane by the nominal distance of ±2 mm.

Test quantity	Acceptable	Achievable
External CT positioning lasers	±2 mm	±1 mm

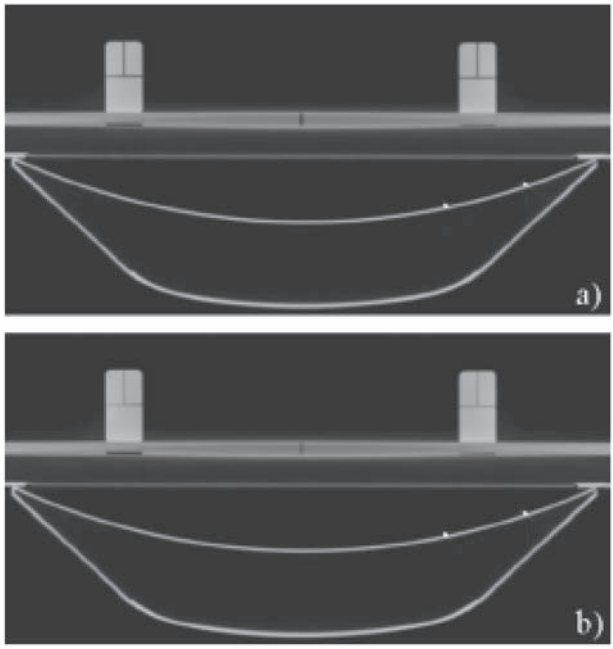

FIG. 29. CT laser QC device. (a) Alignment lights aligned with the tomographic plane and (b) centre of test device offset by 1 mm from the tomographic plane (from Ref. [2]).

10.2.8. Recommendations and corrective actions

If external CT positioning lasers are out of tolerance, no patients for radiotherapy treatment planning should be examined until the source of error has been determined and corrected. If a scanner is not equipped with external CT positioning lasers and only gantry lights are available and if these are out of tolerance, the scanner should not be used for radiotherapy scanning until the source of error has been determined and corrected.

10.3. COUCH TOP ALIGNMENT AND POSITIONAL ACCURACY

10.3.1. Introduction

In radiotherapy applications, it is most commonly required that the CT scanner be equipped with a flat couch top. The couch and the couch top should be orthogonal to the tomographic plane. This ensures that scanned patient geometry can be accurately reproduced on the treatment machine.

10.3.2. Scope

Objectives: To ensure that:

(a) The couch top is level and orthogonal with respect to the tomographic plane;
(b) Both couch and vertical motions, according to digital indicators, are accurate and reproducible.

Reference: [2].

Frequency: Daily, or at least on those days prior to using the system for treatment planning purposes, or any time after repositioning the couch top.

10.3.3. Equipment, materials and instrumentation

(1) Section I.2.2 CT laser QC test tool [2] or similar test tool;
(2) Ruler (1 m long);
(3) Spirit level;
(4) Weights of 70 kg and 140 kg to be distributed on the couch to simulate a patient.

10.3.4. Scan protocol

Single axial 1 mm thick CT slice.

10.3.5. Methodology

(a) Methodology for ensuring the couch top is level and orthogonal with respect to the tomographic plane:
    (1) Place on the couch top a 70 kg weight to simulate the weight of a patient.
    (2) Place the laser QC device on the extreme head end of the couch top (gantry side) carefully, so that the support edge is aligned with the couch edge.
    (3) Align the laser QC device with gantry alignment lights and spirit level.
    (4) Acquire a single 1 mm thick CT slice.
    (5) Place the laser QC device on the extreme foot end of the couch top carefully, so that the support edge is aligned with the couch edge, and again align it with the gantry alignment lights. For the success of this test, it is imperative that the laser QC device be aligned for both scans in the same position relative to the centre of the couch.
    (6) Acquire a single 1 mm thick CT slice.

(7) Measure the location of the centre hole in the laser QC device of both images using the scanner cursor tool. If the scanner is not equipped with a cursor tool, use image overlay graphics, which are reproducible between two scans, as the reference.
(8) Repeat steps (1)–(7) with a 140 kg weight.

(b) Methodology for ensuring that both couch and vertical motion according to digital indicators are accurate and reproducible:
(1) Place on the couch top a 70 kg weight to simulate the weight of a patient.
(2) Place a long ruler vertically on the couch top.
(3) Observe the light's position on the ruler as the couch is raised and lowered.
(4) Record all measurements made on the ruler, but ensure that the ruler is perpendicular to the couch top for all measurements.
(5) Repeat steps (1)–(4) with a 140 kg weight.

Details of these procedure are discussed in Appendix E of Ref. [2].

10.3.6. Analysis

(1) Compare the position of the horizontal holes in both pegs on the laser QC device in both images using the scanner cursor tool.
(2) For manual and computer controlled couch position accuracy and reproducibility, the measurements made on the ruler, using the light marker, are recorded.

10.3.7. Interpretation of results

Tolerances: The alignment is acceptable if the full lengths of the marker segments are visible with high contrasts in the image.

Test quantity	Acceptable	Achievable
Couch top alignment	±2 mm	±1 mm
Vertical positional accuracy	±2 mm	±1 mm

10.3.8. Recommendations and corrective actions

Measurements exceeding the suggested tolerances indicate a need for equipment repair by a qualified service engineer. Repairs and follow-up QC checks should be carried out prior to imaging patients for treatment planning purposes.

10.4. CT NUMBER OF MULTIPLE MATERIALS

10.4.1. Introduction

Dosimetric calculations, which account for differential dose distribution in various materials, rely on an accurate representation of the electron densities of organs shown in the CT image. An inaccurate determination of electron density can lead to erroneous radiotherapy dose calculations and, ultimately, to patient mistreatment.

10.4.2. Scope

Objective: To ensure that the CT number of a range of materials falls within the required values for the use of the scanner for radiotherapy planning.
References: [2, 4, 67, 139].
Frequency: Monthly.

10.4.3. Equipment, materials and instrumentation

Phantom containing inserts covering a range of different CT numbers (Section I.4.1), including an IAEA recommended phantom [67].

10.4.4. Scan protocol

The most frequently used scan protocol for radiotherapy patients. If multiple kV settings are used for radiotherapy patients, each kV used should be tested (see Appendix II).

10.4.5. Methodology

(1) Centre the phantom at the isocentre in the tomographic plane.
(2) Select the relevant scan protocol.

(3) Scan the phantom.
(4) Repeat for additional protocols as required.

10.4.6. Analysis

CT numbers are measured in a centrally placed circular ROI of appropriate diameter. In the ROI, the mean value of the CT number is given.

10.4.7. Interpretation of results

Test quantity	Acceptable	Achievable (IEC)
CT number	±20	±4 from baseline value

10.4.8. Recommendations and corrective actions

If measured values are out of tolerance, user calibrations of the scanner should be performed and measurements repeated. If the problem persists, corrective actions should be initiated. All out of tolerance results should be communicated to radiotherapy planning staff so that they can determine if any patients have been affected by equipment malfunction.

10.5. INTRODUCTION TO MEDICAL PHYSICIST'S TESTS

Table 11 lists the tests to be performed by the medical physicist for diagnostic purposes. Table 12 lists the additional tests to be performed by the medical physicist for radiotherapy purposes. Ideally, the tests from Table 12 should be performed by a medical physicist specializing in radiation oncology or, if performed by a medical physicist specializing in diagnostic radiology, should be done in consultation with the therapy physicist.

10.6. EXTERNAL CT POSITIONING LASERS

10.6.1. Introduction

The CT gantry alignment lights accuracy test shows the congruence of the scan localization light and the tomographic plane. Alignment lights on the scanner may be located within the gantry at the tomographic plane or outside the gantry at a reference distance from the tomographic plane, or both. If both

TABLE 11. MEDICAL PHYSICIST'S TESTS (DIAGNOSTIC RADIOLOGY)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
9.2	Visual inspection and review of programme	E	Acceptance and annually	All items must meet local standards	
9.3	CT alignment lights	E	Annually	±5 mm	±1 mm
9.4	SPR accuracy	E	Annually	±2 mm	±1 mm
9.5	kV and HVL	E	Acceptance and after changes	kV ±5% nominal HVL ≥ specified by radiation protection regulations and within the tolerances specified by the manufacturer	kV ±2% nominal HVL ≥ specified by radiation protection regulations and within the tolerances specified by the manufacturer
9.6	Radiation dose	E	Annually	±20% of manufacturers' specifications ±20% of baseline	±10% of baseline
9.7	Image noise	E	Commissioning and annually	±25% of baseline	±10% of baseline
9.7	Image uniformity	E	Commissioning and annually	±10 HU[b] for head and body	±4 HU for head and body
9.7	CT number accuracy	E	Commissioning and annually	±5 HU (water) ±10 HU (other material) compared to baseline values	±4 HU

TABLE 11. MEDICAL PHYSICIST'S TESTS (DIAGNOSTIC RADIOLOGY) (cont.)

Test number	Test name	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
9.8	Image display and printing	E	Commissioning and annually	See text	
9.9	Imaged slice width	E	Annually	+0.5 mm for <1 mm; ±50% for 1 to 2 mm; ±1 mm above 2 mm	
9.10	X ray beam width	D	Acceptance and after changes	Within manufacturers' specifications	
9.11	High contrast spatial resolution (MTF or modulation)	D	Acceptance and after changes	Within manufacturers' specifications	

[a] E: essential, basic requirement; D: desirable.
[b] HU: Hounsfield unit.

TABLE 12. MEDICAL PHYSICIST'S TESTS (RADIOTHERAPY)

Test number	Test	Priority[a]	Suggested frequency	Criteria	
				Acceptable	Achievable
10.6	CT positional lasers	E	Monthly	±2 mm	±1 mm
10.7	Couch top alignment and index accuracy	E	Annually	±2 mm couch positional accuracy	±1 mm
10.8	Gantry tilt	E	Annually	±1° from vertical	
10.9	CT number accuracy	E	Commissioning and annually	±20 HU[b] (all materials) compared with values specified by manufacturer	±4 HU

[a] E: essential, basic requirement.
[b] HU: Hounsfield unit.

internal and external alignment lights are supplied, and are independently aligned, both should be tested (see Section 8.2).

External CT positioning lasers are used for referencing the patient position and need to be regularly tested to a tight tolerance, as described below.

10.6.2. Scope

Objective: To verify that external CT positioning lasers are properly aligned with the CT tomographic plane isocentre. The methodology is the same for fixed and movable lasers.

References: [2, 3].

Frequency: Daily, or at least on those days prior to using the system for treatment planning purposes.

10.6.3. Equipment, materials and instrumentation

CT laser QC test tool [2] or similar test tool, as described in Section I.2.2.

10.6.4. Scan protocol

A thin slice width should be used, preferably 1 mm in axial mode.

10.6.5. Methodology

(1) Ensure that the flat couch top is attached to the couch.
(2) Use the CT laser test tool or similar device and establish that the internal light field is parallel to the tomographic plane. This involves aligning the markers on the device with the internal gantry laser. A scan is then taken to ensure that the lasers are aligned with the radiation field and a couch position for this scan is recorded.
(3) The device is then aligned with the external CT positioning lasers and the couch position recorded. At this point, the external CT positioning lasers should bisect the markers on the device.
(4) Alternatively, if the internal lights are not of sufficient accuracy, the CT laser test tool can first be scanned and a couch position, which has the alignment between the centre of the test device and tomographic plane, identified. The test device is then aligned with the wall lasers and the longitudinal couch position noted.

10.6.6. Analysis

To ensure that the internal gantry lights align with the tomographic plane, the test device holes should appear similar from left to right, as shown in Fig. 29, and should appear as large as possible. If there is any misalignment or rotation, the image of the holes in the test device will not be solid (they will appear dim, as in Fig. 29(b) or will not be symmetrical between the left and right pegs). At this time, the couch position at which the test device is aligned with the tomographic plane should be recorded.

Next, the couch is retracted so that the test device is aligned with the external CT positioning lasers and the couch coordinate is recorded. Alternatively, as described in Section 10.2.5.4, if the gantry CT lights are not used, the couch position at which the holes in the test device appear the largest and the couch position at which the test device is aligned with external lasers are recorded.

10.6.7. Interpretation of results

The CT positioning lights should identify the tomographic plane within. The wall lasers should be spaced from the tomographic plane by the nominal distance (±2 mm).

Test quantity	Acceptable	Achievable
External CT positioning lasers	±2 mm	±1 mm

10.6.8. Recommendations and corrective actions

If external CT positioning lasers are out of tolerance, no patients for radiotherapy treatment planning should be examined until the source of error has been determined and corrected. If a scanner is not equipped with wall lasers and only gantry mounted lasers/lights are available and if these are out of tolerance, the scanner should not be used for radiotherapy scanning until the source of error has been determined and corrected.

10.7. COUCH TOP ALIGNMENT, DEFLECTION AND POSITIONAL ACCURACY

10.7.1. Introduction

In radiotherapy applications, it is most commonly required that the CT scanner be equipped with a flat couch top. The couch and the couch top should be orthogonal to the tomographic plane. This ensures that scanned patient geometry can be accurately reproduced on the treatment machine.

10.7.2. Scope

Objectives: To ensure that the:

(a) Couch top is level and orthogonal with respect to the tomographic plane;
(b) Couch and the vertical and longitudinal motions, according to digital indicators, are accurate and reproducible;
(a) Couch indexing and positioning under scanner control is accurate.

References: [2, 3, 7].

Frequency: These tests must be performed by a medical physicist at the time of acceptance testing, as part of the end of warranty testing, annually, and whenever servicing of the CT system is carried out, which might impact couch top alignment and positional accuracy, including, but not limited to, system calibration and software changes or upgrades.

10.7.3. Equipment, materials and instrumentation

(1) CT laser QC test tool (see Section I.2.2);
(2) Ruler (1 m long);
(3) Weights of 70 kg and 140 kg to be distributed on the couch to simulate a patient.

10.7.4. Scan protocol

Single axial 1 mm thick CT slice.

10.7.5. Methodology

(a) Methodology for ensuring that the couch top is level and orthogonal with respect to the tomographic plane:

(1) Place on the couch top a 70 kg weight to simulate the weight of a patient.
(2) Place the alignment QC device on the extreme head end of the couch top (gantry side).
(3) Align the alignment QC device with the gantry alignment lights and the spirit level.
(4) Acquire a single 1 mm (or the thinnest available) thick, transaxial CT slice.
(5) Place the alignment QC device on the extreme foot end of the couch top and again align it with the gantry alignment lights. For the success of this test, it is essential that the alignment QC device be aligned for both scans in the same position relative to the centre of the couch.
(6) Acquire a single 1 mm (or the thinnest available) thick, transaxial CT slice.
(7) Measure the location of the centre hole in the alignment QC device of both images using the scanner cursor tool. If the scanner is not equipped with a cursor tool, use image overlay graphics, which are reproducible between two scans, as the reference.
(8) Repeat steps (1)–(7) with a 140 kg weight.

(b) Methodology for ensuring couch and vertical motion according to digital indicators are accurate and reproducible:
(1) Place a 70 kg weight on the couch top to simulate the weight of a patient.
(2) Place a long ruler vertically on the couch top.
(3) Observe the lights' position on the ruler as the couch is raised and lowered.
(4) Record all measurements made on the ruler.
(5) Repeat steps (1)–(4) with a 140 kg weight.
(6) Ensure that the ruler is perpendicular to the couch top for all measurements.

(c) Methodology for ensuring couch indexing and positioning under scanner control is accurate:
(1) Place a 70 kg weight on the couch top to simulate the weight of a patient.
(2) Place a long ruler along the couch top.
(3) Observe the lights' position on the ruler.
(4) Advance the couch under scanner control a known distance.
(5) Return the couch the same distance and note the position on the ruler.
(6) Repeat steps (1)–(5) with a 140 kg weight.

For both scans, the couch should be moved under scanner control. Details of these procedures using film are discussed in Appendix E of Ref. [2].

10.7.6. Analysis

(a) The positions of the horizontal holes in both pegs on the alignment QC device in both images should have the same coordinate, to within 2 mm, when measured by the scanner cursor tool.
(b) For the manual and computer controlled couch position accuracy and reproducibility, the measurements made on the ruler, using the light marker, are recorded.

Ensure that the:

(1) Couch top is level and orthogonal with respect to the tomographic plane;
(2) Couch and vertical motions, according to digital indicators, are accurate and reproducible;
(3) Couch indexing and positioning under scanner control is accurate.

10.7.7. Interpretation of results

Tolerances:

Test quantity	Acceptable	Achievable
Couch level and orthogonality	Visual assessment	
Digital indicator accuracy	±1 mm	
Couch positional accuracy	±2 mm	±1 mm

10.7.8. Recommendations and corrective actions

Measurements exceeding the suggested tolerances indicate the need for equipment repair by a qualified service engineer. If the CT images are to be used for radiation therapy treatment planning, the repairs and the follow-up QC checks should be carried out prior to imaging patients for treatment planning purposes.

10.8. GANTRY TILT

10.8.1. Introduction

Almost all CT scanners are capable of acquiring axial CT scans with a tilted gantry, though most MDCT systems do not permit helical scanning in these circumstances. Scanner tilt is generally not desirable in radiotherapy scanning and the use of tilted images could potentially lead to some unnecessary uncertainties in the treatment planning process. If the scanner is used only in a radiotherapy setting, the gantry is typically left in a vertical position all the time. In the case of shared use scanners, the gantry should be returned to the vertical position prior to therapy scanning and the gantry tilt indicator should correctly indicate the vertical position.

10.8.2. Scope

Objective: The digitally indicated angle of the CT scanner gantry in the vertical position should be accurate to within ±1°.
Reference: [2].
Frequency: Annually.

10.8.3. Equipment, materials and instrumentation

CT laser QC test tool [2] or similar test tool (see Section I.2.2).

10.8.4. Scan protocol

Not applicable.

10.8.5. Methodology

(1) Align the laser QA device with the gantry lasers and verify that the device is aligned with the side vertical gantry lasers over the full range of vertical couch travel by moving the couch up and down.
(2) Tilt the gantry in both directions and then return to the vertical position. The alignment of the laser QA device with vertical side gantry lasers should remain within 1°.
(3) Repeat the test by tilting the gantry in the opposite direction and returning it to the vertical position.

10.8.6. Analysis

After returning the gantry to the vertical position, as indicated by angle readout, the gantry lights/lasers should be in the same position relative to the laser QA device. It is important to move the couch up and down to ensure that the alignment is maintained over the full range of motion. If there are any discrepancies, it is expected that it would be the easiest to observe them at the extremes of the vertical couch travel range.

10.8.7. Interpretation of results

Tolerances: The alignment is acceptable if the relative position of the gantry lights/lasers remains in a constant position over the full range of vertical couch travel.

Test quantity	Acceptable	Achievable
Gantry alignment in vertical position	±1° from vertical	±1° from vertical

10.8.8. Recommendations and corrective actions

Measurements exceeding tolerance indicate the need for gantry tilt indicator recalibration or repair by a qualified service engineer. This test should be repeated after any such procedure to ensure correct operation. If this test fails, patient scans can still be performed as long as the correct vertical position is verified prior to each scan. This can be performed by following the procedure outlined in this section.

10.9. ELECTRON DENSITY CALIBRATION

10.9.1. Introduction

Reference [67] discusses in detail the importance for radiotherapy planning of an accurate relationship between CT numbers and electron density for a range of materials. Dosimetric calculations which account for differential dose distribution in various materials rely on an accurate representation of the electron densities of organs shown in the CT image. An inaccurate determination of electron density can lead to erroneous radiotherapy dose calculations and ultimately to patient mistreatment.

10.9.2. Scope

Objective: To establish the CT number and electron density relationship for the use of the scanner for radiotherapy planning.
References: [2, 4, 5, 67].
Frequency: This should be carried out at acceptance and annually, and also at the time of any software or hardware upgrades.

10.9.3. Equipment, materials and instrumentation

Phantom containing inserts covering a wide range of different CT numbers and with known electron density values (see Section I.4.1).

10.9.4. Scan protocol

Typical clinical parameters for radiotherapy (300 mAs, 1 cm reconstructed slice thickness axial mode) should be used, noting particularly that the correct reconstruction algorithm (filter or kernel) is used, and for all kV settings used clinically as discussed in Section 10.4.6.

10.9.5. Methodology

(1) Ensure that CT number and uniformity tests have been successfully completed at the time of testing.
(2) Centre the electron density phantom in the tomographic plane.
(3) Check centring with SPR.
(4) Select the relevant scan protocol.
(5) Scan the phantom.
(6) Place ROIs inside the relevant samples and record HU values.
(7) Repeat for additional protocols as required.
(8) Send acquired data to the treatment planning system for analysis under the treatment planning system QA procedures [67].

10.9.6. Analysis

The CT number is plotted against the known values of electron density for the materials supplied in the phantom.

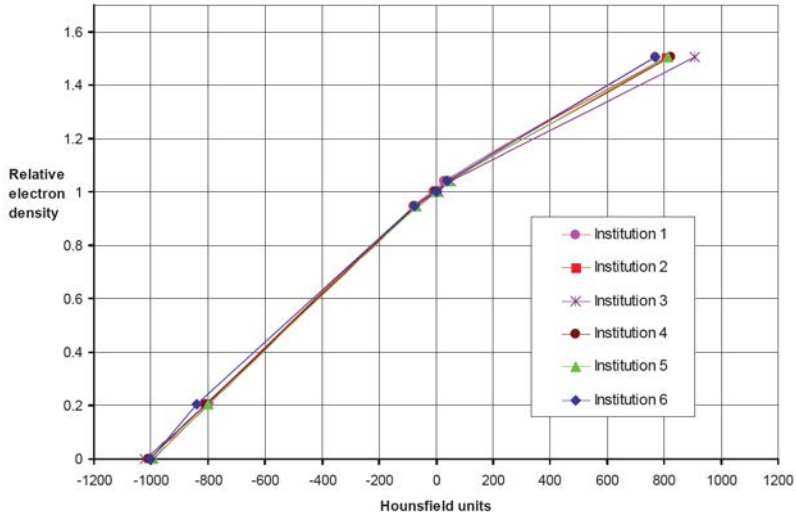

FIG. 30. CT calibration curves measured with the CIRS phantom at different hospitals [67].

10.9.7. Interpretation of results

Tolerances: ±5 HU (water), ±20 HU (other materials) compared with phantom manufacturer recommendations and baseline values established during system commissioning.

The plot of relative electron density against CT number is approximately linear, although in practice it deviates from the linear relationship in the high CT value region due to the greater proportion of photoelectric interactions (Fig. 30).

The actual CT values measured will differ from scanner to scanner, depending on differences in effective energy and as a function of scan energy as discussed in Section 10.4.6. However, the aim is to establish the relationship of CT number and electron density for a particular scanner, and this information is input into the radiotherapy treatment planning computer.

10.9.8. Recommendations and corrective actions

If measured values are out of tolerance, user calibrations of the scanner should be performed and measurements repeated. If a problem persists, corrective actions should be initiated. All out of tolerance results should be communicated to the radiotherapy planning staff so that they can determine if any patients have been affected by equipment malfunction.

Appendix I

PHANTOMS AND EQUIPMENT

CT phantoms and equipment can be organized into three general categories: (i) image performance phantoms, (ii) geometric phantoms and (iii) quantitative/dosimetry phantoms and instrumentation. Many of these phantoms are available from the scanner suppliers and are 'packaged' in the purchase agreement. In other cases, it may be necessary to use commercial and 'home-grown' solutions. These solutions vary in complexity, cost and availability. While discussing all available designs and features is beyond the scope of this publication, several general descriptions can be made.

I.1. IMAGE PERFORMANCE PHANTOMS

I.1.1. Noise and uniformity phantom (general)

Noise and homogeneity section: Homogeneous phantom or phantom section that may contain an outer rim of plastic (preferably PMMA). This will be the case in phantoms containing liquid water. Alternatively, these phantoms may be constructed from uniform plastic (Fig. 31). These phantoms typically come in

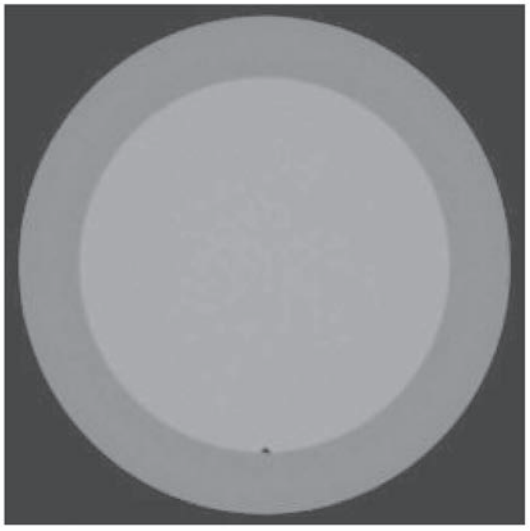

FIG. 31. Noise and uniformity phantom consisting of uniform plastic core and surrounding plastic ring.

two diameters: body and head. Phantoms are identical in design and only differ in their size in simulating the body or head.

I.1.2. Noise and uniformity phantom (head)

- Homogeneous phantom or phantom section simulating the head.
- Homogeneous in attenuation or phantom section and may contain an outer rim of plastic (preferably PMMA). This will be the case in phantoms containing liquid water.
- Outside diameter of 16–20 cm and a total attenuation equivalent to at least 16 cm of water should be used to simulate the patient's head [7].
- Length of the phantom or water section in the phantom should be substantially longer than the maximum total collimation, this being especially important for MDCT.

I.1.3. Noise and uniformity phantom (body)

- Homogeneous phantom or phantom section simulating the body.
- As in Section I.1.2, but with an outside diameter of 30–35 cm and a total attenuation equivalent to at least 30 cm of water used to simulate the patient's body [7].

I.1.4. Low and high contrast spatial resolution phantoms

While this publication does not discuss in detail low and high contrast tests, a brief description of phantoms used for evaluating these image performance parameters is given here.

I.1.4.1. Low contrast (contrast resolution) phantom

Low contrast phantoms consist of relatively large cylindrical test objects, of varying sizes and contrast, which are imbedded in a uniform background which differs only slightly in density from the test objects. The phantom imaged in Fig. 32 contains three sets of cylindrical rods of varying diameters and contrast levels to measure low contrast performance. The rod diameters at each contrast level are 2, 3, 4, 5, 6, 7, 8, 9, 10 and 15, with nominal contrast levels of 0.3, 0.5 and 1%. These phantoms are usually purchased rather than developed in-house. This publication advocates the use of noise values as a sufficient test of scanner performance in relation to low contrast detectability, as is consistent with the performance testing of other digital imaging equipment.

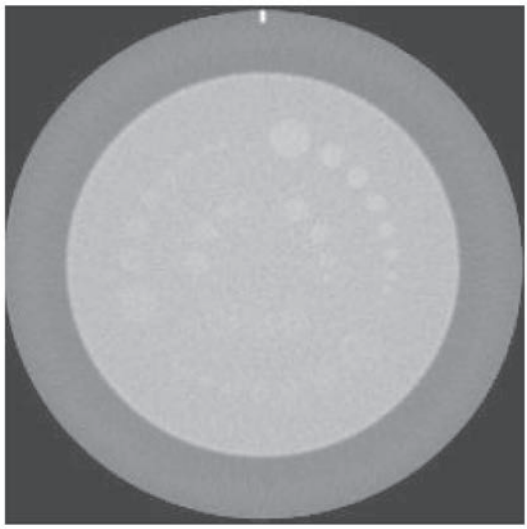

FIG. 32. Low contrast phantom.

I.1.4.2. High contrast (spatial resolution) phantom

High resolution measurements are performed with objects which have a high contrast (contrast difference of 12% or greater) with respect to uniform background and which are usually supplied by the manufacturer or are available from commercial manufacturers. High contrast resolution is most commonly measured using either a resolution pattern (line pair phantom with a range of spatial frequencies) or by the MTF method. The line pair phantom is shown in Fig. 33(a), where the line pair pattern ranges in frequency from 1 lp/cm to 21 lp/cm. Increasingly, however, this is not able to demonstrate the high spatial resolutions now available with modern scanners. The use of PMMA test objects in water is also problematic due to long term absorption of water in the PMMA. Figure 33(b) shows an MTF phantom with a high density tungsten carbide bead which is used to create an impulse, or point source, from which the MTF can be calculated.[29] The high contrast wire or bead needs to have a cross-sectional

[29] Alternatively, an edge response phantom can be developed. To measure an edge spread function, a phantom containing the block of high density material and having a flat face needs to be used. Suitable dimensions would be approximately 80 mm × 40 mm in the image plane and a thickness which is at least three times the nominal irradiated slice width (more for helical scanning or for MDCT scanners). Perspex can be used for low resolution scans. However, a higher density material for the higher resolution scans, such as polytetrafluoroethylene (PTFE), is more suitable [4].

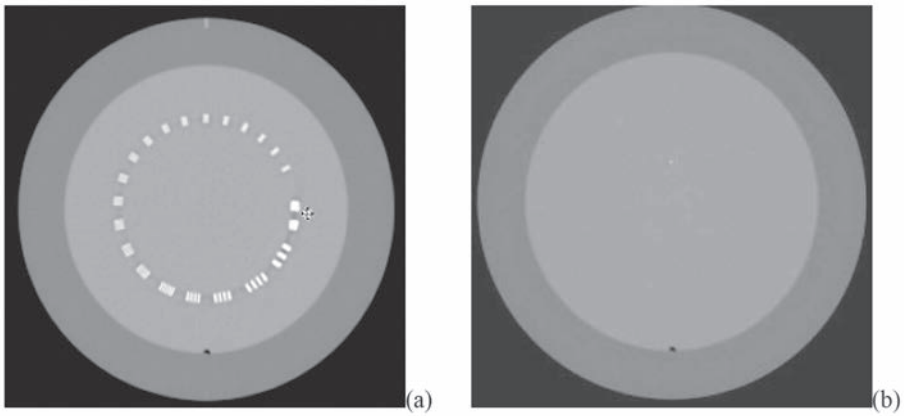

FIG. 33. (a) Line pair high contrast phantom, (b) MTF based high contrast phantom.

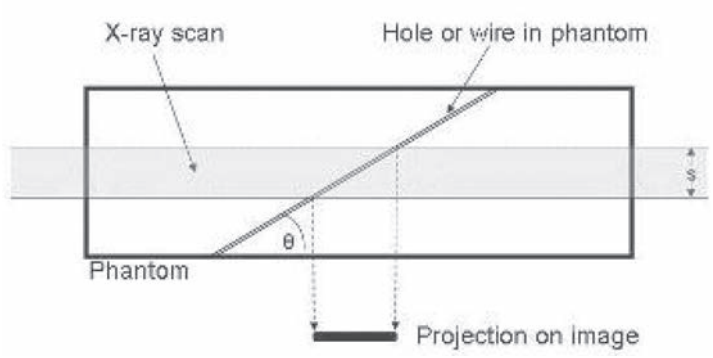

FIG. 34. View of slice width phantom with superimposed CT slice projection and resulting test object projection on image.

diameter of the order of, or preferably less than, the resolution that is being tested. For all phantoms used to evaluate the scanner, MTF appropriate software needs to be available to complete the analysis.

I.1.5. Imaged slice width phantom

Imaged slice width can be measured in either the axial or helical mode.

In the axial mode, the test object typically contains one, or preferably two, thin metal inclined planes[30], as seen in Fig. 34.

[30] A ramp with discrete beads, discs or wires may also be used.

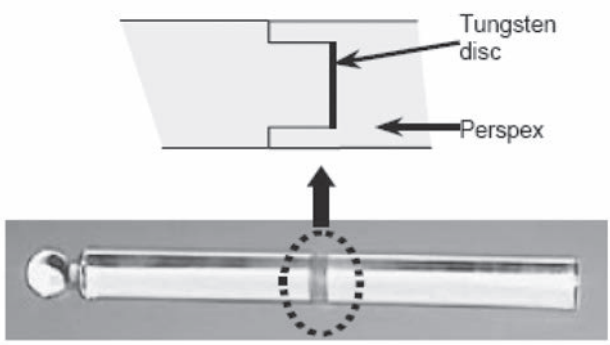

FIG. 35. A typical helical z sensitivity phantom (courtesy: ImPACT) [140].

For helical acquisition, the phantom contains a thin (submillimetre thick) metal plate (Fig. 35), or a submillimetre diameter air hole embedded in a uniform background cylinder (usually PMMA). The thickness of the metal plate or size of the air hole must be less than the smallest nominal imaged slice width that will be measured. Such phantoms can commonly be manufactured locally, using materials such as the lead foil found in dental film packs.

I.2. GEOMETRIC ACCURACY PHANTOMS

I.2.1. SPR accuracy test tool

A suitable test tool can be easily fabricated from a PMMA slab at least 25 cm, and preferably 50 cm or more, long. It can be shaped like a ruler and should have accurately placed markers at a set distance apart (e.g. 50 cm). Another alternative is to use a phantom or material block of a precisely known length. In this case, the SPR distance can be checked by comparing with the distance between the ends of the phantom.

I.2.2. CT laser QC test tool or similar test tool

This device, as shown in the Fig. 36, consists of a Lucite base and two Lucite pegs mounted on the base. The pegs are 5 cm high, 2.8 cm wide and 25 cm apart. Vertical and horizontal holes 1 mm in diameter are drilled through the centre of each peg, forming a cross inside the peg.

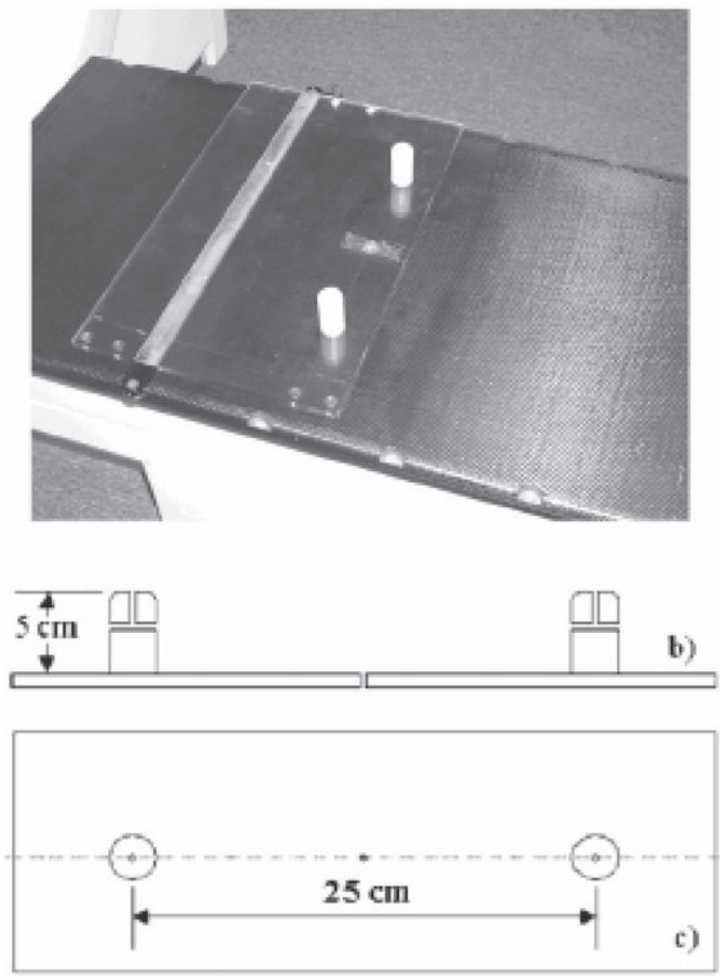

FIG. 36. CT laser QC test tool. (a) CT laser QC device attached to the couch top, (b) side elevation of CT laser device through the centre of the pegs showing holes drilled inside the pegs and (c) plan view of the device (adapted from Ref. [2]).

I.3. DOSIMETRY PHANTOMS

I.3.1. Standard CT dosimetry phantoms and equipment

Phantoms: The standard head and body CT dosimetry phantoms (Fig. 37) are constructed of PMMA and each is a right circular cylinder of at least 140 mm in length. The head and body phantoms have diameters of 160 mm and 320 mm, respectively, and may be separate or combined as in Fig. 37. Note that the body phantom cross-section is larger than a typical adult. Each phantom has a central bore and at least four bores at the periphery to accommodate the ionization chamber as shown in Fig. 38. All the holes are filled with plugs, which may be individually removed to facilitate measurements.

Dosimetry equipment: The CT air kerma indices $C_{a,100}$ and C_w ($CTDI_{100}$ and $CTDI_w$) are measured with a calibrated pencil detector compliant with IEC standards [141], which typically comprises an ionization chamber specially designed for CT (Fig. 39) and an electrometer.

Active semiconductor detectors and passive solid state detectors (thermoluminescence and optically stimulated luminescence) have also been used for CT dosimetry, but their use is limited [26].

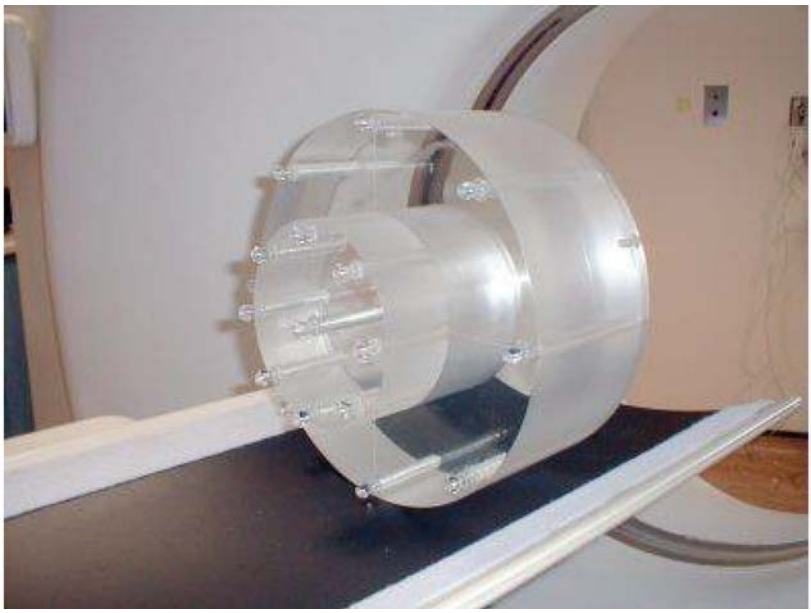

FIG. 37. Photograph of typical standard head and body CT dosimetry phantoms. The head phantom is the innermost cylinder (which has been partially removed for clarity) and when surrounded by a PMMA annulus, the body phantom is formed (from Ref. [26]).

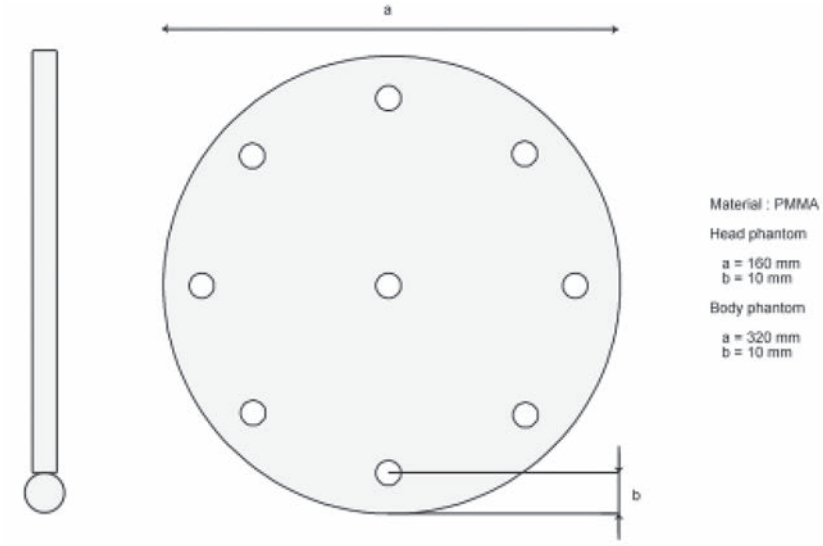

FIG. 38. Cross-section through the standard phantom (head and body dimensions given). The small circles indicate the positions of the bores for inserting a CT ionization chamber. A PMMA plug is shown on the left (not to scale) (from Ref. [26]).

FIG. 39. Pencil ionization chamber used for dosimetry of CT examinations. The active volume of the chamber is typically 3 cm^3 and the active length is 100 mm [26].

Paediatric phantoms: Currently, there is no internationally agreed phantom or set of phantoms to represent the various sizes for non-adult patients. Typically, the standard head CT dosimetry phantom (16 cm diameter) is used to provide C_{VOL} values for paediatrics. However, the phantom size should always be specified and care should be taken that the correct filters (head or body) are applied.

I.4. QUANTITATIVE PHANTOMS

I.4.1. CT number phantom

There are numerous phantoms which are designed to measure CT numbers or HU. These phantoms are typically cylindrical in shape and contain several cylindrical test objects of different composition. Preferably, the electron density is known for each material. Figure 40 shows a phantom which was recommended in a previous IAEA publication [67]. This phantom is specifically designed for radiotherapy applications and a number of alternatives are also available for primarily diagnostic work. Figure 41 shows one such phantom. If a commercial phantom is not available, a CT phantom may be constructed locally, provided due care is taken.

FIG. 40. Thorax phantom (CIRS Model 002LFC) [67].

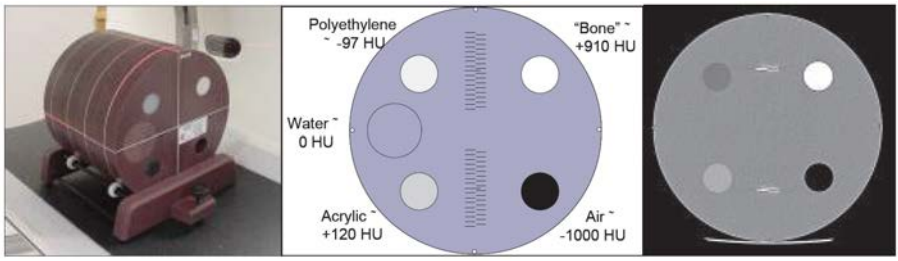

FIG. 41. CT number regions in RMI ACR accreditation phantom.

Appendix II

RADIOTHERAPY CT SCAN PROTOCOL PARAMETERS

II.1. INTRODUCTION

In addition to anatomical information, radiotherapy applications rely on CT studies to provide accurate information about patient geometry and also quantitative CT values which contain information about patient tissue composition. Radiotherapy CT scan protocols commonly require an image slice thickness of 3 mm, longer scan volumes than are typically used in diagnostic scanning and scans that are acquired in rapid fashion. Therefore, CT scan protocols for radiotherapy applications often require modification for the common diagnostic protocols used for the same scan site. This section provides some elementary guidance and explanations for radiotherapy scan protocol parameter selection.

The appropriate settings needed for various parameters for single slice and multislice scanners are also addressed. A number of CT scanners routinely come from the manufacturer preloaded with basic radiotherapy protocols and these are generally a good starting point for scan protocol development.

II.2. PROTOCOL PARAMETERS

Tube potential (kV): As discussed in Section 10.9, the relationship between CT number and electron density that is stored in radiotherapy treatment planning system computers is unique for each available CT scanner tube voltage setting on the scanner. In order to avoid radiotherapy treatment planning errors and to simplify the process, most radiotherapy facilities rely on a single kV setting for all their examination scans. This ensures that the correct CT number to electron density conversion file is always used. Tube voltages of 120 or 130 kV should produce satisfactory images and information in almost all patients. These values are equally applicable to single slice and multislice CT scanners. The following points are important:

- It is imperative that all radiotherapy patients be scanned using the predetermined kV setting as given in the scan protocol and this setting should not be altered without a specific communication to radiotherapy planning staff. If a facility chooses to vary the tube voltage setting among patients, in order to optimize imaging, this must be clearly communicated

by the CT radiographers to radiotherapy staff to ensure that the correct conversion table is used in the treatment planning system for each patient.

- In this discussion of CT kV settings it is assumed that the radiotherapy facility is performing heterogeneity corrected dose calculations. If the facility is exclusively using homogenous dose calculations, then kV settings can be used as needed. It is imperative to contact the radiotherapy physicist to determine which dose calculation formalism (heterogeneous or homogenous) is used in the clinic.

Tube current (mA): Soft tissue contrast can be of critical importance for tumour delineation in radiotherapy applications and higher than usual mA settings may be desirable. As an increase in mA is directly proportional to an increase in dose, radiotherapy physicians should determine if such an increase in mA is justified. In most instances, the additional dose from a CT scan is a very minor fraction of the dose that will be delivered from radiotherapy treatment and it is commonly considered that the benefit of improved images outweighs the risks to patient from higher dose.

As discussed in Section 4.3 and in this appendix, radiotherapy examinations often require 100 or more images in a single study. The large number of images result in increased demand on X ray tube heat loading and some CT scanners may not be able to accommodate such scans without modification of protocols (this is mainly an issue with single slice scanners, as MDCT scanners employ a more efficient utilization of X rays). One of these modifications is a reduction of the mA settings from commonly used values. The lower mA will result in a lower tube loading. However, this will also reduce image quality and it is in direct contradiction to the above paragraph. Radiotherapy physicians and/or physicists should be consulted to evaluate the options for mA settings.

It is possible that some large bore CT scanners require an increase in mA setting to produce the same image quality as conventional bore scanners. This is due to increased separation between the X ray tube and detectors. The manufacturer should be contacted to evaluate this issue and determine if adjustment in technique is needed, based on scanner geometry.

Scan time: The scan time setting is commonly selected so that the chance of patient motion during the scan is minimized while allowing the entire volume to be scanned without running into heat storage problems. For most single slice scanners, scan time will be set to 1–2 seconds per rotation. For MDCT scanners, this is typically less than one second, depending on scanner capabilities.

Slice thickness and spacing: Typically, the slice thickness used for radiotherapy applications is 3 mm with continuous coverage of the patient. This provides a very good balance between tumour and normal structure delineation accuracy and the generated number of images. Thicker slices may lead to the

inadequate definition of tumours, while thinner slices may result in unnecessarily large data sets. At a slice spacing of 3 mm, common radiotherapy scans will result in 100–200 images. The majority of treatment planning systems should be able to accommodate such amounts of data with little difficulty. However, the radiotherapy physicist should be consulted to confirm data storage capacity.

In single slice CT scanners, it may not be possible to scan the entire volume with 3 mm thick slices without reaching the X ray tube heat limit. In such situations, it is acceptable to use a 3 mm slice thickness through the tumour volume and 5 mm above and below this volume with the use of thicker slice settings (5 or 8 mm) towards the two ends of the scan volume. As MDCT scanners typically do not have tube heat storage issues, a single spacing can be used for the entire scan.

Pitch: The pitch setting for single slice scanners is commonly set between 1.5 and 2.0 in order to allow full volume image acquisition without reaching the X ray tube heat storage limit. Pitch settings greater than 2.0 may lead to unacceptable image quality and should be evaluated prior to use. For multidetector scanners, the pitch setting is less critical and can be set according to manufacturer recommendations.

Reconstructed display FOV: Radiotherapy applications most commonly require that the full extent of the patient be included in axial images. This is necessary for accurate dose deposition calculations. Therefore, since the FOV must be large enough to accommodate the entire patient and in virtually all applications except paediatric scanning, it is desirable to set the FOV to the largest setting available. For most scanners, this will be 48 or 50 cm. This requirement is the same for single slice and multislice scanners.

Scan limits: In diagnostic applications, it is typically acceptable to acquire a scan only through the volume of interest (i.e. through the tumour and the volume in the immediate vicinity) and to limit the extent of the scan to as short a time as necessary. In radiotherapy applications, it is often required to include the entire organs in the scan in order to evaluate treatment effects on organ function. For example, if a patient has a 3 cm long lung lesion, the entire lung volume would commonly be included in the scan so that the effects of radiotherapy dose on lung function can be determined. Therefore, scan limits for radiotherapy applications will be different from diagnostic ones and the radiotherapy physician should be contacted to determine these. It is possible that scan limits are predetermined for individual scan sites and outlined in the institution's policies and procedures or, alternatively, the physician can provide them with each scan request. This requirement is the same for single slice and multislice scanners.

Image reconstruction: Image reconstruction algorithms and other parameters for radiotherapy and diagnostic applications are typically the same and no modifications are needed. It is possible that some options may result in better soft tissue contrast and these possibilities should be explored. This requirement is the same for single slice and multislice scanners.

Appendix III

CONFIGURING EXTERNAL CT POSITIONING LASERS FOR RADIOTHERAPY APPLICATIONS

III.1. INTRODUCTION

External CT positioning lasers can significantly improve the accuracy, value and efficiency of CT imaging for radiotherapy applications. The lasers in the CT scanner are needed to align and mark a patient in a position which subsequently can be reproduced for radiotherapy treatments, as shown in Fig. 42.

In the simplest form, this marking and patient positioning can be performed with gantry mounted lights/lasers. Unfortunately, gantry lights are often not very stable and/or accurate and, additionally, it can be very difficult to reach inside the gantry to mark the patient. Generally, it is preferred to have lasers which are

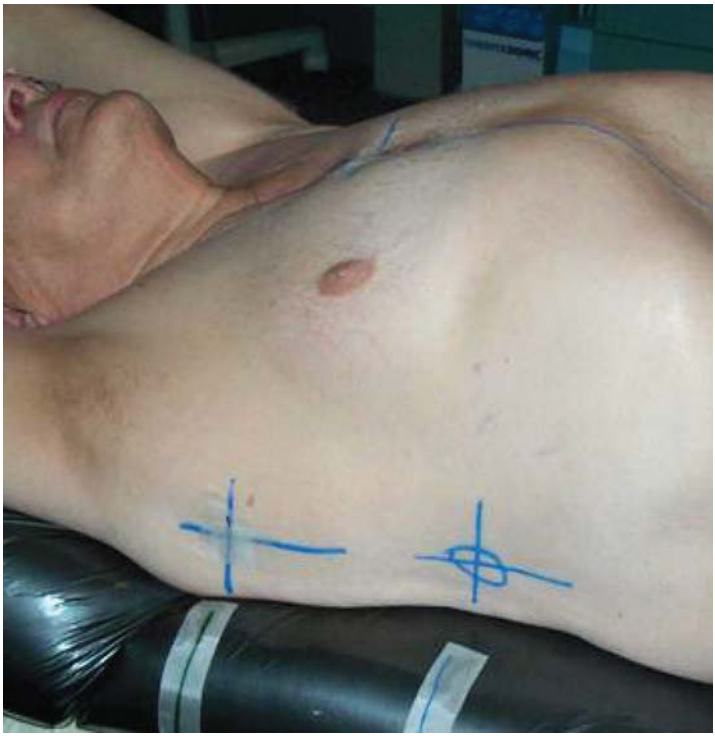

FIG. 42. Example of skin marks that are placed on a patient's anterior and sides during a radiotherapy CT scan.

mounted outside the gantry. External lasers can be wall mounted if walls are sufficiently stable (usually less than 1 mm movement is expected) or on a metal arch frame or on posts. External CT positioning lasers consist of overhead (sagittal), side-horizontal (coronal) and side-vertical (axial) lasers. Side lasers are mounted on both sides of the scanner.

III.2. CONFIGURATION

Wall or arch mounted lasers are available in four configurations (listed in order of increasing complexity and cost):

(i) *Fixed lateral cross hair and fixed overhead sagittal line lasers*: In this configuration, all three lasers are fixed and cannot be moved. This is a very reasonable configuration where resources are limited and radiotherapy patients can be remarked prior to treatment while on the treatment couch according to the radiotherapy plan. With well-designed procedures, this technique provides acceptable results, but with less efficiency and perhaps an increased chance for error. Therefore, it is essential that communication channels and procedures be established to ensure that patients are correctly remarked while on the treatment machine.

(ii) *Fixed lateral cross hair and movable overhead sagittal line lasers*: In this configuration, lateral lasers are fixed while the overhead laser is movable (see Fig. 7(b)). This configuration typically offers the optimal compromise between complexity and cost and clinical utility and efficiency. This configuration provides great clinical flexibility and patients can efficiently and accurately progress from CT scanner to treatment. The main advantage of this option is the capability to move the overhead laser in either the left or right direction, which allows patient marking away from the midline. This cannot be achieved with couch movement, as CT couches do not move in this direction.

(iii) *Fixed side-vertical, movable side-horizontal and movable overhead sagittal line lasers*: This configuration is shown in Fig. 7(a). This is a slightly more advanced version of the configuration shown in Fig. 7(b). Generally, this option can provide somewhat greater operational efficiency but it does not significantly improve clinical utility or accuracy. Additionally, in some systems, the couch cannot be zeroed in the vertical axis, requiring the operator to add or subtract the shift value from the couch position, which can be confusing, especially since the coordinate system for IEC 61217 [142] may not correlate to the coordinate system of the CT.

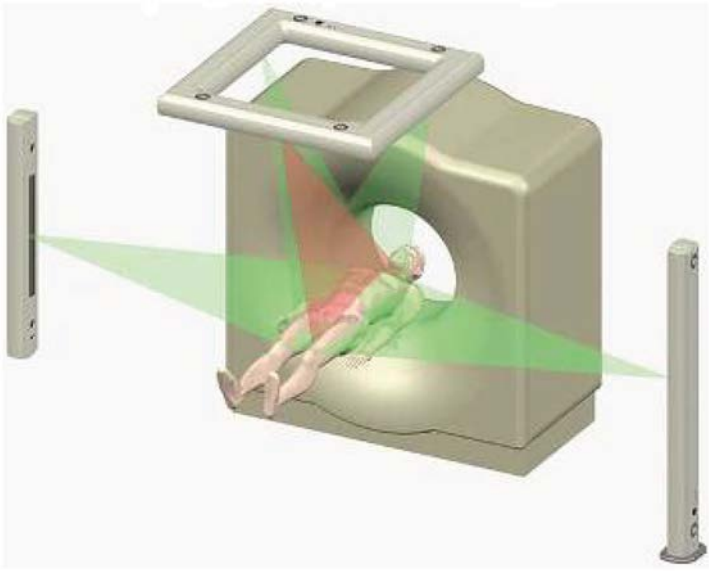

FIG. 43. An advanced laser system for patient positioning and marking (courtesy: LAP Laser Systems for image and comments).

(iv) *Movable side-horizontal line, movable side-vertical line and movable overhead sagittal line laser*: This configuration is shown in Fig. 43. In complexity, this is the most advanced external laser configuration. While system flexibility and efficiency are exceptional, the increased cost may be hard to justify in many clinical settings.

III.3. GEOMETRY

(a) *Orientation*: When configuring external CT scanner lasers, the laser location and orientation is referenced to the tomographic plane. The lasers need to be installed orthogonally or parallel (depending on their orientation) to the tomographic plane. As the laser projection can be 40 cm or more in length, it is important to ensure that the lasers are parallel/orthogonal over their entire projection.

(b) *Origin/isocentre*: As the lasers are oriented with respect to the tomographic plane, the laser origin is also referenced to the tomographic plane. The sagittal and horizontal laser origins coincide with the image centre (typically x and y coordinates); the vertical laser is spaced some predetermined distance away from the tomographic plane. This distance

depends on the scanner installation and the room layout, but it is typically 500–600 mm from the tomographic plane. It may be important that this distance is set to a simple value as it assists in manual calculations when necessary, during the marking of patients. This is especially important for CT simulation software with limited available capability.

(c) *Projection length*: In many radiotherapy applications, several alignment marks will be placed along any laser projection/patient side. Multiple alignment marks greatly improve the reproducibility and accuracy of patient positioning for radiation treatment. Therefore, laser projection at the level of the patient should be at least 40 cm in length and longer projections are preferable.

III.4. INSTALLATION

Any installation of external/wall CT lasers should start with localization of the tomographic plane. The tomographic plane can be localized using the CT laser test tool (Fig. 36) or similar device. The procedure starts by scanning the QA device and ensuring that it is parallel with the tomographic plane in both x and y directions. This may require several scans to first align the QA tool in the x direction. After the tool has been aligned in the x direction, two scans with different couch vertical positions should be acquired to ensure that the tool is aligned in the y direction as well. Once it is verified that the QA tool is aligned with the tomographic plane, the origin of the tool is then aligned with the centre of the tomographic plane by adjusting the couch scan height. Once the QA device has been aligned with the image origin, the couch is retracted by an amount equal to the desired separation between the tomographic plane and the vertical side lasers. This position will be used to install and align all lasers. Installation will proceed, depending on laser configuration and facility design. Once the installation is complete, it should be verified that lasers are orthogonal/parallel along the entire projection. This is accomplished by moving the couch in/out of the gantry and also up/down while observing the laser projection on the laser QA device [2].

The above procedure relies on proper couch alignment with the tomographic plane. This can be verified by the method described in Section 10.7. More efficient methods for accurate laser installation and alignment are available, but the above outlined procedure can be implemented in virtually all clinical settings with minimal resources.

Appendix IV

AVAILABLE DOSIMETRY SOFTWARE

Freeware

CT Dose 1.0.1
http://www.mta.au.dk/ctdose/index.htm (2008-04-26)
Dose calculations for CT-exams (last update 2003-11-11)

Omni mAs
http://omnimas.arwen.se/ (2008-04-26)
Calculation of mAs settings for different patient diameters
Effective dose via conversion factors for DLP

D2ED (Palm)
http://www.mh-hannover.de/1604.html
Estimation of patient exposure using conversion factors
(DAP, CTDI and DLP)

QuickDose (Windows Mobile 6)
http://www.mh-hannover.de/1604.html
Estimation of patient exposure using conversion factors
(DAP, CTDI and DLP)

Commercial/shareware

CT-Expo V 2.1 (Shareware)

http://www.mh-hannover.de/1604.html (2008-04-26)

ImpactDose

http://www.vamp-gmbh.de/software/impactdose.php (2008-04-28)

CTDosimetry.xls v1.0

http://www.impactscan.org/ctdosimetry.htm#CTDoseDownload (2009-09-15)

free but requires NRPB-SR250: Normalised Organ Doses for X-Ray Computed Tomography Calculated Using Monte Carlo Techniques

http://www.hpa.org.uk/webw/HPAweb&HPAwebStandard/HPAweb_C/1957337533330?p=1153846674387

P-Dose

http://www.qualitycontrol.org/index.php?page=modulo_p_dose_ct_en (2008-04-28)

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Annex

VISUAL INSPECTION AND PROGRAMME REVIEW CHECKLIST

Item to be evaluated	Date				
Scan protocol book:					
Located at operator's console					
Protocols up to date					
Automatic exposure control, if available, for all protocols					
Technique chart for all manual scans with mAs varying with patient size					
Techniques for chest lower than abdomen scans					
Adult protocols available and optimized					
Paediatric protocols available and optimized					
Window between operator's room and scanner room provides clear view of patient (no clutter on the window)					
Ensure image headers are correct, including patient identification, date and time					
Written radiation safety procedures up to date					
Staff (including radiologists) understand and follow radiation safety procedures					
Scanner room:					
Door closes properly					
Room clean					
Lead aprons available					
Portable barriers available					

Item to be evaluated	Date				
Quality control logs:					
Properly completed					
Up to date					
Required corrective action noted and requested					
Quality control tests performed after corrective action					
Emergency equipment available and complete					
Emergency phone numbers posted and clearly visible beside telephones					
Medical physicist should add items specific to the facility					

Note: P = pass, F = fail.

GLOSSARY

3-D. Imaging techniques used to display images of a three dimensional volume on a flat (2-D) display, such as a computer screen. Should be distinguished from true 3-D displays, such as virtual reality or stereoscopic systems.

archive. Storage of large amounts of idle data. Accuracy of data recovery is important, but as the data are no longer active, speed of reading is not as important. Can be stored on various types of device or media, such as tape or optical disc.

artefact. Any structure or pattern visible in the image that is not part of the object being imaged.

automatic exposure control (AEC). A device which automatically determines and provides the exposure needed to produce a preselected image quality by sampling the X ray intensity at the image receptor.

beam hardening. The process of filtration of a polychromatic beam by the preferential absorption of lower energy photons in tissue, with a subsequent increase in effective energy.

coefficient of variation (COV). COV = $\sigma_x/\bar{x}$, where x_i represents a set of measured data, with mean $\bar{x}$, and σ_x is the standard deviation.

collimation. Geometrical limitation of the extent of the X ray beam in the z direction.

contrast medium. A substance introduced into structures to increase or decrease their contrast, enhancing otherwise low contrast resolution on the CT image. Most commonly, a radio-opaque substance is injected intravenously, but an oral contrast medium may also be used.

CT fluoroscopy. Use of CT to produce almost real time images.

CT number. Number used to represent the mean X ray attenuation associated with each elemental area of the CT image (see also Hounsfield units).

diagnostic reference level (DRL). Indicative dose level for a given examination of patients with standard body size against which actual doses can be audited.

effective dose. Representative dose measurement that gives the uniform whole body dose, which is equivalent, in terms of stochastic (carcinogenic) risk, to the non-uniform irradiation of various organs. Effective dose (E) is quoted in units of millisieverts (mSv).

effective mAs. The tube current–exposure time product (mAs) divided by the pitch; an estimate of the mAs per slice.

field of view (FOV). Size of an area being imaged. Can be either the scanned field (SFOV) or the reconstructed field (RFOV).

filter. This can be:

(a) X ray filter: Material introduced to remove soft X rays from the X ray beam.
(b) Beam shaping filter or bow-tie filter: Material used to shape the intensity of the X ray beam to suit the subject.
(c) Reconstruction filter: Mathematical kernel used during the reconstruction process to modify responses at different spatial frequencies (see reconstruction algorithm).

full width at half maximum (FWHM). Method for characterizing the width of the X ray beam by measuring the distance between the points at which the intensity is 50% of the peak.

gantry tilt. Capability to offset the gantry rotation from the couch axis. Used in head scans to avoid critical dose sensitive organs, such as the eyes.

geometric efficiency. One measure of dose efficiency of the scanner; it is the ratio of the slice width imaged to the total slice width irradiated. Strictly, it should be the ratio of dose used to create images to total dose.

half-value layer (HVL). The thickness of a specified material which when introduced into the path of a beam of radiation reduces the exposure rate by half.

health level seven (HL7). Used here to refer to standards and frameworks related to health care interoperability, i.e. the transfer of data between different systems in health care. Also, it is the name of the non-profit organization that develops such standards.

helical (spiral) CT. A CT scan where the patient is moved constantly through the rotating gantry. The X ray beam describes a helix about the patient.

imaged slice width. Effective thickness of the tomographic section, as measured by the full width at half maximum of the sensitivity profile in the centre of the scan field.

Integrating the Healthcare Enterprise (IHE). Global initiative designed to advance the state of data integration in health care across all hospital systems. IHE promotes the coordinated use of established information standards, such as DICOM and HL7, through common functional profiles.

isocentre. The point around which the CT system rotates. Also, the visual centre of the reconstructed image.

longitudinal (z) mA modulation. Addresses the varying attenuation of the patient among anatomical regions by varying the mA along the z axis (length) of the patient (e.g. shoulders versus the abdomen versus the pelvis). The task of z modulation is to produce relatively uniform noise levels across the various anatomical regions.

low contrast resolution. Lowest contrast detail of an object of a specified shape and area that can be resolved from a uniform background.

mean CT number. Mean value of the CT numbers of all pixels within a certain defined region of interest (ROI).

MDCT (multidetector CT). MDCT systems are CT scanners with a detector array consisting of more than a single row of detectors. Also called multislice, or multirow CT scanners.

modulation of X ray tube current (mA). A method of managing the dose in CT by automatically adjusting the current to achieve the selected image quality. May be angular or longitudinal mA modulation or both.

modulation transfer function (MTF). The capability of imaging equipment to reproduce spatial detail of an object in an image.

multiphase examination. An examination in which two or more sets of images of a defined anatomical area are taken (e.g. before and after the injection of a contrast medium).

noise. Variation of CT numbers from a mean value in a defined area in the image of a uniform substance. The magnitude of noise is indicated by the standard deviation of the CT numbers of a uniform substance in the region of interest.

nominal tomographic section thickness. In CT scanners, the nominal tomographic section thickness is selected and indicated on the control panel. It should be noted that in helical scanning, the thickness of a reconstructed image depends on the helical reconstruction algorithm and pitch, and hence this thickness may not equal the nominal tomographic section thickness. The thickness of the reconstructed image may be indicated or selected prior to the helical scan.

organ dose. The mean absorbed dose in an organ or tissue, i.e. the total energy absorbed by the organ or tissue divided by the total mass of the organ or tissue.

over-beaming. The situation arising when the X ray beam incident to the patient extends beyond the active detector area and hence is not used for imaging purposes.

over-ranging. The increase in dose–length product due to the additional rotations required for the helical (spiral) interpolation algorithm. Typically, an extra rotation is required at the beginning and at the end of each helical acquisition so that the extent of the effect will depend on the pitch and the X ray beam collimation.

partial volume effect or averaging. The inaccuracy in a CT number caused by the presence of a structure that is only within part of a slice.

phantom. A device that absorbs or scatters radiation in an equivalent manner to that of a patient and which is utilized to estimate radiation dose and to test imaging systems without actually exposing a patient. A phantom may be an anthropomorphic or a physical test object.

picture archiving and communication system (PACS). A system that manages the storage and retrieval of digital images.

pitch. In helical scanning, pitch is the ratio between the distance moved by the patient during one gantry rotation (the couch feed) and the width of the image sample being scanned (the beam collimation).

polymethylmethacrylate (PMMA). A commercially available plastic polymer, e.g. Perspex, acrylic or Lucite.

radiology information system (RIS). A computer system which stores the appointment information for a radiology department and may be linked to the hospital information system. A PACS may take examination booking information and demographics from the RIS to form worklists.

reconstruction algorithm (convolution kernel). A mathematical operation applied during image creation that can, for example, reduce image noise or enhance edge details.

reconstruction matrix. The array of pixels that is displayed as the CT image.

region of interest (ROI). Localized part of an image, which is of particular interest at a given time.

resolution. The capability to distinguish small discrete objects with an imaging system. This may be measured using a bar pattern and is frequently described in terms of the spatial resolution in line pairs per cm (lp/cm) or MTF.

scan time. For a single exposure, the time interval between the beginning and the end of the acquisition of attenuation data. This may be longer than the exposure time owing to pulsing of the X ray emission with some CT scanners.

scan projection radiograph (SPR). Initial (low dose) scan acquired using a static tube and moving bed. This is used to plan the full CT scan sequences and may also be used for tube current modulation. Sometimes referred to as a ‘scout’ view or ‘scanogram’.

slice. A tomographic section defined by the position of a test phantom or patient under investigation during a single CT exposure.

spatial resolution. Defines the smallest feature that can be detected in an image. This is usually quoted as line pairs per cm (lp/cm) for CT scanners.

temporal mA modulation. Temporal mA modulation alters the tube current according to a time based criterion. This is most commonly used in CT examinations of the heart, reducing the dose for projections of limited interest, such as in early systole where the rapid cardiac motion compromises image quality.

tolerance values. Express the range over which the parameter is allowed to vary before the item is no longer considered to be operating within limits. These ranges are classified into two categories: achievable and acceptable.

tube current–exposure time product (mAs). The product of the X ray tube current (mA) and the exposure time in seconds (s).

tube current modulation. Variation of tube current according to the relative attenuation of the object.

workstation. A computer running applications software. There are several types, the main ones being:

(a) Diagnostic or reporting workstation: Used for viewing images for primary diagnosis and/or production of clinical reports. A reporting workstation will commonly have multiple diagnostic (high quality) display monitors, usually two or four screens, capable of meeting specific guidelines for resolution, contrast and brightness. These may be for CT reporting only, or shared for reporting all radiology images.
(b) PACS workstation: Capable of retrieving and displaying image data from the PACS. Varies from a simple review workstation on the wards to a diagnostic workstation with specialist reporting tools and RIS integration.
(c) Review workstation: Used for checking that acquired images are of the required quality and that all necessary views have been performed. Often, at least one display on the CT control console will act as a review workstation. Review workstation may also refer to a workstation used on a ward or in a clinic to view images in conjunction with the associated clinical report. A review workstation will usually have one or two display monitors, which are of lower display quality than those used in reporting workstations.
(d) Scanner acquisition workstation: Workstation supplied with the CT scanner and optimized to run specialist applications.

x–y plane. The plane in which the CT scanner gantry rotates. Also called the scan plane.

z axis. Axis about which the gantry of a CT scanner rotates. Also called the scan axis. Normally the superior–inferior axis of the patient.

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