IAEA HUMAN HEALTH SERIES

No. 1

Quality Assurance for PET and PET/CT Systems

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QUALITY ASSURANCE FOR PET AND PET/CT SYSTEMS

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INTERNATIONAL ATOMIC ENERGY AGENCY
VIENNA, 2009

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Sales and Promotion, Publishing Section
International Atomic Energy Agency
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PO Box 100
1400 Vienna, Austria
fax: +43 1 2600 29302
tel.: +43 1 2600 22417
email: sales.publications@iaea.org
http://www.iaea.org/books

Printed by the IAEA in Austria
July 2009
STI/PUB/1393

IAEA Library Cataloguing in Publication Data

Quality assurance for PET and PET/CT systems. — Vienna : International Atomic Energy Agency, 2009.
p. ; 24 cm. — (IAEA human health series, ISSN 2075–3772 ; no. 1)
STI/PUB/1393
ISBN 978–92–0–103609–4
Includes bibliographical references.

1. Tomography — Quality control. 2. Radiography, Medical — Instruments — Quality control. I. International Atomic Energy Agency. II. Series.

IAEAL 09–00582

FOREWORD

Improvement in quality assurance (QA) in nuclear medicine and, in particular, in quality control (QC) of related equipment has for a long time been a major field of interest of the IAEA. Starting from the late 1970s, several publications were produced, including Quality Control of Nuclear Medicine Instruments (IAEA-TECDOC-317) in 1984, and the still widely used revision of this publication, IAEA-TECDOC-602, issued in 1991. Additional QC of single photon emission computed tomography (SPECT) systems has been addressed in the IAEA Quality Control Atlas for Scintillation Camera Systems, which provides a comprehensive set of sample SPECT artefacts. Positron emission tomography (PET) scanners and related performance assessment and QC were not included in the previous publications, as PET has been mainly a research tool, with limited distribution until the 1990s. The tremendous role played at present by PET and PET/CT in oncology, as well as in cardiology and neurology, associated with the increasing use of PET for multiple purposes has prompted the need for updated guidelines specific to PET and PET/CT in terms of acceptance testing, as well as in terms of QC and QA.

This publication provides guidelines for the implementation of QA and QC programmes concerning the combined medical diagnostic modality of PET and CT technologies. The use of these independent, but complementary, imaging techniques is frequent and growing within the fields of diagnostic imaging, oncology, cardiology and neurology, where they allow physicians to locate and diagnose malignant diseases accurately. Specific topics of discussion include the frameworks for reference values, tolerances and action levels, minimal required configurations with corresponding performance characteristics, and the management of ancillary equipment.

The IAEA wishes to express its thanks to the main contributors to this publication; in alphabetical order, G. El Fakhri (USA), R. Fulton (Australia), J.E. Gray (USA), M. Marengo (Italy) and B. Zimmerman (USA). The IAEA officers responsible for this publication were M. Dondi, I.D. McLean and S. Palm of the Division of Human Health.

EDITORIAL NOTE

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CONTENTS

1. INTRODUCTION

The refinement of standardized performance measurements for positron emission tomography (PET) scanners has been an ongoing process over the last ten years. The initial efforts, initiated by the Society of Nuclear Medicine and further elaborated by the National Electrical Manufacturers Association (NEMA) of the Unites States of America, resulted in the creation of an initial standard, the NU2-1994 document. In the same period, the European Union started to develop a standardized performance test, which resulted in the International Electrotechnical Commission (IEC) standard. Despite some similarities in the way some procedures were performed, there were distinct differences in the way the performance tests, including use of different phantoms, the data acquisition procedures as well as the image reconstruction procedures, were performed. The early 1990s saw the clinical introduction of fully three dimensional (3-D) PET systems that operate without interplane septa. These scanners were much more sensitive to scattering and randoms originating from radioactivity outside the field-of-view (FOV), and there was a need for performance tests that were more relevant to this mode of operation. In 2001, the 1994 NEMA standard was updated to NEMA NU2-2001, to specifically cater for 3-D scanning and the effects of out-of-field radioactivity. The NEMA NU2-2001 standard introduced a new 70 cm long phantom with an off-centre line source to provide a more realistic whole body radioactivity distribution with out-of-field radioactivity. In addition, the introduction of an image quality test that assessed the overall performance of the scanner using a torso phantom with out-of-field radioactivity allowed the performance of different scanners to be compared under more realistic conditions. A revised version of this standard, NEMA NU2-2007, was released in 2007, which incorporates changes to cater for the introduction of PET scanners with intrinsically radioactive components.

The benefit of these standards is that they allow direct comparison of PET scanners from different vendors, as well as providing standardized and well described tests for assuring that the scanners meet their specified performance. PET technology is continually evolving, and the NEMA standards do not address the relatively recent addition of the computed tomography (CT) component, and the accompanying need to ensure proper registration of the PET and CT data. The present guidelines are intended to provide standard testing procedures that address both the PET and CT components.

Since PET manufacturers currently specify the performance of their systems using NEMA NU2-2001 parameters, and may be expected to follow

NEMA NU2-2007 procedures in future, it is essential that acceptance tests follow the same procedures so that performance parameters may be compared. Therefore, many of the PET acceptance tests described in this publication adhere closely to the NEMA 2007 standard.

Once the instrument that is being tested passes all of the acceptance tests, 'benchmark tests' must be performed. These tests are a set of quality control (QC) tests that are performed in the same way as the routine QC procedures. The benchmark tests serve as a baseline for instrument performance and are used to evaluate subsequent QC tests. The tests should also be used to evaluate instrument performance after major services and updates in software, and must be repeated after upgrades in hardware.

The present publication is a technical reference book that provides guidance about the specifications and prerequisites required for acceptance testing of PET and PET/CT scanners, including the professionals to be involved, definitions of applications, minimal required configurations and corresponding performance parameters, as well as ancillary equipment.

It also provides guidelines for routine QC of PET and PET/CT scanners, as well as a framework for setting reference values, tolerance values and action levels. Following these guidelines would ensure operation of a scanner under optimal conditions that yield the best performance in routine clinical tasks that involve lesion detection as well as quantitation of the radioactive material concentration. Such tasks are crucial for early detection of lesions in whole body oncological PET as well as staging, follow-up and therapy monitoring in oncological PET. These tasks are also crucial for quantitation of radioactivity when assessing the response to therapy or quantitating the uptake of a radiopharmaceutical. The same considerations apply to other indications of PET/CT in cardiac, neurological and inflammation imaging.

2. BASIC PET/CT TECHNOLOGY

2.1. INTRODUCTION

Positron emission tomography is based on the detection, in coincidence, of two 511 keV photons emitted, at 180 ± 0.5º from each other, following the annihilation of a positron and an electron. By using appropriate radiation detectors and coincidence electronics circuitry, the two photons are detected in coincidence within a limited time window, allowing the identification of a line

of response (LOR), i.e. a line along which the positron annihilation was located. By acquiring a large number of LORs (several hundred millions), it is possible to reconstruct the distribution of the radioactive nuclei inside the volume studied. This characteristic is of paramount importance, as it obviates the need for a collimator to determine the direction of emission of the photon and yields a significantly increased sensitivity as compared with single photon emission computed tomography (SPECT).

2.2. PET TECHNOLOGY

2.2.1. PET radionuclides

Positron emission tomography is now a well established diagnostic modality that is extensively used in oncology for tumour diagnosis, staging, radiotherapy planning and monitoring, as well as cardiology for myocardial viability and perfusion, and neurology for perfusion and neuro-receptor imaging. Positron emission tomography also remains a strong molecular imaging modality, as radionuclides with adequate physical and biochemical characteristics are available and are being developed. Ideal PET radionuclides need to be:

(a) Readily available or (relatively) easy to produce, in adequate quantities, and with the required purity;
(b) Suitable for synthesis of radiopharmaceuticals that allow the study of biochemical processes in vivo.

Among positron emitters are several interesting radioisotopes of fundamental 'building blocks' of organic molecules such as carbon, nitrogen and oxygen, as well as radioisotopes of other elements, such as fluorine, which can be efficiently used to label a large variety of substrates and molecules of pharmaceutical relevance, allowing the study of numerous biological processes in vivo at the molecular level.

Table 1 shows the principal PET radionuclides, with some of their relevant physical characteristics.

The short half-lives of these radionuclides require, in most cases, that they be produced using dedicated cyclotrons or generators at the same site where they will be used. However, ^{18}F based radiopharmaceuticals can be distributed over a reasonable distance, a property that has greatly contributed to the widespread use of PET as a diagnostic modality in the oncological setting. It is worthy of note that ^{68}Ga and ^{82}Rb are available from generators.

TABLE 1. PHYSICAL CHARACTERISTICS OF THE PRINCIPAL PET RADIONUCLIDES

Radionuclide	Source	Half-life (min)	Maximum (and mean) positron energies (keV)	Mean positron range in water (mm)
C-11	Cyclotron	20.4	970 (390)	1.1
N-13	Cyclotron	9.96	1190 (490)	1.3
O-15	Cyclotron	2.07	1720 (740)	2.5
F-18	Cyclotron	110	635 (250)	0.5
Ga-68	Generator	68	1899 (836)	0.8
Rb-82	Generator	1.25	3356 (1532)	1.5

2.2.2. The fundamental limits of PET

From a physics point of view, the ability of a PET system to correctly position the annihilation events is affected by several factors:

(a) The range of the emitted positron in matter before annihilation;
(b) The dimensions of the crystals used in the detector;
(c) The non-exact collinearity of the annihilation photons.

A thorough discussion of these factors can be found elsewhere [1], and only a short summary is presented in Sections 2.2.2.1–2.2.2.4 below.

2.2.2.1. Positron range

Positrons do not immediately annihilate when they are emitted. Instead they travel some distance in matter, depending on their initial kinetic energy and the electron density of the absorbing material. The emitted positrons have a continuous distribution of kinetic energy values, ranging from zero to a maximum energy (Table 1). Therefore, the range of positrons is not a fixed value but rather a distribution of values that can be characterized by a full width at half-maximum (FWHM). For ^{18}F and ^{11}C, this value, $FWHM_p$, is of the order of 0.1–0.5 mm in water, and for ^{13}N and ^{82}Rb, which are widely used in cardiac imaging, it is of the order of 1.3–2.5 mm.

2.2.2.2. Detector dimensions

The finite dimensions of radiation detectors do not allow us to draw an LOR between two detectors but rather a small volume.

For a point source at the centre of the detection volume, and for small crystals (small compared with the distance between a pair of detectors), the response of the system can be described by a triangular function with a spread equal to:

$$\mathrm{FWHM}_D = w_D/2 \tag{1}$$

where w_D is the dimension of a detection element. This is shown graphically in Fig. 1.

In modern scanners, the transverse dimension of crystals is about 4–6 mm, as compared with a length of 2–3 cm, and the contribution to total system spatial resolution is of the order of 2–3 mm FWHM.

2.2.2.3. Non-collinearity

Annihilation photons are not emitted exactly at 180° because the positronium (positron plus atomic electron) has some residual momentum. This results in a spread of the system's response function, which is given by:

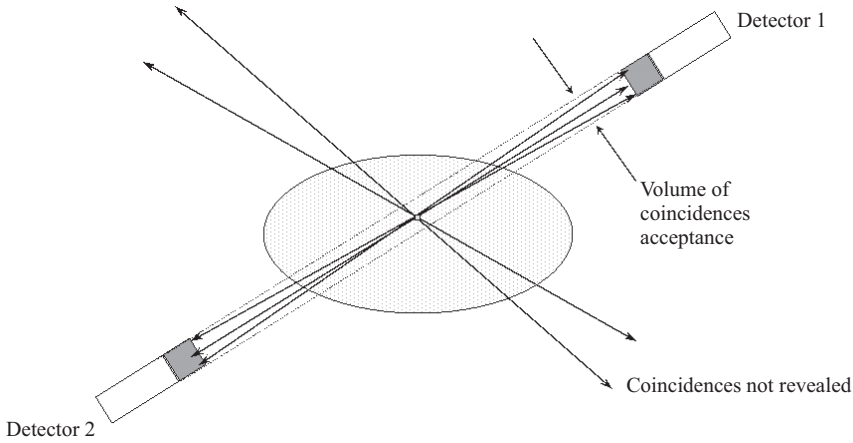

FIG. 1. A pair of detectors 180° apart, operating in coincidence mode. The finite dimensions of the detectors define a volume of acceptable LORs.

$$FWHM_N = 0.0022 d_s \quad (2)$$

where d_s (in millimetres) is the distance between opposite detector elements or the diameter of the detection system.

In clinical scanners that are state of the art, this spread is of the order of 800–900 mm, and thus the contribution of non-collinearity to the spatial resolution of the scanner is about 1.8–2.0 mm.

2.2.2.4. Technological factors

The resolution of a PET scanner is influenced by the factors described above, and depends mainly on the physical characteristics of the radionuclide and the detection system. However, other factors must be included in order to model correctly the spatial resolution of a PET system. As pointed out by Moses and Derenzo [2], for scanners based on detector elements organized in blocks, a spread of $FWHM_B = 2$ mm should be added.

The theoretical limit of the spatial resolution of a PET scanner can thus be estimated by:

$$FWHM_{tot} = (FWHM_D^2 + FWHM_N^2 + FWHM_p^2 + FWHM_B^2)^{1/2} \quad (3)$$

The above equation assumes a perfect spatial resolution restoration by the reconstruction algorithm. In practice, an additional degradation may result from the specific reconstruction algorithm used and the choice of reconstruction parameters. For example, in the case of ^{18}F, the overall spatial resolution that can be achieved in the reconstructed volume by a clinical PET scanner that is state of the art is in the range 4–6 mm FWHM.

2.2.3. Detector crystals for PET

Annihilation photons are not only more energetic than those in SPECT and planar imaging, but also need to be detected in coincidence. This places specific requirements on the crystal detector materials used in clinical PET scanners. The characteristics of some scintillator materials used in commercial PET scanners are listed in Table 2.

Sodium iodide activated with thallium, NaI(Tl), is a reference scintillation crystal widely used in SPECT for its excellent light yield and energy resolution, in combination with a relatively high effective atomic number Z_{eff}. It has also been used in PET scanners, despite its lower stopping power for 511 keV photons and relatively slow light decay constant. Bismuth germanate oxide (BGO), on the other hand, has a very good stopping power for 511 keV

TABLE 2. CHARACTERISTICS OF THE MOST IMPORTANT SCINTILLATOR CRYSTALS USED IN PET

Crystal material	Light yield (photons/MeV)	Emitted light wave-length (nm)	Light emission decay time (ns)	Density (g/cm^3)	Effective atomic number	Refractive index	Energy resolution at 511 keV (%)
NaI(Tl)	38 000	415	230	3.7	51	1.85	10
BGO ($Bi_4Ge_3O_{12}$)	9 000	480	300	7.1	75	2.15	20
LSO (Lu_2SiO_5)	26 000	420	40	7.4	66	1.82	15
LYSO $(Lu_{1-y}Y_y)_{2(1-x)}SiO_5$	32 000	430	40	7.1	66	1.82	12
GSO (Gd_2SiO_5)	13 000	440	50	6.7	59	1.85	15
$LaBr_3$ (5% Ce)	60 000	370	25	5.3	47	1.9	10
LuAP[a] (0.4% Ce) ($LuAlO_3$)	12 000	365	18	8.3	65	1.94	7

[a] LuAP: Lutetium aluminium perovskite.

photons and has been widely used in clinical PET scanners since the 1990s. Both NaI(Tl) and BGO are slow scintillating materials, i.e. both these materials are relatively slow in re-emitting, in the form of visible light photons, the energy absorbed in the interaction with ionizing radiation. As a result, these scintillators require a relatively long time coincidence window. There has been a significant effort made by several manufacturers to introduce faster scintillating materials such as lutetium or lutetium/yttrium oxyorthosilicate (LSO/LYSO), which have similar densities and Z_{eff} to BGO, a higher light yield than BGO and a faster light decay. It is noteworthy that naturally occurring lutetium comprises two isotopes, ^{175}Lu and ^{176}Lu; the latter is naturally radioactive and, although present in small amounts (2.6%), produces undesired single events in the PET detector that, although negligible at clinical count rates, can affect low count rate measurements such as those in some QC procedures.

Another interesting scintillator is gadolinium oxyorthosilicate (GSO), which has slightly lower density and light yield than LSO, but better energy resolution than BGO and LSO, and a similar light decay constant to LSO and LYSO. Other crystals such as yttrium aluminium perovskite (YAP) have been used in small animal PET scanners, and lanthanum bromide (LaBr) has been proposed for time-of-flight (TOF) clinical PET, by virtue of its excellent energy

resolution and very fast light decay. Cost issues and availability remain limiting factors to the widespread adoption of new scintillating materials.

2.2.4. Detector configuration

Positron emission tomography scanners use a large number of small crystals, with cross-sectional dimensions of 4–8 mm (e.g., cross-sections of 4 × 4, 4 × 6, 6 × 6 and 4 × 8 mm^2) and depths of 20–30 mm. In many scanners, detectors are organized in blocks, for example, an array of 8 × 8 crystals encoded on an array of 2 × 2 photomultipliers. About one hundred blocks form a ring of detectors, and state of the art scanners have three to four complete rings of blocks, yielding an axial FOV of 15–22 cm. As a result, 12 000 to 18 000 individual crystals are needed, depending on the specific characteristics of each scanner, to build a scanner. Other designs use flat panels, about 9 cm × 18 cm, of pixelated crystals, every crystal element being 4 mm × 6 mm. Each panel has then more than 600 crystals and is coupled to an array of 15 photomultipliers; about 28 panels are used to build a full ring, yielding a total of about 18 000 individual crystal elements. The large number of crystals, added to the complexity of manufacture and assembly required to build the PET detector, are responsible for a large part of the cost of PET scanners.

2.2.5. Components of the response of a PET scanner

The events detected in a PET scanner consist not only of 'true' coincidence events but also of unwanted events arising from other mechanisms of interaction in the patient and detector that do not contribute useful information.

When a single event is detected in a crystal, it has to satisfy an energy acceptance criterion; the energy of the pulse produced must be higher than the lower level energy discriminator (LLD), i.e. set depending on the type of crystals used and the trade-off between sensitivity and spatial resolution (typically 350 keV for BGO and 420 keV for LSO/LYSO). The system then checks whether another valid event has been detected within a limited coincidence time window (CTW). The CTW duration is set, taking into account the light decay characteristics of the crystal used and the design of the electronics.

When two valid events are detected within the CTW, the position of the two events defines an LOR and the event is recorded. Spurious events such as randoms and scattering satisfying the energy and timing acceptance criteria are also detected (Fig. 2).

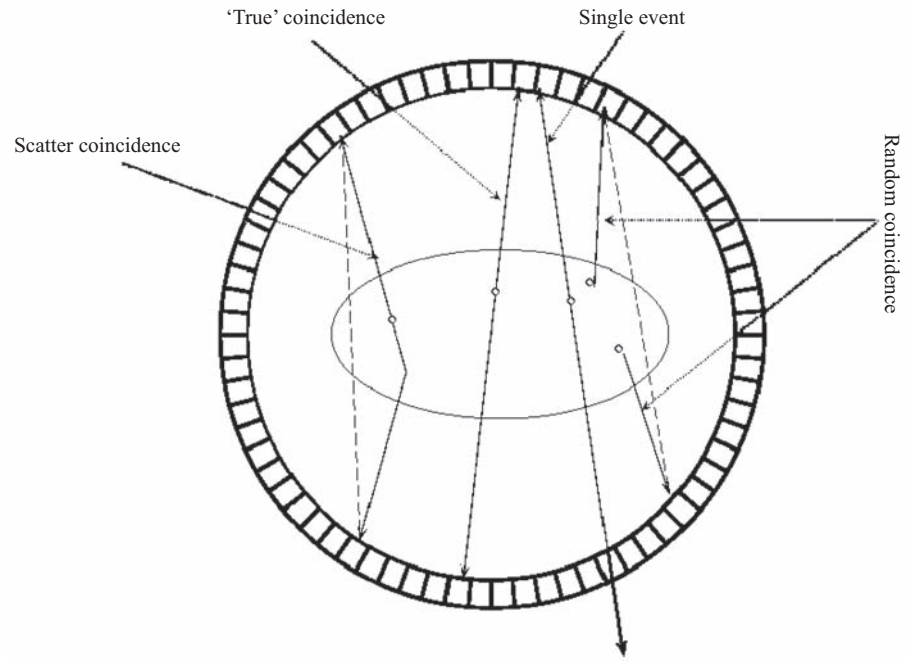

FIG. 2. Illustration of the events detected in a PET scanner.

Randoms are detected when two independent photons arising from separate annihilation events are recorded in coincidence. The probability of detection of randoms increases with the total single event count rate. If R_i is the single event rate on a detector i, the randoms coincidence rate between two detectors is given by the following expression:

$$R_{12} = 2\tau R_1 R_2 \tag{4}$$

where τ is the length of the CTW.

The above equation can be used for correction of randoms, provided the system is able to record the single event rate on each detector. Alternatively, PET scanners can correct for randoms by using a second delayed time window, which provides an estimate of the rate of randoms. Correction is applied by subtracting these randoms from the data acquired in the first window.

If one of the annihilation photons undergoes a Compton interaction, the original direction of propagation is changed, but the energy of the scattered photon may still be greater than that of the LLD, particularly if the latter is set low due to limited energy resolution of the crystal material. If the unscattered

and scattered photons are detected in coincidence, this will produce a mispositioned LOR, increasing blurring in the reconstructed image. For a description of scatter correction techniques, see Refs [3–5].

The total number of prompt coincidence events, P, detected in a PET scanner can be expressed as the sum of true, random and scattered coincidence events:

$$P = T + R + S \tag{5}$$

2.2.6. Two dimensional and three dimensional operation

In the so-called two dimensional (2-D) mode of operation, septa made of a high Z material, such as tungsten, are placed between adjacent rings of detectors, in order to limit the acceptance angle of LORs and to reduce scatter and random coincidences, at the expense of a reduction in the efficiency of the scanner to true coincidences.

In order to improve the efficiency of detection of true coincidences, scanners operating in 2-D mode can optionally accept coincidences not only in the direct planes but also in adjacent cross-planes.

In modern scanners, septa may be retracted or not present. In this case, the system operates in three dimensional (3-D) mode, and all coincidences detected within a sensitive volume of the scanner can be accepted (Fig. 3). Therefore, 3-D septa-less whole body PET yields increased sensitivity to true

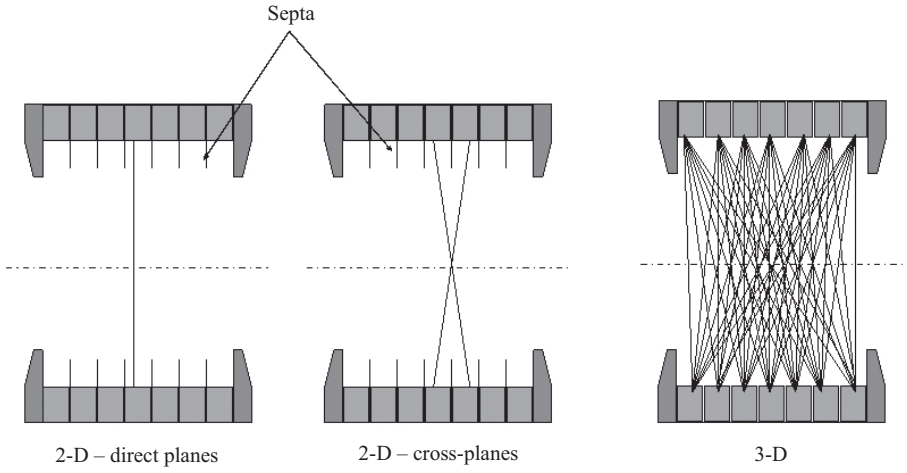

FIG. 3. Illustration of 2-D and 3-D modes.

coincidences at the expense of increased sensitivity to scattered and random coincidences. Newer crystal materials used in 3-D scanners have better energy resolution (≈15% for GSO and LSO and 12% for LYSO as compared with ≈20% for BGO), allowing the use of a higher LLD to discard a higher proportion of scattered events. Furthermore, these crystals have shorter light decay constants, allowing the use of a shorter CTW and a reduction of random coincidences.

2.2.7. Time-of-flight PET

Fast scintillating crystals with fast rise times and short decay times allow the measurement of the difference of arrival times of the two 511 keV photons emitted following the annihilation. The difference in arrival times will yield information regarding the distance travelled by each of the two annihilation photons and would, therefore, restrict the likely location of the annihilation event to a portion of the LOR. Since the likely location of emission associated with a 500 ps full width at half-time timing resolution (typical of the fastest crystals available at present) is known within 3×10^8 m·s^{-1} $\times$ 500 $\times$ 10^{-12} s ≈ 7 cm (uncertainty, Δx, in Fig. 4), no direct improvement in spatial resolution is expected, as the crystal size is typically 4–6 mm. However, the reduction in the

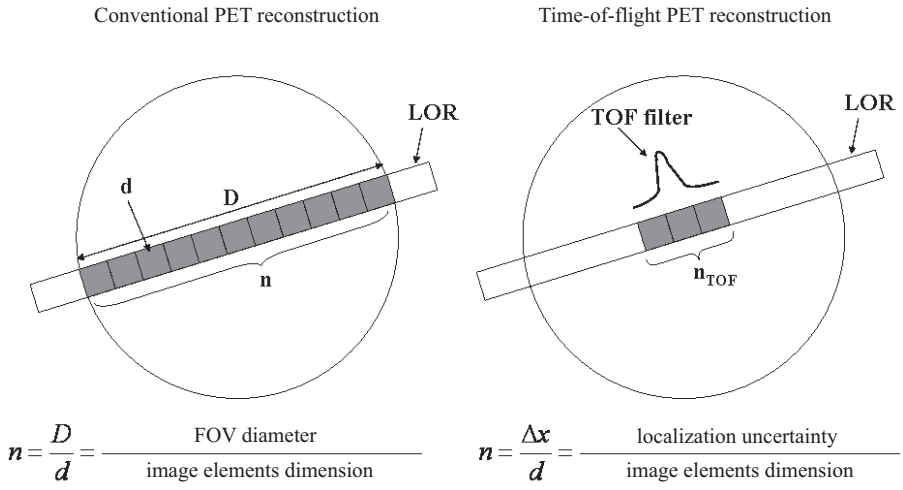

FIG. 4. Illustration of the implementation of TOF information in PET reconstruction. The length, n, of the region representing the likely location of the annihilation is shown schematically in units of detector element widths.

uncertainty of the likely location of emission will improve the signal-to-noise ratio within the tomographic reconstruction.

2.3. COMPUTED TOMOGRAPHY TECHNOLOGY

Computed tomography is a mature diagnostic modality that is still undergoing rapid development. The combination of CT with PET is an example of a new technical and clinical application as in isotropic imaging for cardiac assessment. The basic principles of CT imaging are found in many texts [6, 7]. A brief introduction will now be given.

Computed tomography 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 contrast to isotope imaging, where detected photons are emitted from the patient, an external source of X rays is projected through the patient to form a transmitted attenuation profile at the detectors. In order to obtain a cross-sectional image, the beam is restricted to a thin fan across the patient (in the x–y direction) of between 0.5 and 10 mm thickness for a single 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. 5).

2.3.1. Computed tomography gantry: Tube, collimator, filters and detector

The large X ray tube located within the gantry (Fig. 6) operates at between 80 and 140 kV. This tube can generate over 10^9 photons/(mm^2·s) at 75 cm from the tube focus for typical CT radiographic settings of tube voltage (120 kV) and current (300 mA). This is many orders of magnitude higher than is possible with isotope imaging and explains why a CT scanner can produce a low noise image in less than a second while isotope images require 10–30 min and produce images with significant noise.

Operation of X ray tubes at such high voltage and current values requires rapid dissipation of heat to avoid tube failure. Tube cooling systems are 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

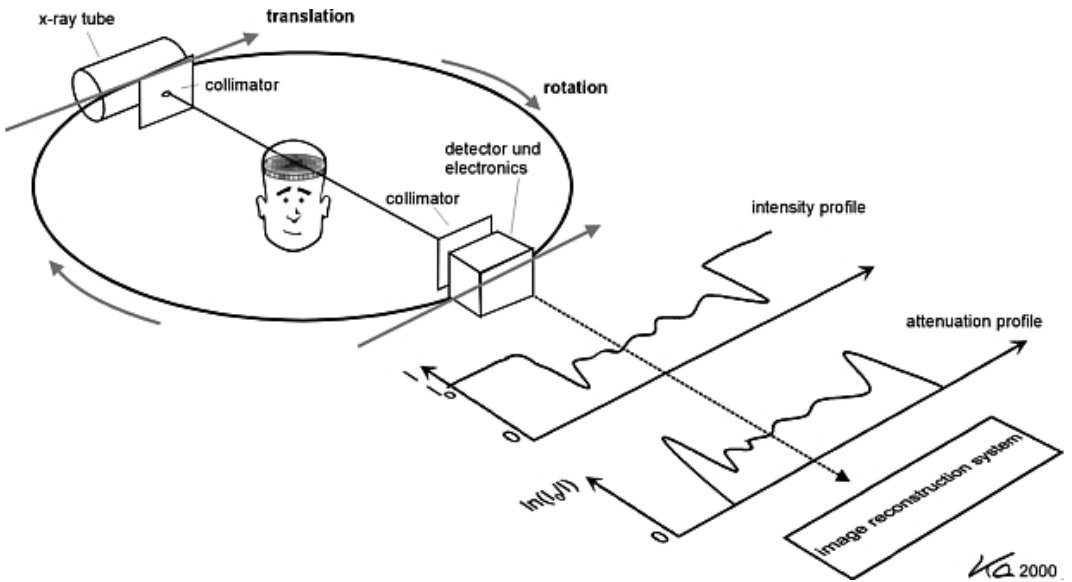

FIG. 5. Simplified diagram of creation of an attenuation profile in a CT scan. Note that modern scanners use a fan beam to acquire the attenuation profile in one exposure (courtesy: W.A. Kalendar [6]).

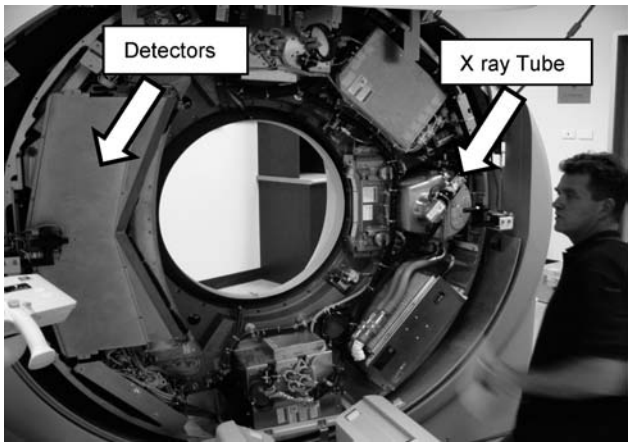

FIG. 6. Computed tomography scanner with gantry cover removed. Note the X ray tube on the right hand side with collimator and filters towards the scan aperture. The detectors are on the left hand side.

essential to use the correct filter for the part of the body being investigated. 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 256) are acquired simultaneously. The 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. Typical materials include Gd_2O_2S, xenon, YGdO and $CdWO_4$. Important specifications for such detector elements include a high dynamic range, high quantum absorption efficiency and a fast temporal response with a low afterglow. For a single slice axial scanner, the detector will have over 700 elements along an arc to intersect the exit beam of the scan plane. This is known as third generation scan geometry (Fig. 7) and is the basic design for modern CT scanners. In multislice scanners, the detector has additional adjoining arcs of detector elements. Such multirow detectors may have up to 256 rows, allowing a total acquisition width of 132 mm (measured at the isocentre). This type of acquisition can produce slice thicknesses of from 0.5 to 10 mm. With such a detector, the acquisition time is reduced and the occurrence of motion artefacts is considerably reduced.

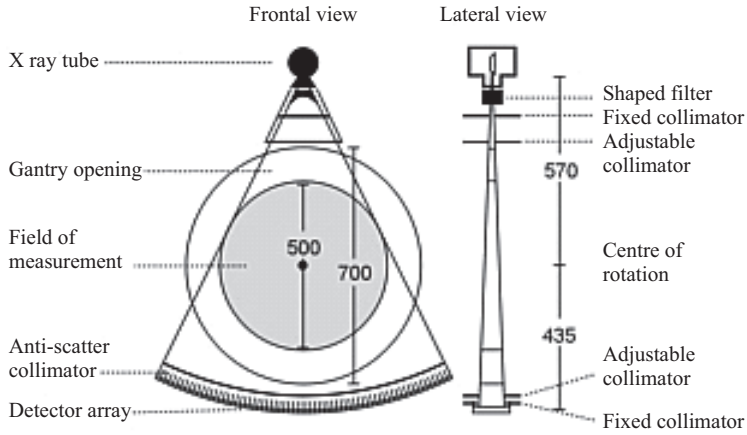

FIG. 7. Schematic representation of the scanning geometry and important components of the CT measuring system in both frontal (x–y plane) and lateral (y/z plane) views (adapted from Ref. [6]).

2.3.2. 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 from as little as 180° of rotation, while in spiral CT scanners, variable reconstruction angles are used. The reconstruction is primarily done by a filtered back-projection method (Fig. 8) that allows reconstruction in almost real time.

2.3.3. Scanning procedures

The simplest image acquisition procedure is the scout scan (Fig. 9), which is taken to plan a CT slice acquisition. This scan is really a digital radiograph, 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.

Axial slice scans involve 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 table is then moved to allow the acquisition of a new slice for a different anatomical region.

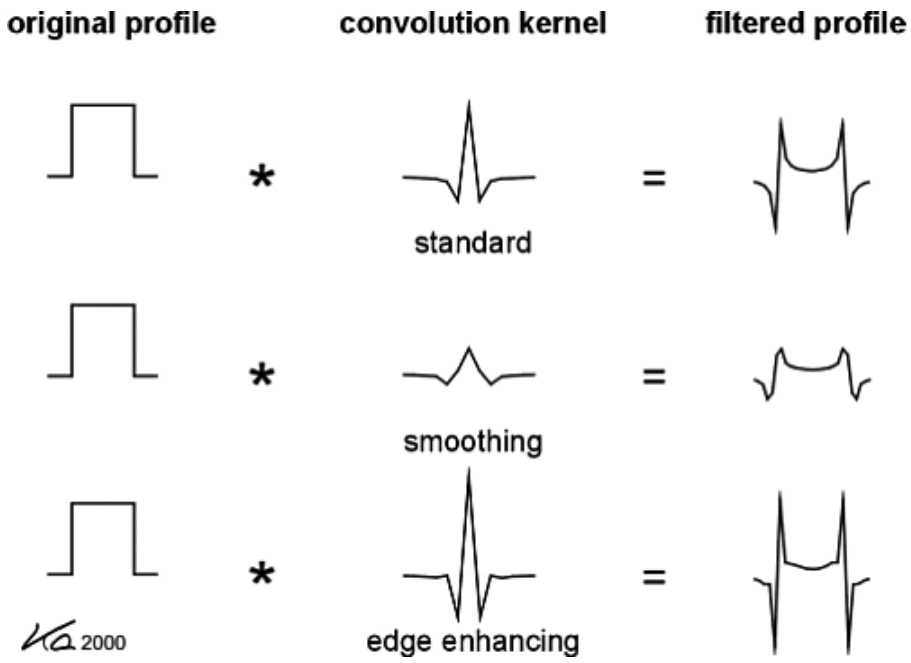

FIG. 8. Image characteristics can be influenced by the choice of convolution kernel, whereby increasing spatial resolution or edge enhancement also means increasing image noise (courtesy: W.A. Kalendar [6]).

Spiral or helical scanning is achieved when the table 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 greatly reduced. However, the reconstruction is complicated as the profiles are no longer in the same plane and need to be interpolated (so called z-interpolation) to a pseudo-planar state before reconstruction. Maximum benefit from spiral CT acquisition is achieved with multislice acquisition. Here, a process known as z-filtering is implemented. These algorithms have different characteristics than those of axial reconstruction algorithms. One notable feature is the ability to alter the reconstructed slice thickness to any thickness equal to or greater than the original acquisition thickness for an individual slice. This leads to the possibility of generating many sets of images from one acquisition, and can greatly increase the image storage requirements and data management practices of a department.

The key parameter describing spiral CT acquisition is pitch, defined as the ratio of the table advance during a 360° rotation and the width of the total

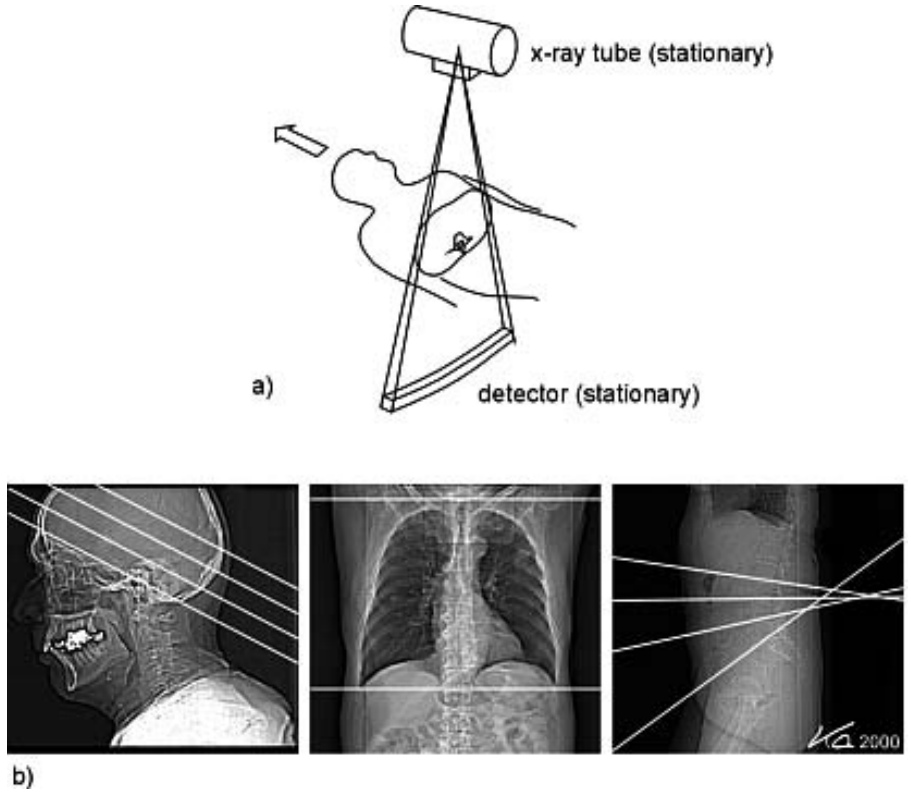

FIG. 9. Scout scan radiographs are taken with low dose and low spatial resolution by transporting the patient through the field of measurement with the X ray tube in a fixed position (a). The projection direction (here anterior–posterior) in principle is arbitrary. Scout scans allow selection of the position and gantry tilt (note that it is not possible to tilt the gantry in PET/CT systems) for single slices or complete scan regions as shown schematically (b) (courtesy: W.A. Kalendar [6]).

X ray collimation (i.e. the detector width multiplied by the number of detectors). Consequently, a large pitch implies faster acquisition with reduced dose, but with the cost of reduced resolution along the z axis, while low pitch has a slower acquisition and an increased dose but with better z resolution.

2.3.4. Image quality

As mentioned above, CT scanning utilizes a large photon flux in acquisition to achieve low noise images. These images allow the identification of very small differences in photon attenuation or low contrast differences in

tissue composition. The spatial resolution in a CT image is, however, limited to approximately 0.5 mm FWHM, which is inferior to most other radiological procedures but better than that possible with radioisotope imaging. Computed tomography scanners can also acquire slices of thin tissue thickness (with a minimum of 0.5 mm), which allows very precise cross-sectional delineation of structures without excessive partial volume effects. Partial volume effects are consequences of limited spatial resolution. They lead to a loss of signal in tissue regions of a size similar to the FWHM of the point spread function, distorting or reducing the contrast of features. The importance of this is reflected by the clinical desire for small slice width images for display.

Perhaps the greatest problem in CT imaging is the occurrence of artefacts. Many of these are due to equipment alignment. Because of this, it is essential that the room preparation be considered carefully. 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 phases to check for any indication in phantom images. Another cause of artefacts is electronic component failure or poor board connections.

2.3.5. Computed tomography doses

The large photon flux gives rise to good image quality as well as high patient dose. The main reason for excessive doses to patients is the use of inappropriate CT scan protocols. This is particularly true in the case of paediatric examinations [8]. It is essential for a medical physicist to determine the dose parameters (the weighted computed tomography dose index ($CTDI_w$) and the dose length product (DLP)) that would deliver an acceptable dose to typical patients. This is achieved by consideration of the CT scan protocol technique factors selected by the scanner staff and the measured scan dose data. These doses should be compared with national or international diagnostic reference levels (DRLs). If the doses are not significantly below these levels, the procedures should be examined to find out if the protocol modification is appropriate to allow dose reduction without impairment of diagnostic outcome. The scan factors that should be examined to determine what changes ought to be made should include [9]:

(a) Examining the reconstruction kernel or filter. By using a ‘smoother’ filter, the dose can be reduced by up to tenfold in extreme cases.
(b) Examining the scan length on the patient.
(c) Reducing the scan mA. While low mA can increase image noise, often this does not reduce the clinical significance of the image.

(d) Increasing the displayed slice width. This will reduce the noise, and often the clinical image quality is not impaired.

If children are involved in PET/CT imaging, special care should be taken. It should be recognized that DRLs are not well defined for patients of these sizes. It may be useful to estimate the effective dose for such procedures, as has been described in the literature [10].

It is important to remember that the CT images used in PET/CT are typically not used for diagnostic purposes. Consequently, it is possible to reduce the dose to the patient from the CT scan significantly, as a slight increase in the CT image noise will not be detrimental to the images for this application.

2.4. COMBINED PET/CT TECHNOLOGY

Positron emission tomography and CT provide useful and complementary clinical information: PET can identify functional abnormalities that might be undetectable on CT alone, while CT provides detailed anatomical information, but can normally identify malignancies only after structural changes have occurred. Identifying the precise location of a site of malignant disease often has a profound effect on decisions affecting the diagnosis, prognosis, staging, treatment and overall patient management. However, precise localization is difficult in PET due to the absence of identifiable anatomical structures in the reconstructed images.

The combined PET/CT scanner, introduced by Townsend and co-workers in 2000 [11], overcomes this problem by providing precisely co-registered anatomical and functional images in a single multimodal imaging session. Commercial versions of this device, which first appeared in 2001, comprise separate PET and CT scanners with a common scanning bed, placed in tandem. After injection with a tracer and an appropriate uptake period, the patient is positioned on the bed and undergoes a spiral CT scan, followed immediately by a PET scan.

The CT transmission images provide attenuation information for attenuation and scatter correction, as well as anatomical localization information on the same slices as the PET scan for precise localization of abnormalities seen on the PET scan. Software tools allow the display of transverse, coronal and sagittal sections of the PET and CT image volumes, either side by side, or as fused images in which the PET images are superimposed on the CT images. The blending of the two images can be varied to enhance the display.

2.5. COMPUTED TOMOGRAPHY BASED ATTENUATION CORRECTION

Prior to the introduction of PET/CT, the attenuation correction in PET was typically based on transmission measurements made with one or more rotating positron (typically ^{68}Ge) or single photon (typically ^{137}Cs) emitting sources prior to the PET emission scan. The advantages of PET/CT are that the transmission data can be acquired very quickly in a spiral CT scan, thus improving patient comfort and throughput, and that the transmission data are unaffected by radioactivity within the patient. A further advantage is that CT provides essentially noiseless attenuation correction factors. Since attenuation coefficients are energy dependent, attenuation correction factors derived from a CT scan must be scaled to account for the difference in attenuation of the X ray beam ($E \approx 70$ keV) and the energy of annihilation photons (511 keV).

Several practical problems need to be considered when using a CT based attenuation correction:

(a) The CT scan duration is very short (a few seconds) compared with the PET scan duration (a few minutes). This can result in artefacts in the attenuation corrected PET image due to mismatch between the CT data and the average attenuation distribution during the PET scan because of internal (e.g. breathing) or external patient motion.
(b) The scaling algorithms used to convert CT attenuation factors to values appropriate for 511 keV photons account correctly for the different properties of soft tissue and bone, but are not designed for other materials such as CT contrast agents. Attenuation correction based on CT in the presence of a CT contrast agent can therefore result in the application of invalid attenuation correction factors and an erroneous reconstructed image.
(c) There is an increase in the effective dose to the patient with CT as compared with a transmission source.
(d) There is a need to have a procedure to ensure that the CT data are properly registered with the emission.
(e) It is recommended to consider the uncorrected PET images as well as attenuation corrected images based on CT when there is a suspicion of artefacts associated with the use of attenuation based on CT.

3. CLINICAL APPLICATIONS OF PET/CT

3.1. INTRODUCTION

In the 1970s, PET scanning was formally introduced to the medical community. At that time, it was seen as an exciting new research modality that created possibilities through which medical researchers could watch, study and understand the biology of human disease. In 1976, the radiopharmaceutical [^{18}F]2-fluoro-2-deoxyglucose ([^{18}F]FDG), a marker of sugar metabolism with a half-life of 110 min, enabled tracer doses to be administered safely, with low radiation exposures, to patients. The development of radiopharmaceuticals such as FDG made it easier to study living beings, and set the groundwork for more in-depth research into using PET to diagnose and evaluate the effect of treatment on diseases in humans.

During the 1980s, the technology that underlies PET advanced greatly. Commercial PET scanners were developed that gave more precise resolution and images. As a result, many of the steps required for producing a PET scan became automated and could be performed by a trained technician and an experienced physician, thereby reducing the cost and complexity of the procedure. Smaller, self-shielded, cyclotrons were developed, making it possible to install cyclotrons at more locations.

Over the last few years, the major advance in this technology has been the combining of a PET scanner and a CT scanner in one device. Modern PET/CT scanners allow studies to be carried out in a shorter amount of time but still provide more diagnostic information.

Positron emission tomography and conventional nuclear imaging are both diagnostic radionuclide imaging techniques, and involve the use of radiopharmaceuticals (pharmaceuticals labelled with a radioactive isotope). These radionuclides can be localized by a variety of physiological or pathological processes using sophisticated imaging systems. Unlike conventional imaging techniques (diagnostic X rays, CT, magnetic resonance imaging (MRI) and ultrasound), which provide predominantly anatomical information, radionuclide imaging provides functional information on metabolic activity in physiological or pathological processes and only limited anatomical information. The detection of an abnormal lesion with these modalities is based on the differential radionuclide uptake within the lesion and in the surrounding tissues. Whether or not a lesion can be detected is related to the degree of radionuclide avidity, the size of the lesion and the background radioactivity.

Expectations about the utility of PET scanning are very high within the medical community. This technique combines the metabolic and localization

approaches and could, in theory, detect pathological process still invisible to classical imaging techniques. Where classical imaging techniques provide information about the structure and localization of lesions, PET scanning is used, as a complementary tool, to detect greater or lesser radiopharmaceutical uptakes and to characterize the function, metabolism, biochemical processes and blood flow of organs.

To reach this goal, a positron emitter is combined with a biochemical substance, active in the tissues. This is the case for the glucose analogue, which becomes [^{18}F]FDG when combined with the radioisotope ^{18}F. FDG-PET imaging in cancer is based on the property of increased glucose uptake into several malignant cell types and glycolysis within them. FDG, as a glucose analogue, undergoes glycolysis within tumour cells and is converted to FDG-6-phosphate intracellularly. However, in all tissues except those of the liver, FDG-6-phosphate is only slowly metabolized and is 'trapped' within the cell; hence its uptake becomes proportional to the glycolytic rate at tissue level.

[^{18}F]FDG is the most commonly used radiopharmaceutical in PET, since its half-life of 110 min allows commercial distribution as far as a travelling time of two to three hours away from production sites. Once in the body, FDG emits positrons, leading to the release of photons that are detected by PET scanners and the production of an image, to be interpreted by a nuclear medicine specialist. For other isotopes with much shorter half-lives (ranging from 2 min for ^{15}O to 20 min for ^{11}C), on-site production is required, typically using biomedical cyclotrons.

The determination of a positive result depends on the comparison between a specific region and the adjacent 'normal' regions. However, certain regions of the body are known to be physiologically glucose-avid. Therefore, the categorization of a region with augmented uptake is a very difficult process, based on a careful inspection of the ROI, contrasting the supposed lesion with the adjacent tissue. The difficulty of standardizing the reading of PET examinations explains why reported sensitivity and specificity may show such variations for the same indication. It is therefore of the utmost importance that PET scanners be checked on a regular basis to ensure proper scanner performance.

With such a process, the experience of the reader is the most important issue and, for that reason, there have been various attempts to objectify readings, at least in a semi-quantitative way, with the standardized uptake value (SUV) being the most common. The SUV is the ratio of the tissue concentration of the radiopharmaceutical (in Bq/g) to the injected dose (in becquerels) divided by the body mass (in grams). As such, it can be thought of as the ratio of the FDG uptake per gram in the tissue of interest to the injected dose per gram of body mass. It is important that the tissue concentration and injected dose are decay corrected to the same time, such as the start of the scan.

Owing to clearance of the tracer, the SUV obtained depends on the time between injection and imaging, and this may explain why reported SUVs vary considerably for the same indication.

Most frequently, clinical PET is used for the detection of lesions, and images are qualitatively assessed. It has been suggested that both attenuation corrected and uncorrected images should be used for lesion detection. While the need for attenuation correction for lesion detection remains debatable, it is certainly required in quantitative measurements of lesion uptake.

In a stand-alone, dedicated PET scanner, about 1 h is required to complete the emission and transmission acquisitions from skull base to thigh. The recent development of the faster scintillating crystals LSO and GSO as well of PET/CT systems has reduced total scanning times to less than 30 min.

The lack of anatomical detail in the PET images requires that the interpretation is usually made along with anatomical information obtained from CT or MRI. The introduction of hybrid machines, combining PET and CT scanners was aimed at overcoming this major limitation of PET. In the recently developed PET/CT system, a CT scanner is combined with a PET imager, typically in the same gantry. The CT acquisition is performed first, followed by PET acquisition. This set-up allows co-registration of PET data and CT data, producing fusion images with combined functional and anatomical details. In addition, attenuation correction is based on CT data, thereby reducing the total scanning time to less than 30 min. It has been proposed that PET/CT can improve PET images through fast and accurate attenuation correction, improve localization of abnormalities detected in PET, improve radiotherapy and surgery planning, improve evaluation of therapy outcome by localizing regions of oedema and scarring, and that it can produce the highest quality PET and CT information with the least inconvenience. The costs related to the acquisition and maintenance of a PET/CT scanner may be higher than those of a stand-alone PET scanner, but they may be outweighed by their potential for producing diagnostically superior images and for reducing scan time, thus allowing higher patient throughput.

Positron emission tomography has proven to be useful in a variety of medical fields, helping to detect certain types of cancer, cardiac disease and neurological disease. Although the list of recognized clinical applications of PET continues to grow as the research into it advances, at present most applications are in the field of oncology. The current indications are given in the following sections.

3.2. ONCOLOGY

Oncology remains by far the largest application of PET scanning [12], mostly in the staging/restaging phase, but in some cases also in the diagnostic process. Lung cancer was the first application, and PET is used for malignancy diagnosis of solitary pulmonary nodules (SPNs), for which there is evidence of diagnostic efficacy up to diagnostic applications based on the existence of a pre-test probability and a likelihood ratio, allowing the computation of a post-test probability. Moreover, a post-test probability threshold for cost effectiveness is provided by economic models [13]: evidence is supportive for the use of PET. In non-small-cell lung carcinoma (NSCLC), there is also evidence of diagnostic accuracy and that adding PET to CT is cost effective.

In lymphomas, there is evidence in the initial staging and recurrence diagnosis (involvement of lymph nodes and extralymphatic localization). There are also studies supporting the use of PET to evaluate possible treatment changes in patient management [14]. For residual mass evaluation, there is clinical evidence up to the diagnostic evaluation level, because PET contributes to medical decisions on follow-up strategy. For prognosis and evaluation of response to treatment, there is evidence of diagnostic accuracy, including the determination of sensitivity and specificity.

In head and neck cancers, PET is used to diagnose occult primary tumours suspected from cervical lymph node metastasis when clinical examination, panendoscopy with biopsy and/or conventional imaging modalities (CT/MRI) have failed to identify a primary tumour. In these applications, there is evidence of diagnostic accuracy, including the determination of sensitivity and specificity when occult primary tumours may be suspected from a single metastatic site outside the cervical lymph nodes following an unsuccessful initial diagnostic work-up, as well as when local or regional therapies are considered as part of a treatment plan for a single metastatic carcinoma outside the cervical lymph nodes [15].

In colorectal cancer, hierarchy levels of diagnostic efficacy of PET are established for the initial diagnosis and staging, as well as for detection and localization of local, hepatic and extrahepatic recurrences. The evidence of diagnostic efficacy includes changes in patient management and therapeutic decisions. Moreover, there is evidence for the cost effectiveness of PET in this application [16].

In melanomas, there is evidence of diagnostic accuracy, including the determination of sensitivity and specificity for staging, i.e. assessment of regional lymph node involvement or of distant metastatic disease in patients with primary or suspected recurrent melanomas [17].

In breast cancer, there is evidence of diagnostic inaccuracy for the use of PET in diagnosing patients referred for breast biopsies with abnormal mammograms or palpable breast masses, as benefits do not appear to outweigh risks. On the other hand, there is evidence of diagnostic accuracy including the determination of sensitivity and specificity, and evidence seems supportive for the use of PET, for staging/restaging, i.e. detection of distant metastatic disease if there is a clinical suspicion that metastatic disease is high at initial diagnosis or when recurrent breast cancer is suspected [18]. For the detection of locoregional recurrence, there is evidence of diagnostic accuracy, including the determination of sensitivity and specificity.

In oesophageal cancer, there is evidence of diagnostic accuracy including the determination of sensitivity and specificity, for staging, i.e. staging of lymph nodes (locoregional, distal or all lymph nodes) and distant sites other than lymph nodes. Evidence, although limited, seems supportive for the use of PET.

For restaging after patients who are eligible for curative surgery have received neoadjuvant therapy (comparative with initial staging PET results), there is evidence up to diagnostic evaluation based on diagnostic accuracy and prognosis.

In thyroid cancer, there is evidence of diagnostic accuracy including the determination of sensitivity and specificity, for restaging, i.e. detection of the recurrence of differentiated thyroid cancer in previously treated patients with elevated biomarkers not confirmed by whole body ^{131}I scintigraphy [19]. The same situation prevails for the detection of recurrence of medullary thyroid cancer in previously treated patients with elevated biomarkers not confirmed by other imaging methods.

3.3. CARDIOLOGY

Cardiac PET has been gaining great clinical acceptance in the assessment of myocardial perfusion using [^{13}N]ammonia and ^{82}Rb, as well as myocardial viability using FDG. With the advent of multiple slice CT PET/CT scanners (at present 64 slice CT PET/CT), PET/CT is poised to play an important role in the assessment of coronary obstruction, using CT angiography (CTA) along with myocardial perfusion PET [20]. This would have great potential in identifying patients eligible for revascularization and monitoring the results of such a procedure, yielding a 'one stop shop' with a PET/CT/CTA scanner. Furthermore, ^{82}Rb does not require access to a cyclotron, as it can be produced by an on-site generator.

One advantage of the integrated approach to the diagnosis of coronary artery disease (CAD) is the added sensitivity of PET and CTA, potentially

providing correct diagnoses in virtually all patients. The integration of PET and multidetector CT technology provides a potential opportunity to delineate the anatomic extent and physiological severity of coronary atherosclerosis and obstructive disease in a single setting. It allows detection and quantification of the burden of the extent of calcified and non-calcified plaques, quantification of vascular reactivity and endothelial health, identification of flow-limiting coronary stenoses, and, potentially, identification of high risk plaques in the coronary and other arterial beds. Together, by revealing the degree and location of anatomic stenoses and their physiological significance, as well as the plaque burden and its composition, integrated PET/CT can provide unique information that may improve non-invasive diagnosis of CAD and the prediction of cardiovascular risk. In addition, this approach expands the diagnostic capability of nuclear cardiology to include atherosclerosis and may facilitate further study of the progression of atherothrombosis and its response to therapy, thus allowing assessment of subclinical disease.

Thus far, the lack of widespread availability of PET scanners and radiotracers, their high cost, the limited amount of data supporting their use, and reimbursement problems have all contributed to the limited clinical acceptance of this imaging technology. However, widespread clinical acceptance by the imaging and clinical communities is unlikely to occur until more data documenting that PET can actually realize its clinical promise become available. Although data obtained more than 15 years ago supported the superiority of PET compared with SPECT for diagnosing CAD, it is unclear whether this superiority is still apparent at the present time because of major advances in SPECT instrumentation and imaging protocols [21, 22].

3.4. NEUROLOGY

Since glucose is the primary source of energy for cells in the brain, the radiopharmaceutical FDG, a glucose derivative, helps to create a map of normal versus abnormal brain function, as imaged in a PET scan. Distinctive patterns of glucose metabolism assist physicians in accurately diagnosing patients and treating them appropriately.

It now appears possible to diagnose Alzheimer's disease early on in its development, non-invasively and reliably, with PET [23]. It is now over ten years since the first report from PET laboratories described decrements in cerebral blood flow, oxygen utilization and glucose metabolism in the parietal and temporal lobes of patients with Alzheimer's disease. Since then, these findings have been extensively confirmed by numerous independent laboratories. The

lack of any other clinical, biochemical or genetic marker for Alzheimer's disease in living patients makes these findings unique and of clinical relevance.

Low grade gliomas (LGGs) account for 30–40% of all gliomas and are primarily treated with surgery. Since both timing and use of other oncological treatments in LGGs are a matter of controversy, there has been a constantly increasing demand to characterize these often slowly growing neoplasms with functional imaging methods, such as PET. Positron emission tomography yields information on growth rate and heterogeneity of LGGs and is especially useful for follow-up purposes, as metabolic changes tend to precede structural changes detected with imaging methods based on structures. Furthermore, for planning of LGG surgery or radiotherapy, co-registration of functional images with CT and MRI is invaluable. This is increasingly performed with the new generation of hybrid scanners with integrated PET and CT [24].

For pre-surgical evaluation of refractory epilepsy, there is some evidence of diagnostic accuracy. Evidence, although limited, seems supportive for the use of PET. However, this is a rare indication [25].

4. QUALITY ASSURANCE IN PET/CT

4.1. INTRODUCTION

In order for QC tests of PET and PET/CT scanners to be effective, operational and technical aspects of their use need to be governed by a quality management system (QMS). Management aspects also need to be included that ensure that all procedures related to image quality and radiation dose to patients are properly addressed and documented. Possible problems and malfunctions also need to be addressed. The QMS should include several basic components:

(a) A clear definition of responsibilities for the defined actions regarding quality assurance (QA);
(b) A series of documents illustrating correct use of the imaging equipment, and of test objects, phantoms and sources, detailing test modalities and procedures to follow in the case of abnormal results that do not correspond to what is expected or in the case of malfunction.
(c) Records of all tests, calibrations and corrective actions performed.

(d) Proper training of all the staff involved in the correct and safe use of the equipment, its QC procedures and all aspects pertaining to QA.

The QMS control life cycle regarding medical imaging equipment is described in Fig. 10, which is based on the IEC 61223-1 standard [26]. On this basis, it should be clear that QA and QC do not merely consist of simply performing routine tests during the operation of the equipment. A proper QMS should also include the specification and acquisition phases, and thus starts well before the actual installation and operation of the equipment.

In particular, the process of specification and acquisition should involve a multidisciplinary team of professionals in order to properly define the needs of an institution and prepare the technical specifications that the proposed equipment should meet in order to ensure satisfactory results.

In the case of PET or PET/CT scanners, the team of professionals should include at least the following:

(a) A nuclear medicine and radiology physician;
(b) A medical physicist with expertise in diagnostic radiological physics, especially CT;

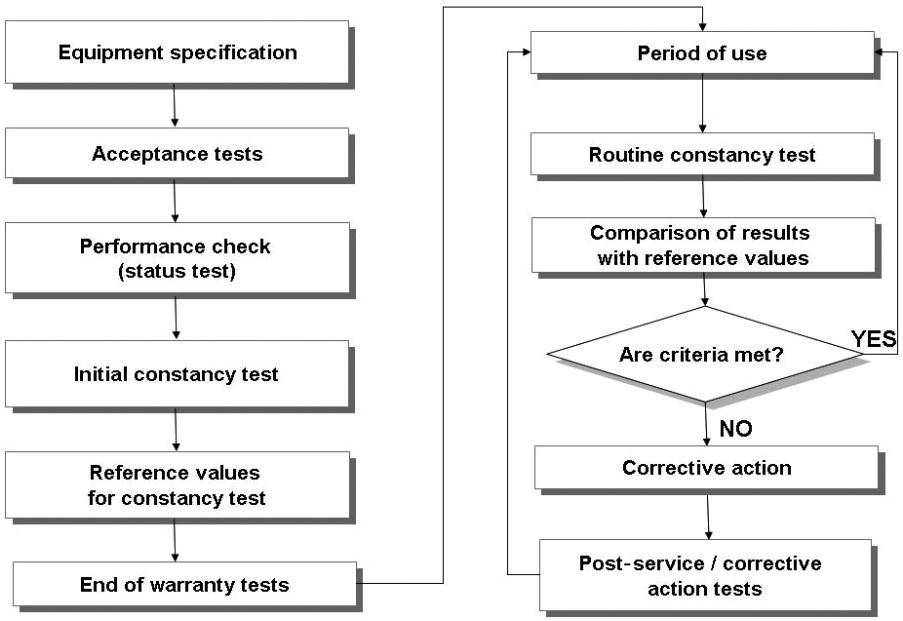

FIG. 10. Quality assurance and QC cycle for a medical imaging device (based on information from Ref. [26]).

(c) A medical physicist with expertise in nuclear medicine;
(d) A medical physicist specializing in radiation oncology physics (if the PET/CT images are to be used for radiation treatment planning);
(e) Facilities management staff;
(f) An architect;
(g) A radiation protection expert;
(h) A person qualified in radiochemistry or radiopharmacy (in the case of in-house production of radiopharmaceuticals);
(i) A nuclear medicine technologist, also trained in CT technology;
(j) A hospital management expert.

The specification document should include information regarding acceptance and end-of-warranty testing, so that the vendor understands the requirements and schedule. In addition, the manufacturer should ensure that a service engineer is present during acceptance and end-of-warranty testing to correct any problems encountered by the medical physicists doing the testing. Once the equipment has been adequately specified, identified and purchased, the equipment must be properly installed. Acceptance tests shall then be performed, preferably by a qualified independent physicist, in order to verify that the scanner meets all the requirements in terms of performance and operational parameters.

Some confusion exists about the differences between QA and QC. Quality assurance refers in general to the concept of taking actions to ensure that delivered products or services meet performance requirements. The QMS is the programme that controls how quality is maintained and ensured throughout an organization. Quality assurance may encompass various aspects such as quality of medical care based on specific indicators, for example:

— The infection rate in the hospital;
— The satisfaction of patients with their care;
— The credentials of the medical staff;
— Any continuing education of the hospital staff.

The QMS defines what steps will be taken to ensure that the desired level of care is maintained and how these will be documented. Quality control for PET/CT applies to a specific set of measurements focused on monitoring the performance of installed imaging equipment relative to image quality and dose on a periodic basis, for example, monthly.

The present publication is intended to assist with the process of acceptance testing and QC of PET scanners. It supplements the material found

in international and national standards, such as the IEC and NEMA publications [27, 28], as well as other relevant documents referenced in this publication. To establish reference values and action levels to compare with the results of routine tests, an initial series of QC tests must be performed immediately after completion of the acceptance procedures. During the operational life cycle of the equipment, regular QC tests should be performed as described in Section 4 of this publication and the other sources of information indicated above.

4.2. INTRODUCTION TO ACCEPTANCE TESTING

Acceptance testing of medical imaging equipment serves several purposes, it:

(a) Ensures that equipment (both hardware and software) performs to the manufacturer's specifications prior to final payment for the equipment;
(b) Establishes the baseline performance of the equipment to which future quality tests will be compared;
(c) Provides data that can give guidance in the determination of the optimal operating parameters for routine use;
(d) Ensures that the imaging equipment meets regulatory requirements for radiation safety.

In addition to the traditional acceptance tests, a second set of tests (the same as the acceptance tests) should be carried out about one month before the end of the warranty period. The equipment should perform similarly to its performance during acceptance testing. If not, the manufacturer should be advised immediately and be required to make the appropriate calibrations and modifications to ensure that the system again meets the manufacturer's specifications.

The following sections cover acceptance testing of PET imaging systems, CT scanners and PET/CT combined systems. In all remaining parts of this publication, the term 'PET/CT' will be used to refer generically to these systems. The terms 'PET' and 'CT' will be used when discussing aspects of acceptance testing specific to these modalities.

4.2.1. Defining acceptance testing

The American Association of Physicists in Medicine (AAPM), in its report by the Imaging Committee Task Group 2 [29], defines acceptance testing as follows:

> "The acceptance test is a series of measurements performed by the clinical medical physicist to verify that a CT system conforms to vendor technical specifications or to specifications mutually agreed upon by buyer and vendor. Often a proviso for acceptance testing is written into the bid request which indicates who will do the testing, what tests will be performed, and what level of performance is acceptable to the buyer."

Modern medical imaging systems provide more potential combinations of operating conditions than can be practically tested. Acceptance testing must acquire sufficient data to adequately characterize system performance. Consequently, it is essential to review clinical needs, scanner specifications and design principles, to determine the most important performance variables that should be tested, and to include only those that could affect important performance characteristics.

It is important to select a short list of four or five standard scanning conditions that are the most relevant to clinical imaging with the system. For CT applications, this will most often include CT of an adult's head and body, and paediatric body imaging. If thin slice imaging (e.g. 1 mm slice imaging) is not anticipated for PET/CT imaging, then thicker slice imaging modes should be the focus of acceptance testing. If all scans are in helical mode, then acceptance testing should emphasize helical mode data acquisition.

There are two other tests that fall under the broad heading of acceptance tests: end-of-warranty tests and post-service tests.

4.2.1.1. End-of-warranty tests

As the name implies, end-of-warranty tests are carried out near the end of the warranty period. These tests should be carried out four to six weeks before the end of the warranty, to ensure that the equipment meets the performance determined from the original acceptance testing and the manufacturer's performance specifications. Any measurement that fails to meet either of these two criteria indicates that a change has occurred in the equipment and that the manufacturer should make appropriate repairs at no cost to the customer as soon as possible — before the end of the warranty period. Consequently, it is essential to notify the manufacturer (in writing, including the date of the

notification) of such failures sufficiently in advance of the expiry of the warranty period, to allow for repairs to be performed and also for subsequent testing by the medical physicist to confirm that the problem has been resolved. The intention to conduct end-of-warranty tests should be included as part of the purchasing agreement.

4.2.1.2. Post-service tests

When any service is performed on medical imaging equipment, there is the potential that significant changes may occur to the hardware or software. Such changes may result from calibration of the equipment, repair or replacement of components, or upgrades to the system. Consequently, it is necessary to carry out tests after equipment service and before the equipment is placed back in clinical use.

The post-service tests do not need to cover all of the equipment aspects evaluated in the initial acceptance tests. For example, if an X ray tube is changed, one needs to focus the acceptance testing on the radiation producing aspects of the equipment. In this case, one should check whether the radiation exposure level produced is the same as that prior to replacement. If not, one should check whether the kVp calibration has changed, if appropriate filtration has been replaced by the service engineer, or if the radiation dose profile (pre-patient collimator width) is the same as before. Because it is not affected by a change in the X ray tube, there is no need to evaluate the image display, the hard copy printing or any other parts of the system that are not impacted by such a change.

4.2.2. Responsibilities for acceptance testing

Acceptance testing is a very important part of the acquisition process for medical imaging equipment. PET/CT scanners combine sophisticated features of two complex imaging modalities and must be properly maintained and monitored to ensure their correct operation.

A qualified medical physicist, or physicists, must be responsible for acceptance testing of a PET/CT system. It is often very difficult to find a single medical physicist who is experienced in the technical aspects of both PET and CT. Consequently, it may be necessary to have two physicists working together in the acceptance testing of a PET/CT system: one an expert in PET technology (medical physicist qualified in nuclear medicine) and the other an expert in CT technology (a medical physicist qualified in diagnostic radiology). Furthermore, if the PET/CT system data are to be used for radiation therapy treatment planning, then the PET and CT medical physicists should consult

with and work together with a medical physicist qualified in radiation therapy who is familiar with the application of the data for treatment planning purposes. In order to ensure that the scanner performance truly meets the customer's specifications, it is strongly advisable for the acceptance tests to be performed by a qualified medical physicist who is independent of the vendor.

Qualified medical physicists are usually recognized by their professional organizations through certification and continuing education in clinical medical physics. Qualified medical physicists evaluating PET/CT systems must be recognized for their competence in nuclear medicine physics and diagnostic medical imaging. The qualified medical physicist must be familiar with:

— The principles of radiation protection;
— The guidelines of national radiation protection organizations;
— The laws and regulations governing the use of the equipment being tested;
— The function, optimal clinical uses and performance specifications of the imaging equipment;
— Calibration processes and limitations of the measurement instruments;
— The techniques used for testing performance.

The qualified medical physicist(s) is/are responsible for acceptance testing and annual testing of the equipment, and must review, interpret and approve all data, as well as summarize the tests performed and provide conclusions. Even if the tests are performed by an independent physicist, the responsible physicist from the purchasing institution must be present during the tests. The test report should be signed by the responsible medical physicist(s) of the purchasing institution, and, if applicable, by the physicist who performed the tests.

While the medical physicists are responsible for acceptance testing of the equipment, it is recommended that a qualified service engineer from the equipment manufacturer be on-site during this testing. This will allow for corrections of problems as they are found and reduce the overall amount of time required for acceptance testing.

4.2.3. Sequence of acceptance tests

The sequence in which acceptance tests are carried out can have an impact on the efficiency of testing. Calibrations generally need to be conducted prior to the start of the tests. Electromechanical tests are then typically carried out before other types of test. Discrepancies in the results may affect image quality and dose, and must be corrected before carrying out other tests.

Appendices I and II list the acceptance tests recommended for the different components of a PET/CT imaging system, along with the equipment needed, the values measured and the acceptance criteria, where appropriate.

4.3. QUALITY CONTROL

Quality control helps to ensure that the equipment performs, throughout its useful life, at the levels specified by the manufacturer and measured during the acceptance testing process and that there is ongoing compliance with regulatory requirements for radiation safety.

4.3.1. Post-service testing

Post-service testing is an important part of the QA and QC programme and is always required when there are changes made to the equipment that could affect its performance. When there are changes to the equipment, the functions of the equipment potentially affected by these changes must be evaluated, to ensure that the equipment once again performs at the levels determined during acceptance testing. Many types of changes will require post-service testing, some of which are not obvious. Equipment calibration may result in changes to many aspects of the imaging equipment. Software upgrades can affect X ray tube output, system sensitivity, image quality and other aspects of the function of the system. Software upgrades frequently require that all acceptance tests be carried out, since such upgrades can have an impact on virtually every aspect of the system.

As an example, when an X ray tube is replaced, it is essential to determine if the patient dose is similar to that used previously. If the dose has changed significantly, then the reason why this has occurred must be determined. Is the kVp value the same? Did the service engineer replace the appropriate aluminium filtration unit in the X ray beam?

Whatever is the case, the service engineer must ensure that the output is similar to that prior to the tube replacement. Consequently, it is beneficial if these tests can be carried out while the service engineer is still on the site.

4.3.2. Equipment required for quality control testing

Appendices I and II provide lists of the equipment and materials needed for each of the acceptance and QC tests. Much of this equipment is only needed on an annual basis and is usually brought to the site by the consulting medical physicist or by the manufacturer (in the case of acceptance testing).

All test equipment needed for daily or monthly tests must be available at the facility for use by the person performing the QC tests.

4.3.3. Control charts

One key to a useful QC programme is the use of control charts. These types of diagram involve plotting a quantity of interest (e.g. energy or spatial resolution and X ray dose) as a function of time in order to check for trends. Such trends in the data, either monotonic or oscillatory, can be an early indication of problems developing in the system. By monitoring over long time periods, effects might be revealed that might not be evident when merely looking at small changes over short time periods.

In any control chart, there are three primary values of significance, which are:

(1) The operating level — This is the ideal value for the measurement being made, and is generally measured during acceptance testing.
(2) The upper control limit (UCL) — This is the value which, if exceeded, requires immediate corrective action.
(3) The lower control limit (LCL) — This is similar to the UCL, and is the value for which immediate correction action is required if the measured value falls below the LCL.

To be effective, data must be plotted on the control chart immediately after a measurement has been made. It is quick and easy to then see if the measured value exceeds the UCL or the LCL, and to see if any trends in the data are apparent. If the limits are exceeded or if trends are apparent, corrective action must be taken in accordance with the QMS of the institution. Corrective action usually includes repeating the test to ensure that the result obtained was real and that there was not an error in the measurement. If the measured value is still out-of-bounds (beyond either the UCL or the LCL), then it will be necessary for the medical physicist to determine the appropriate course of corrective action.

It should be noted that even in the absence of true changes in the system performance, there is some degree of variability that will appear in the measured values over the course of time. The magnitude of the variability should be apparent in the early stages of operation as data are acquired.

The basics of the use of control charts are explained, and several examples are given, in the 'Radiologic Technologist's' section of the Mammography Quality Control Manual [30] published by the American College of Radiology (ACR).

4.3.4. Responsibilities for quality control tests

The qualified medical physicist is responsible for the overall supervision of the QC programme, including supervision of tests performed by other professionals (e.g. technologists[1]). As such, the physicist must ensure that the technologist is able to carry out the tests and interpret the results, to request follow-up testing by the physicist, or to request corrective action when needed. For a combined PET/CT system, the responsible physicist must be qualified to provide physics support for both imaging modalities. As indicated, many of the tests described in Appendices I and II can be done by a PET/CT trained technologist, but the overall responsibility remains with the qualified medical physicist.

Physicists in nuclear medicine normally have other responsibilities in addition to acceptance testing and QC, such as supporting diagnostic and therapeutic procedures, teaching, radiation protection, computer system administration, and development [31–36]. However, they should be available for immediate consulting on matters regarding QC. It is therefore recommended that a physicist be available on-site at the PET or PET/CT facility.

Tests are carried out by the physicist when the equipment is installed (acceptance tests), at the end of the warranty period, on an annual basis (QC tests) and after the equipment has been serviced. It is necessary for the physicist to carry out post-service testing to ensure that changes to the equipment made by the service engineer do not affect either image quality or patient doses. In addition to annual testing, the physicist should review the technologist's test results at least quarterly and be available for consultation with the technologist as needed.

The technologist is responsible for carrying out the daily and monthly QC tests. If the tests indicate significant changes in the parameters that affect image quality or patient dose, the technologist should contact the physicist for consultation. As an example, if the noise changes significantly on the CT images, or if abnormal variations in the detector response appear in the daily PET QC tests, the technologist should discuss this with the physicist, as this may indicate a change in system operation. The requirements (action limits) for contacting the

[1] Throughout this publication, the term 'technologist' has been adopted as a generic descriptor for the person normally associated with the job titles of 'CT radiographer' or 'nuclear medicine technologist', who is trained in both the CT and PET modalities.

medical physicist in this regard should be determined by the medical physicist and be documented in the operating procedures.

In addition to periodic tests, the technologist is responsible for notifying the physicist every time that equipment is serviced (preventive maintenance, calibration or repairs) or that there are software upgrades. The physicist will then determine if the technologist can carry out tests to determine the impact of the equipment service or upgrades on image quality and patient dose, or if the physicist must carry out on-site testing.

4.3.5. Frequency of quality control tests

The frequency of QC tests is typically specified along with a test procedure. However, the frequency can be modified by the medical physicist on the basis of the performance and reliability of the equipment, the criticality of the application of the equipment and the results of earlier QC tests.

All diagnostic imaging equipment requires a certain amount of calibration, preventive maintenance and component replacement. There is usually no impact on these from QC programmes. When a component fails, it will fail whether it is monitored with a QC programme or not. For example, QCs on the CT photographic processor are almost universally required on a daily basis, before processing any clinical films. However, review of six months of control charts indicates that the monitored values never shift by more than ±0.05 optical density units. Furthermore, all the processing chemicals are mixed in-house in large batches, and each batch of chemicals is tested sensitometrically. Under these conditions, it is acceptable to reduce the frequency of testing, perhaps to one to three times per week, with one test always being carried out before processing of any clinical images at the beginning of the working week.

As noted above, the frequency also depends on the criticality of the application. As another example, it is recommended to check the gantry lasers daily. Even although there may not have been any problems with the alignment of the lasers for six months, it is probably not prudent to reduce the frequency of this check if the data are to be used for radiation therapy applications. If there were to be a shift in the laser alignment, this could result in an error in the location of the treatment volume when the images are used for radiation therapy treatment planning. This is a critical application and, therefore, one for which daily checks are warranted.

4.3.6. Important points

While all QC tests are important to ensuring optimum image quality and patient radiation dose, two procedures require special emphasis: artefact evaluation and QC tests for equipment used for radiation therapy.

Image artefacts can be created by several mechanisms, including mismatch of the CT transmission attenuation map with PET data, patient motion, external radioactive contamination, presence of prostheses and implants in the patient and detector malfunction. Artefacts create poor quality images and can lead to incorrect diagnosis or staging if left unaccounted for. It is essential during acceptance testing and QC tests to *evaluate each and every image* for obvious artefacts. Uniform phantom images often do not demonstrate artefacts as clearly as images of a spatial resolution phantom or a contrast scale phantom. Whenever an artefact is suspected in a PET/CT reconstructed image, it is always advisable to also check the PET image reconstructed without attenuation correction, to determine whether the problem is of PET or CT origin.

If PET/CT equipment is used for radiation therapy treatment planning, some of the QC tests become critical to the correct treatment of the patient. The most critical in this respect is the QC test of the scan localization lasers, i.e. the lasers on the CT gantry that are used to locate the patient relative to the image data volume. The location of the image data volume must be known precisely for transfer to the radiation therapy treatment planning computer system. This daily test consists of a simple and quick visual check.

The medical physicist's annual tests are important as part of the QMS programme. In addition, it is essential that the medical physicist carry out a visual inspection and a programme review on an annual basis. As part of the visual inspection, the medical physicist will look for issues that result in risks for patients and staff. These may include a lack of auxiliary shielding or aprons, and sharp corners or covers that do not close properly.

A review of the programme is also an important part of annual testing. This should include, but not be limited to:

(a) Ensuring that appropriate scan protocols are available in writing for the technologist and that these protocols have been programmed into the scanners;
(b) A review of policies and procedures, for example, those for assisting patients during a scan;
(c) Other safety related issues.

4.3.7. Quality control records

A critical element of any QMS is maintenance of a complete set of records. The primary component is a record of the tests that were performed, the time and date they were performed, a brief summary of the results of the tests or a comment about unusual findings, and the name of the person who performed the tests. The images and other appropriate data must be kept in a binder for easy access and review.

4.3.8. Testing of TOF-PET

At present, not all manufacturers have TOF scanners that are a commercial product with QC and QA procedures specific to the TOF component. At the present time, a comprehensive description of TOF testing procedures is premature, since no consensus has been reached on the optimal performance assessment of TOF systems and what parameters are pertinent for such an evaluation. In this publication, limited testing is proposed to help users assess the performance of TOF-PET systems. A more comprehensive description will be considered in a future version of this publication, if TOF scanners become more widespread and baseline values of the principal characteristics of TOF scanners become better established.

4.4. PREVENTIVE MAINTENANCE

Preventive maintenance should be scheduled on a regular basis and can be done by the manufacturer's agent, a third party service organization or trained in-house maintenance staff. This action should put the instrument into its best possible working condition and identify potential problems before major breakdowns occur. In addition, staff must note potential safety problems such as frayed cables or unusual noises, and notify the responsible person (e.g. a qualified medical physicist).

5. ACCEPTANCE TEST PROCEDURES

5.1. PET ACCEPTANCE TESTING

The acceptance testing of a new PET/CT system involves the comparison of the test results with specifications provided by the manufacturer. It is recommended that acceptance tests be performed by the qualified medical physicist and not by the vendor or its representative. The latter does not provide a verifiably independent measure of system performance.

Users of this publication are advised to establish, in consultation with the manufacturer, the precise acceptance values and test procedures that will be used before commencing acceptance tests, and ideally prior to purchasing the system.

As PET/CT manufacturers typically specify PET performance parameters that are in conformance with the NEMA specifications, the PET acceptance tests described in this publication are intended to conform, where applicable, to the NEMA or IEC standards.

For the acceptance tests, it is recommended to obtain the following information in advance from the manufacturer:

— Documentation on test procedures, including a recommendation of the amounts of radioactivity to be used at the beginning of each test;
— Phantom preparation;
— Data acquisition and data analysis;
— Any special equipment required to perform the tests.

It is also advisable to prepare a schedule of testing and radioactivity requirements in advance of testing. The manufacturer is expected to provide all software to perform the acceptance tests, and all tools needed to perform the calibration of the equipment. In some of the test procedures that follow, indicative phantom and source activities are suggested. However, since the optimal radioactivity for each test varies from one system to another, the activities recommended by the PET system manufacturer should be used wherever possible.

The preparation of radioactive sources is an integral part of PET acceptance testing, and requires, for example, accurate recording of times of assays and scan start times. As timing errors can significantly affect results, it is strongly recommended to synchronize wristwatches and clocks with the time shown on the PET/CT computer console before commencing acceptance testing. Source preparation should be done under the supervision of a qualified

medical physicist observing proper radiation protection precautions, to ensure that local radiation safety regulatory requirements are complied with.

Before commencing acceptance testing, it is important that all the calibrations required as part of the installation and commissioning of the scanner have been performed to ensure that the scanner is operating as expected. It should also be verified that the system has passed the daily QC and that there are no problems apparent in the sinograms.

Depending on the scanner, acceptance testing can be completed in three to five days. However, it is prudent to allow at least one week for acceptance testing, to allow time to repeat some tests if required and also to cope with late or non-delivery of ^{18}F radioactivity. It is useful to perform the tests in conjunction with the installation or service engineers, as they are familiar with the system software and can provide access to service menus if necessary. This also avoids duplication of tests being done by the service engineers as part of their testing and verification procedure, and then again by the physicist for final acceptance testing of the scanner. This is best discussed with the vendor prior to installation of the scanner.

5.1.1. Spatial resolution

5.1.1.1. Aim of the test

The aim of this test is to measure the tomographic spatial resolution of the system in air and to ensure that spatial resolution is not degraded by either the tomographic acquisition or the reconstruction process. Tomographic resolution indicates the system's ability to distinguish between two points after image reconstruction and is an important factor in determining the size of a lesion that can be detected. A measurement in air indicates the highest achievable performance. However, it should be noted that spatial resolution is affected by the point at which it is measured and the direction in which it is measured (x, y or z).

This test is based on the NEMA NU2-2007 spatial resolution test [37].

5.1.1.2. Frequency

The spatial resolution test must be carried out by the qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty tests and whenever it is suspected that the performance of the detector system may have changed significantly; for example, after major servicing of the PET detector (e.g., replacement of a large number of detector blocks).

5.1.1.3. Materials

There are three point sources of ^{18}F, with a spatial extent of less than 1 mm in both the transaxial and axial directions. Adequate sources can be prepared using a capillary tube, with an inside diameter of less than 1 mm and an outside diameter of less than 2 mm (Fig. 11). The axial extent of the radioactivity in the capillary tube must be less than 1 mm. (Although some vendors specify an axial length of more than 1 mm for this test and recommend rotating the source through 90°, a length of less than 1 mm is recommended as it allows measurement of both the axial and transaxial resolutions without rotating the source.) In the transverse plane, the sources should be placed in three positions (Fig. 12):

(1) One centimetre vertically from the centre of rotation (to represent the centre of the FOV, but positioned to avoid any possibly inconsistent results at the very centre of the FOV — the 'sweet spot');
(2) At $x = 0$ cm and $y = 10$ cm;
(3) At $x = 10$ cm and $y = 0$ cm.

The sources should be suspended in air, to minimize the effect of scattered radiation. It is recommended to use or construct a source holder to hold the sources securely in the correct positions.

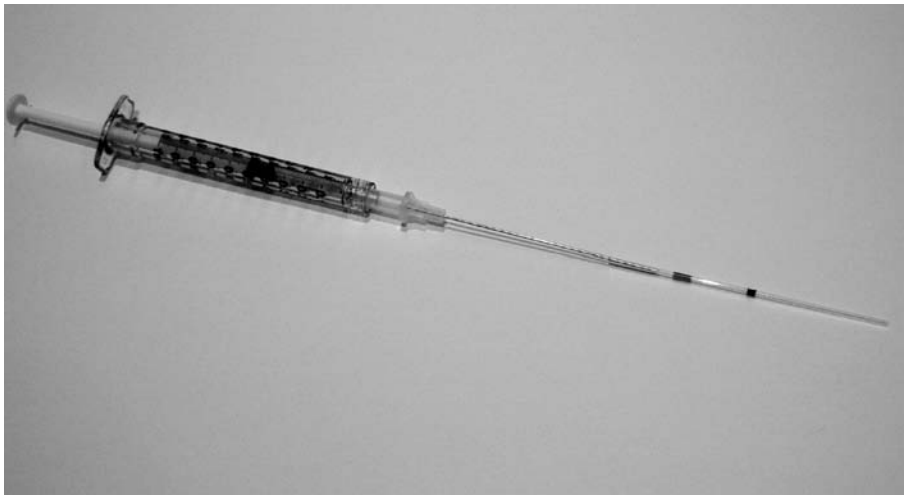

FIG. 11. A syringe of the type with a fine needle to put a small amount of ^{18}F solution inside a capillary tube (on right) for the measurement of spatial resolution.

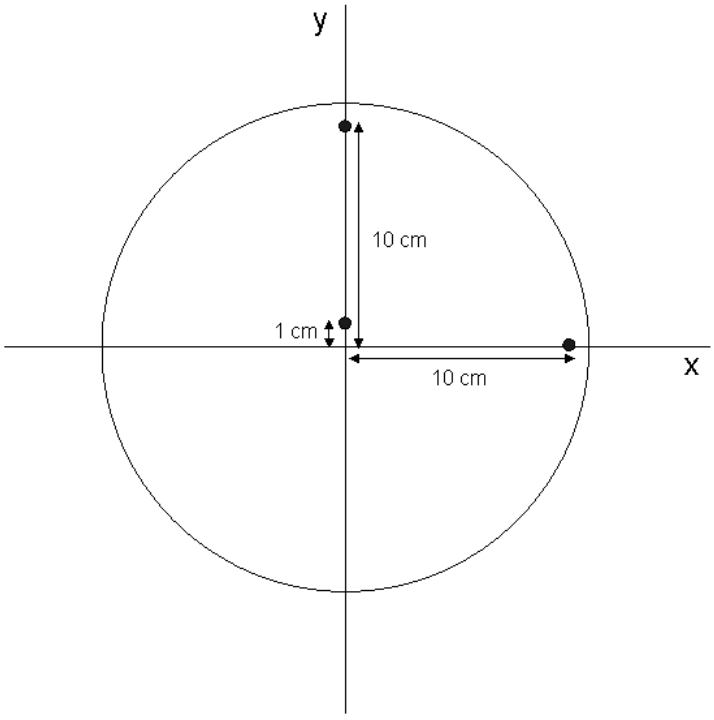

FIG. 12. Position of point sources for measurement of spatial resolution.

The radioactivity of the sources shall be such that when all the sources are inside the FOV the percentage of dead time losses and randoms must both be less than 5% of the total event rate. Typically this can be obtained with a radioactivity of the order of 1 MBq. The radioactivity concentration of the starting radioactive solution should thus be about 1000 MBq/mL (or 27 mCi/mL) or less. It is stressed that for this and other tests, manufacturer recommended activities, if available, may depart from these general recommendations and should be used instead.

5.1.1.4. Data acquisition

In the case of tomographs with significant natural radioactivity in the detector material, the above condition pertaining to randoms as a percentage of the total event rate may not be achieved [38]. For such systems, acquisition of spatial resolution should be performed using delayed window random event correction.

For accurate measurement of the spatial resolution, the reconstructed image pixel size should be set at one third of the expected system FWHM in all three dimensions. For whole body scanners, this typically translates into less than 1.5 mm/pixel. In some cases, it may be necessary to use a zoom factor to obtain satisfactory sampling.

Two separate acquisitions should be performed: in the centre of the axial FOV and at an axial position of a quarter of the FOV.

At least 100 000 counts must be acquired for each response function (point) or, in the case of tomographs with significant natural radioactivity in the detector material, a total of 120 000 counts should be acquired, to account for reduced count statistics after correction for randoms.

The acquisition should be repeated at the same source positions in two dimensions and three dimensions for scanners that have both capabilities. Reconstruction should be carried out using filtered back-projection with a ramp filter; no further smoothing or apodization should be applied.

5.1.1.5. Analysis

For each acquisition position, transaxial and sagittal images should be obtained.

Profiles across the point source response functions in all three directions (radial, tangential and axial) will be generated. The width of the profiles in the two directions at right angles to the direction of measurement will be approximately two times the expected FWHM.

The maximum value of the profile will be determined by a parabolic fit to three points: the peak point and its two nearest neighbours. The FWHM and full width at tenth maximum (FWTM), shown in Fig. 13, for all of the point source response functions in all three directions (radial, tangential and axial) will be calculated using linear interpolation (18 numbers). The calculated FWHM and FWTM values will be converted to millimetres by multiplying by the pixel size.

The radial and tangential resolutions will be averaged using the formulas given in Table 3.

5.1.1.6. Suggested tolerances

Calculated values of FWHM should not exceed the specification given by the vendor. Values for the FWTM are usually not specified; an expected value can be estimated taking into account the fact that for a theoretical Gaussian curve, the ratio between the FWTM and the FWHM is 1.82. Therefore, the

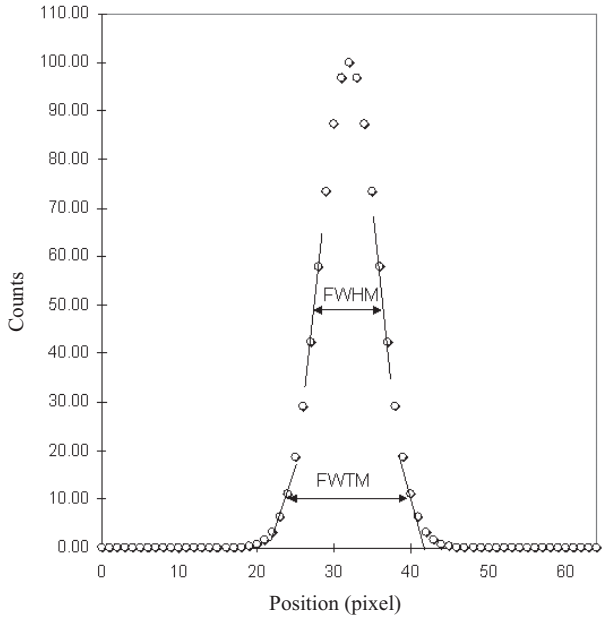

FIG. 13. Example of a response function, with the definitions of FWHM and FWTM.

expected ratio between FWTM and FWHM for an actual PET scanner should be approximately in the range 1.8–2.0.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criterion for FWHM is

$$FWHM_{observed} < 1.05\ FWHM_{expected}$$

5.1.1.7. Corrective action

If tolerance criteria are exceeded for FWHM, some action should be taken. First, the results should be checked and the testing procedure repeated to ensure that the source was properly prepared; next, the manufacturer should be notified and corrective action requested.

TABLE 3. FORMULAS FOR COMPUTING SPATIAL RESOLUTION.

	Description	Formula
At 1 cm radius		
Transverse	Average x and y for both z positions	$RES = (RESx_{x=0,y=1,z=centre} + RESy_{x=0,y=1,z=centre} + RESx_{x=0,y=1,z=¼FOV} + RESy_{x=0,y=1,z=¼FOV})/4$
Axial	Average two z positions (two numbers)	$RES = (RESz_{x=0,y=1,z=centre} + RESz_{x=0,y=1,z=¼FOV})/2$
At 10 cm radius		
Transverse radial	Average two transverse positions for both z positions (four numbers)	$RES = (RESx_{x=10,y=0,z=centre} + RESy_{x=0,y=10,z=centre} + RESx_{x=10,y=0,z=¼FOV} + RESy_{x=0,y=10,z=¼FOV})/4$
Transverse tangential	Average two transverse positions for both z positions (four numbers)	$RES = (RESy_{x=10,y=0,z=centre} + RESx_{x=0,y=10,z=centre} + RESy_{x=10,y=0,z=¼FOV} + RESx_{x=0,y=10,z=¼FOV})/4$
Axial resolution	Average two transverse positions for both z positions (four numbers)	$RES = (RESz_{x=10,y=0,z=centre} + RESz_{x=0,y=10,z=centre} + RESz_{x=10,y=0,z=¼FOV} + RESz_{x=0,y=10,z=¼FOV})/4$

5.1.2. Sensitivity

5.1.2.1. Aim of the test

Tomographic sensitivity relates the count rate measured by the device to the amount of radioactivity within the FOV. The purpose of sensitivity measurement is therefore to determine the rate of detected true coincidence events per unit of radioactivity concentration for a standard source configuration, for example, a cylindrical phantom of given dimensions. In order to be able to compare sensitivity measurements between scanners, these values should be free of confounding effects such as attenuation, scattering and count rate distortions.

In PET, the measurement of attenuation-free radioactivity is complicated by the fact that positrons need a certain pathway in absorbing matter to be converted into gamma radiation. The test takes this into account by utilizing measurements with increasing thickness of absorbing material and extrapolating the measurements to zero absorption [39]. The test requires special equipment consisting of a set of aluminium tubes of varying diameter.

This test is based on the NEMA NU2-2007 sensitivity test [37].

5.1.2.2. Frequency

The sensitivity test must be performed by the qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty tests and whenever it is suspected that the detector system performance may have changed significantly.

5.1.2.3. Materials

The source used is a line source, 700 mm long, uniformly filled with a radioactivity such that count losses are less than 1%, and the random event rate is less than 5% of the true event rate. For dedicated 3-D PET scanners, this is achieved with a radioactivity of about 5 MBq of ^{18}F, but this depends on the sensitivity of the scanner. The vendor is expected to specify an appropriate value of radioactivity for testing. As specified in NEMA NU2-2007 [37], for accurate estimation of the sensitivity the length of radioactivity in the line source should be 700 ± 5 mm.

The radioactivity of the source, A_{cal}, should be accurately measured in a radionuclide radioactivity calibrator (dose calibrator) and the time of measurement, T_{cal}, recorded. The calibration and accuracy of the dose calibrator for measuring ^{18}F must be established prior to performing this test.

The phantom for sensitivity measurements is completed by a set of five sleeves consisting of aluminium tubes 700 mm long, each with a wall thickness of 1.25 mm, with increasing diameters according to Fig. 14 and Table 4.

5.1.2.4. Data acquisition

The phantom is positioned in air, supported at each end by low density materials to minimize scatter, in the centre of the transaxial FOV.

Starting with the smallest sleeve containing the line source only, perform an acquisition such that at least 10 000 true events per slice are collected.

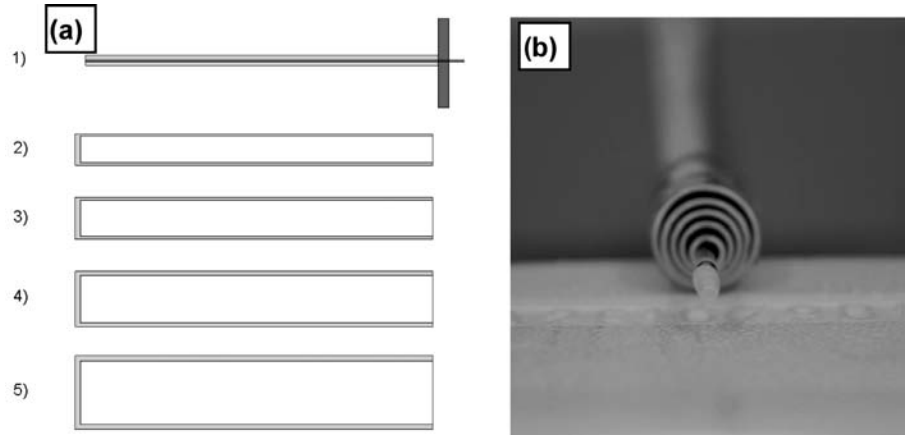

FIG. 14. A sensitivity measurement phantom: (a) schematic diagram, (b) actual phantom viewed end-on.

Increase wall thickness by adding the next smallest sleeve and repeat the acquisition, until acquisitions with all sleeves have been made.

To check the variation of sensitivity within the FOV, perform acquisitions as above for phantoms and line sources at 10 cm offset from the central axis.

For systems capable of acquisition in both 2-D and 3-D, measurements should be performed in both modes.

On systems that can provide measurements of the randoms rate by acquiring data in separate prompt and delayed coincidence windows, the randoms rate may be subtracted, thus permitting the true-events-only

TABLE 4. DIMENSIONS OF THE SENSITIVITY MEASUREMENT PHANTOMS

Tube No.	ID[a] (mm)	OD[b] (mm)	Thickness (mm)	Length (mm)
1	3.9	6.4	1.25	700
2	7.0	9.5	1.25	700
3	10.2	12.7	1.25	700
4	13.4	15.9	1.25	700
5	16.6	19.1	1.25	700

[a] ID: internal diameter.
[b] OD: outer diameter.

sensitivity to be reported. In the case of tomographs with significant natural (intrinsic) radioactivity in the detector material, the assumption that low count rate acquisitions contain a negligible randoms rate is not appropriate, and it is essential that the sensitivity calculations below be performed using count rate data from which randoms have been subtracted. The actual level of radioactivity to be used for testing should be suggested by the vendor.

5.1.2.5. Analysis

Single slice rebinning should be used to assign counts in oblique LORs to the image slice where the LOR crosses the scanner axis. The time of the commencement of a measurement, T_j, and the duration, T_{acq}, including the time required to move the detectors (in the case of scanners that require detector motion to acquire a full 3-D data set) are recorded, along with the number of counts collected.

The rate in counts per second, $R_{j,i}$, shall be determined for each measurement associated with each of the five sleeves, designated by the index j (j = 1–5), and for each slice, designated by the index i (i = 1 to the number of slices), by dividing the number of counts collected in the sinogram of the slice by the duration T_{acq}.

For each measurement associated with each of the five sleeves and for each slice, the count rate for isotope decay will be corrected for radioactive decay using the following formula:

$$R_{CORR,j,i} = R_{j,i}\exp[(T_j - T_{cal})/T_{1/2}] \quad (6)$$

where T_j is the time of the jth acquisition and T_{cal} is the time of phantom radioactivity calibration. After decay correction, the cumulative count rate is calculated using the following expression:

$$R_{CORR,j} = \Sigma_i R_{CORR,j,i} \quad (7)$$

for each accumulated sleeve thickness. The data are then fitted to the following equation:

$$R_{CORR,j} = R_{CORR,0}\exp(-2\mu_M X_j) \quad (8)$$

where $R_{CORR,0}$ represents the unattenuated count rate. The linear attenuation coefficient of the sleeve material, μ_M, is allowed to vary to compensate for scattered radiation, and X_j represent the accumulated sleeve wall thicknesses. The fitting procedure yields estimates of $R_{CORR,0}$ and μ_M.

The system sensitivity, S_{tot}, is then obtained by dividing the unattenuated count rate, $R_{CORR,0}$, by the total radioactivity, A_{cal}:

$$S_{tot} = R_{CORR,0}/A_{cal} \tag{9}$$

The same procedure is followed for the measurements obtained when the phantom and line source are offset 10 cm from the central axis.

The axial sensitivity profile is calculated using the data from the acquisition with the smallest sleeve, for the position at 0 cm offset. Using the corrected count rates, $R_{CORR,1,i}$, for the slices and the total count rate, $R_{CORR,1}$, the axial sensitivity for slice i is obtained using the following formula:

$$S_i = (R_{CORR,1,i}/R_{CORR,1})S_{tot} \tag{10}$$

A sensitivity profile can be obtained by plotting S_i against slice number. Maximum and minimum values can be recorded. Typical plots of axial sensitivity profiles in 2-D and 3-D are shown in Fig. 15.

5.1.2.6. Suggested tolerances

The system sensitivity for 2-D and 3-D modes should be equal to or greater than the vendor's specifications.

The uniformity of the axial sensitivity profile is usually not specified by the vendor.

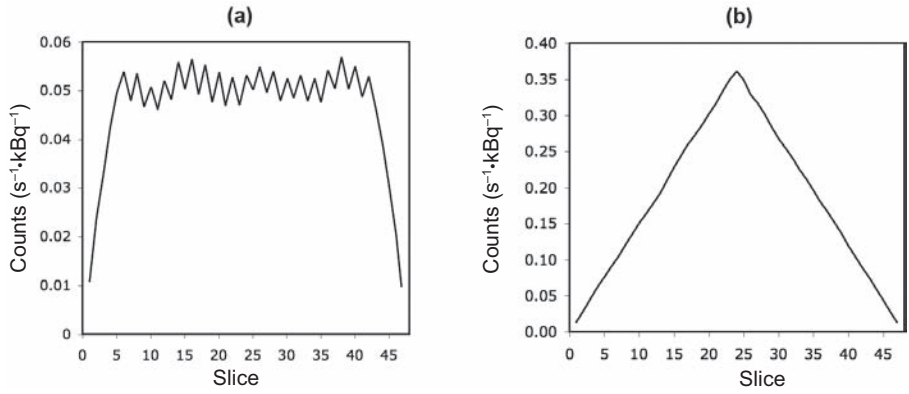

FIG. 15. Typical axial sensitivity profiles in (a) 2-D and (b) 3-D.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criterion for system sensitivity can be:

$$S_{\text{tot,measured}} > 0.95 S_{\text{tot,expected}}$$

5.1.2.7. Corrective action

If tolerance criteria are not met for the sensitivity test, the manufacturer should be notified and corrective action requested.

5.1.3. Scatter fraction, count losses and randoms measurements

5.1.3.1. Aim of the test

Scattering, count losses and randoms affect both image quality and quantitation accuracy.

Scattering and randoms both introduce invalid events. The scatter fraction is defined as the ratio of scatter coincidences to the sum of scattered and true coincidences when random event coincidences are negligible (i.e. at low count rates). This can vary as a function of the design of the tomograph (e.g., its energy resolution, 2-D versus 3-D mode and CTW length). A small scatter fraction is desirable.

Count rate performance reflects the ability of a tomograph to accurately measure high radioactivity sources as well as low radioactivity sources. This is particularly relevant because clinical studies are frequently performed with levels of radioactivity for which count losses due to system dead time are not negligible, while the rate of random coincidences increases with the total single event count rate.

The noise equivalent count (NEC) rate is used to express the tomograph count rate performance as a function of the radioactivity concentration; peak NEC values and the corresponding radioactivity concentration can be used as a guide to determine the optimal radioactivity to be administered to patients in a specific clinical setting. The NEC estimates useful count rates of a scanner by taking into account, assuming Poisson statistics, the contribution of true events and of scattered events and randoms to the total coincidence rate.

This test is based on the NEMA NU2-2007 scatter fraction, count losses and randoms measurement test, which incorporates adaptations for scanners with intrinsic background counts based on those described by Watson et al. [38].

Two methods of analysing and reporting the data are described:

(1) The preferred method, method A, which requires the measurement of random coincidences, by means of either a delayed event window or a calculation based on single detector event rates. This method allows estimates of scatter fraction as a function of count rate. It is the required method for scanners with intrinsic background event counts that cannot achieve a ratio of randoms to true events of less than 1.0%.
(2) An alternative method, method B, for systems that are unable to measure the rate of random coincidences.

The instructions that follow apply to method A, unless otherwise noted.

5.1.3.2. Frequency

This test must be done by the qualified medical physicist at the time of acceptance testing, as part of end-of-warranty tests and whenever it is suspected that the detector system performance may have changed significantly.

5.1.3.3. Materials

The phantom used for testing is composed of a plastic line source, of internal diameter 3.2 ± 0.2 mm (or 1/8 in) and outside diameter 4.8 ± 0.2 mm (3/16 in), and a plastic cylinder (polyethylene, of density 0.96 ± 0.01 g/cm^3) whose dimensions are illustrated in Fig. 16.

The cylinder is traversed from end to end by a hole, 6.4 ± 0.2 mm in diameter, which is parallel to the axis and at a radial offset of 45 ± 1 mm, to contain the line source.

Since the weight of the cylinder is more than 20 kg, it is usually made up of several parts. These parts must be assembled tightly to avoid scatter-free paths through the phantom to the detectors.

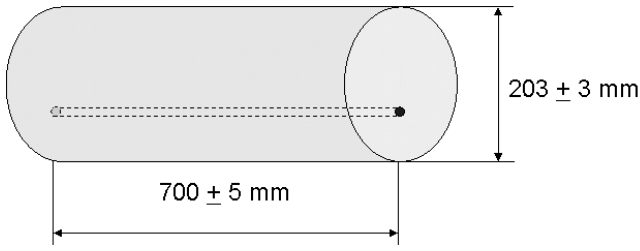

FIG. 16. NEMA scatter fraction phantom.

The radionuclide used is ^{18}F and it should be uniformly distributed in the central 700 ± 5 mm part of the line source. The line source should then be sealed at each end. Note that the results will be affected if radioactivity in the source extends more than 5 mm from the end of the phantom.

The initial radioactivity should be recommended by the manufacturer; relatively high levels of radioactivity are usually specified, to exceed the radioactivity concentration associated with peak NEC. Excessive radioactivity can yield erroneous results on LSO and LYSO scanners due to the crystal afterglow.

The initial radioactivity used to fill the line source is then carefully measured using a radionuclide radioactivity calibrator (dose calibrator), and the time of measurement recorded.

The line source is then inserted into the scatter phantom, placed on the patient bed as shown in Fig. 17, with the line source positioned nearest to the bed. The centre of the phantom must be positioned in the axial and transaxial directions to within 5 mm of the centre of the PET scanner.

5.1.3.4. Data acquisition

Tomographic acquisitions must be performed at time intervals of less than half the half-life of the radionuclide, $t_{1/2}$.

The vendor is expected to provide an acquisition protocol that adequately samples the peak of the NEC curve. Ideally, an acquisition should be made every 15 min or less, around the peak of the NEC curve. To aid in determining

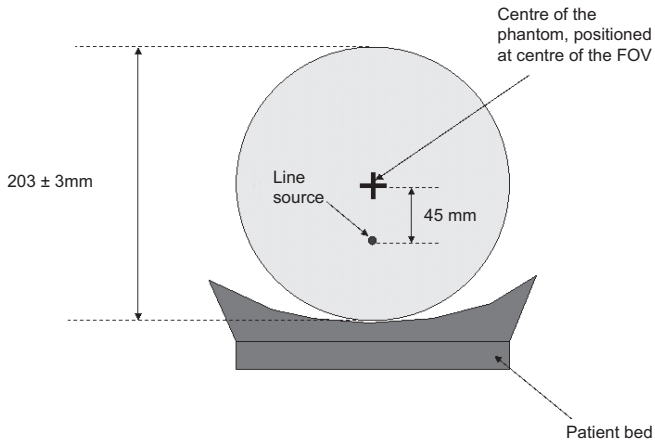

FIG. 17. Positioning of the scatter phantom on the patient bed.

an appropriate sampling frequency, reference can be made to the vendor specification of the expected peak NEC rate. Acquisitions should be performed until true event losses are less than 1%. If method B is to be used, the data must be acquired until the ratio of randoms to true events falls to less than 1.0%.

The durations of individual acquisitions, T_{acq}, should be less than one quarter of $T_{1/2}$, and such that each acquisition accumulates not less than 500 000 prompt counts.

5.1.3.5. Analysis: Trimming of sinograms

If randoms estimation is available (method A), prompt and random sinograms should be generated for each acquisition *j* of slice *i*, for the entire axial FOV, except for scanners with an axial FOV greater than 65 cm, in which case only slices in the central 65 cm should be reconstructed. If no randoms estimate is available, only prompt sinograms are generated. No corrections should be applied for variations in detector sensitivity, motion, randoms, scattering, dead time or attenuation. Oblique sinograms are reformatted into a single sinogram for each slice by single slice rebinning.

For each prompt sinogram *i* of acquisition *j*, all pixels whose distance from the central axis of the phantom is greater than 12 cm should be set to zero (Fig. 18).

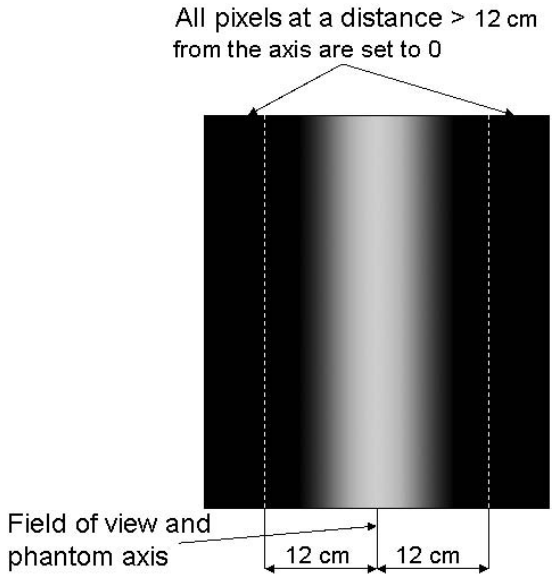

FIG. 18. Schematic example of a sinogram sum profile and definition of the ROI.

For each projection angle, i.e. for each row of the sinogram, the maximum pixel value shall be determined and the projection shifted to align it with the central pixel of the sinogram. A sum projection is then calculated, by summing all the rows of the sinogram (Fig. 19).

In the projection sum profile, counts $C_{Li,j}$ and $C_{Ri,j}$ at a distance of ±20 mm from the maximum pixel are obtained.

The number of scattering and random event counts, $C_{r+s,i,j}$, is then obtained as the sum of all the counts outside the ±20 mm strip and the trapezoidal area delimited by $C_{Li,j}$ and $C_{Ri,j}$ inside the ±20 mm strip. The total event count $C_{Tot,i,j}$ is obtained from the sum of all pixels in the sum projection.

The average radioactivity $A_{ave,j}$ in the phantom for each acquisition j is calculated as:

$$A_{ave,j} = (A_j / \ln 2)(T_{1/2}/T_{acq,j})[1 - \exp(-\ln 2 T_{acq,a}/T_{1/2})] \quad (11)$$

where A_j is the radioactivity at the beginning of the jth acquisition, obtained from the radioactivity measured in the radionuclide radioactivity calibrator at time T_{cal} given as:

$$A_j = A_{cal} \exp[-\ln 2(T_{cal} - T_j)/T_{1/2}] \quad (12)$$

and $T_{acq,j}$ is the duration of the jth acquisition.

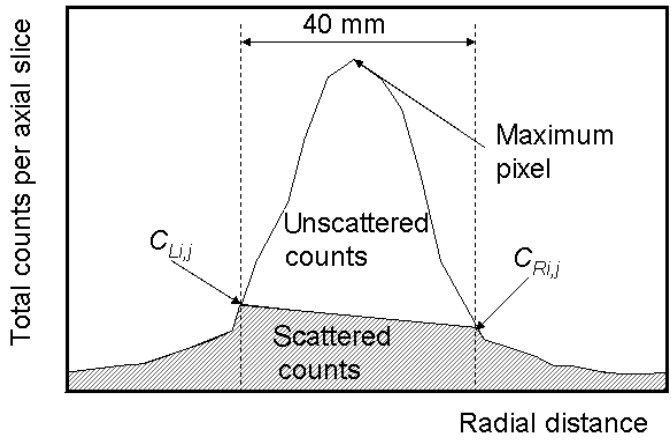

FIG. 19. Integration of scattered counts in the projection sum profile.

(a) Method A: Analysis with estimate of randoms

In each randoms sinogram i of acquisition j, set all pixels located further than 12 cm from the centre of the phantom to zero. Find the number of random counts in sinogram i of acquisition j, $C_{r,i,j}$, by summing the remaining counts.

Calculate the scatter fraction $\mathrm{SF}_{i,j}$ for each slice i and acquisition j as

$$\mathrm{SF}_{i,j} = \frac{\sum_j C_{r+s,i,j} - \sum_j C_{r,i,j}}{\sum_j C_{\mathrm{Tot},i,j} - \sum_j C_{r,i,j}}. \tag{13}$$

Then, compute the system scatter fraction as

$$\mathrm{SF}_j = \frac{\sum_i \sum_j C_{r+s,i,j} - \sum_i \sum_j C_{r,i,j}}{\sum_i \sum_j C_{\mathrm{Tot},i,j} - \sum_i \sum_j C_{r,i,j}}. \tag{14}$$

The NEC rate is computed as follows. First, the total event rate $R_{\mathrm{Tot},i,j}$ for each slice i is calculated as:

$$R_{\mathrm{Tot},i,j} = C_{\mathrm{Tot},i,j}/T_{\mathrm{acq},j} \tag{15}$$

Then, for each slice i, the true event rate $R_{t,i,j}$, the random event rate $R_{r,i,j}$, and the scatter event rate $R_{s,i,j}$ are calculated, respectively, as:

$$R_{t,i,j} = (C_{\mathrm{Tot},i,j} - C_{r+s,i,j})/T_{\mathrm{acq},j} \tag{16}$$

$$R_{r,i,j} = C_{r,t,j}/T_{\mathrm{acq},j} \tag{17}$$

and

$$R_{s,i,j} = (C_{r+s,i,j} - C_{r,i,j})/T_{\mathrm{acq},j} \tag{18}$$

where $T_{\mathrm{acq},j}$ is the duration of frame j.

The NEC rate for each slice i of acquisition j is computed as:

$$R_{\mathrm{NEC},i,j} = R^2_{t,i,j}/(R_{\mathrm{Tot},i,j} + kR_{r,i,j}) \tag{19}$$

where the factor k is set to 0 for tomography equipment that does not perform direct randoms subtraction, and 1 for scanners that use direct randoms subtraction, to account for the fact that the estimation of the randoms is noisy.

The system NEC rate, $R_{\mathrm{NEC},j}$, is calculated as the sum of $R_{\mathrm{NEC},i,j}$ over all slices i.

(b) Method B: Alternative analysis with no estimate of randoms

For scanners with negligible intrinsic natural radioactivity, the scatter fraction can be estimated using the last acquisitions j', for which count losses and random rates are below 1% of the true event rates. Considering the number of randoms to be negligible, $C_{r+s,\,i,\,j'}$ is assumed to be due only to scatter counts.

For each slice i the scatter fraction SF_i is then calculated for the j' low radioactivity acquisitions as:

$$\mathrm{SF}_i = \frac{\sum_{j'} C_{r+s,i,j'}}{\sum_{j'} C_{\mathrm{Tot},i,j'}} \tag{20}$$

and the system scatter fraction is obtained as the count-weighted average of the SF_i values, i.e.

$$\mathrm{SF} = \frac{\sum_{i}\sum_{j'} C_{r+s,i,j'}}{\sum_{i}\sum_{j'} C_{\mathrm{Tot},i,j'}} \tag{21}$$

The total event count rate, $R_{\mathrm{Tot},i,j}$, for each acquisition j and for each slice i is given by:

$$R_{\mathrm{Tot},i,j} = C_{\mathrm{Tot},i,j}/T_{\mathrm{acq},j} \tag{22}$$

and the system total event rate, $R_{\mathrm{Tot},j}$, is the sum of $R_{\mathrm{Tot},i,\,j}$ over all slices i. The true event count rate, $R_{t,i,j}$, for each acquisition j and for each slice i is calculated as:

$$R_{t,i,j} = (C_{\mathrm{Tot},i,j} - C_{r+s,i,j})/T_{\mathrm{acq},j} \tag{23}$$

and the system true event rate $R_{t,j}$ is the sum of $R_{t,i,j}$ over all slices i.

The random event rate, $R_{r,i,j}$, for each acquisition j and for each slice i is calculated as:

$$R_{r,i,j} = R_{\text{Tot},i,j} - R_{t,i,j}/(1 - \text{SF}_i) \tag{24}$$

The scatter event count rate, $R_{s,i,j}$, for each acquisition j and for each slice i is:

$$R_{s,i,j} = R_{t,i,j}\,\text{SF}_i/(1 - \text{SF}_i) \tag{25}$$

and, again, the system scatter event rate $R_{s,j}$ is the sum of $R_{t,i,j}$ over all slices i.

The NEC rate, $R_{\text{NEC},i,j}$, for each slice i of acquisition j is computed as:

$$R_{\text{NEC},i,j} = R_{t,i,j}/(R_{\text{Tot},i,j} + kR_{r,i,j}) \tag{26}$$

where the factor k is set to 0 for tomography equipment that does not perform direct randoms subtraction, and 1 for scanners that use direct randoms subtraction, to account for the fact that the estimation of the randoms is noisy.

The system NEC rate, $R_{\text{NEC},j}$, is calculated as the sum of $R_{\text{NEC},i,j}$ over all slices i.

5.1.3.6. Suggested tolerances

Calculated values of scatter fraction, peak NEC and radioactivity concentration for peak NEC should meet or exceed the vendor's specifications.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criterion for SF is:

$$\text{SF}_{\text{observed}} < 1.05\ \text{SF}_{\text{expected}}$$

The NEC curve, NEC peak value and peak radioactivity concentration shall be reported and registered for future comparison.

5.1.3.7. Corrective action

If the tolerance criteria are not met for this test, it should be verified that the source is correctly assembled and that all procedures were correctly

followed. If the cause cannot be identified, the manufacturer should be notified and corrective action performed before further testing is performed.

5.1.4. Energy resolution

5.1.4.1. Aim

This test is relevant only for tomography equipment using singles-based attenuation correction and calibration. Measurement of energy resolution allows an assessment of proper photomultiplier calibration and ensures that the efficiency of light collection is within the specifications.

5.1.4.2. Frequency

This test, if applicable, must be done by the qualified medical physicist at the time of acceptance testing, as part of end-of-warranty tests and whenever it is suspected that the detector system performance may have changed significantly.

5.1.4.3. Materials

A point source of ^{18}F is employed, less than 1 mm in extent in both the transaxial direction and the axial direction. An adequate source can be prepared using a capillary tube, with an inside diameter of less than 1 mm and an outside diameter of less than 2 mm. The axial extent of the radioactivity in the capillary tube must be less than 1 mm.

A source of the same type used for spatial resolution measurements can be used for this test. The source should be placed at the centre of the FOV, suspended in air, to minimize the effect of scattered radiation.

The radioactivity of the source shall be such that the per cent dead time or randoms loss is less than 5%, as for the spatial resolution test. Typically, this can be obtained with a radioactivity of approximately 1 MBq. The radioactivity concentration of the starting radioactive solution should thus be about 1000 MBq/mL (or 27 mCi/mL) or less. As for other tests, manufacturer recommended activities, if available, may depart from these general recommendations and should be used instead.

5.1.4.4. Data acquisition

The manufacturer's procedure should be followed for energy testing or for energy spectra collection and display (an example of such a spectrum is

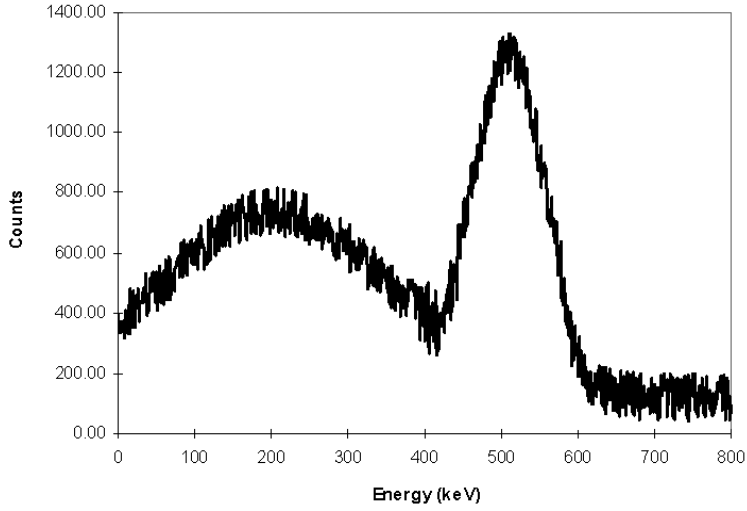

FIG. 20. Example of an acquired energy spectrum; energy resolution can be calculated from the FWHM of the energy peak distribution.

shown in Fig. 20). A time sufficient to obtain not less than 10 000 counts in the peak of the energy distribution should be taken.

5.1.4.5. Analysis

Using the manufacturer's procedure for energy testing, obtain the per cent energy resolution of the system.

Alternatively, if a predefined procedure is not available, the energy spectra should be analysed to obtain the FWHM of the energy peak distribution.

An approximate energy calibration factor can be obtained by calculating the peak position using a parabolic fit to the top of the peak. Using this factor, the FWHM can be converted into units of energy (keV). Energy resolution can then be calculated using the following relation:

$$R_E = 1000 \text{ FWHM}/500 \quad (27)$$

5.1.4.6. Suggested tolerances

The values of energy resolution measured should not exceed those given in the vendor's specification.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criterion for FWHM is:

$$R_{E^{\text{measured}}} < 1.05 R_{E^{\text{expected}}}$$

5.1.4.7. Corrective action

If tolerance criteria are not met for this test, it should be verified that all procedures were correctly followed. If the cause cannot be identified, the manufacturer should be notified and corrective action taken.

5.1.5. Image quality and accuracy of attenuation, and scatter correction and quantitation

5.1.5.1. Aim of the test

Tomographic image quality is determined by a number of different performance parameters, primarily the scanner sensitivity, tomographic uniformity, contrast and spatial resolution, and the process that is used to reconstruct the images. Because of the complexity of the variation in the uptake of radiopharmaceuticals and the large range of patient sizes and shapes, the characteristics of radioactivity distributions can vary greatly and a single study with a phantom cannot simulate all clinical imaging conditions. However, such a study can give some indications of image quality for a particular imaging situation that could be reproduced on different scanners at different times. The purpose of this measurement, which follows closely the NEMA NU2-2007 recommendations, is to produce images simulating those obtained in a total body imaging study involving both hot and cold lesions. Radioactivity is present outside the PET scanner to mimic out-of-field radioactivity, and spheres of different diameters are imaged in a simulated body phantom with non-uniform attenuation. Image quality is assessed by calculating image contrast and background variability ratios for both hot and cold spheres. The same experiment also estimates the accuracy of the attenuation and scatter corrections. Finally, this test allows assessment of the accuracy of the absolute quantification of radioactivity concentration in the uniform volume of interest inside the phantom. It is recommended that three replicates of this test be performed to improve the reliability of results.

5.1.5.2. Frequency

This test must be done by the qualified medical physicist at the time of acceptance testing, as part of end-of-warranty tests, annually and whenever it is suspected that the performance of the detector system may have changed significantly.

5.1.5.3. Materials

The first phantom needed for this test is the 'image quality phantom' described in IEC Standard 61675-1 [40] that is also used to assess the accuracy of PET/CT registration. The phantom consists of:

(a) A 'body compartment' that is at least 18 cm in interior length in order to cover the whole axial FOV of the PET scanner;
(b) Six hollow spheres with internal diameters of 1.0, 1.3, 1.7, 2.2, 2.8 and 3.7 cm, and a wall thickness of not more that 1 mm (Fig. 21);
(c) A cylindrical insert (5.0 ± 0.2 cm outside diameter) filled with a low atomic number material that mimics lung attenuation (average density of 0.3 ± 0.1 g/mL) is centred inside the 'body compartment', and extends axially through the entire phantom. The required density can be achieved by first filling the insert with styrofoam beads, then filling the airspaces between the beads with water.

FIG. 21. IEC/NEMA 2001/2007 body phantom.

In addition to the image quality phantom, this test requires the use of a second phantom to mimic out-of-field radioactivity. The latter phantom shall ideally be the 70 cm long scatter phantom, with the off-centre line source used above for determining the scatter fraction, count losses and randoms measurements. In the absence of such a phantom, a uniform cylindrical phantom containing the same radioactivity can be used instead of the 70 cm long phantom.

5.1.5.4. Phantom preparation

The body compartment is to be filled with an ^{18}F solution of 5.3 ± 0.27 kBq/mL radioactivity concentration. The 5.3 kBq/mL radioactivity concentration corresponds to a total injected radioactivity of

5.3 kBq/mL × 70 000 mL = 371 MBq (~10 mCi)

which is a typical injected dose for whole body PET studies. If a lower dose is recommended by the manufacturer for whole body scans then a lower radioactivity concentration than 5.3 kBq/mL could be used and reported for this test. This test shall be performed for two sphere-to-background ratios of 4:1 and 8:1.

Hint: A practical approach to accurately achieve a 4:1 sphere-to-background ratio without spills is to fill the body compartment with a quarter of its total volume, add the radioactivity intended for the background compartment, use this solution to fill the spheres, and then fill the body compartment with water and cover the phantom with the lid to which the spheres are attached. A similar method can be used to obtain an 8:1 ratio.

The 2.8 and 3.7 cm spheres shall be filled with cold water to mimic cold lesion imaging. The 1.0, 1.3, 1.7 and 2.2 cm spheres are to be filled with an ^{18}F solution that has either 4 or 8 times higher radioactivity concentration than the background. The spheres shall be positioned in such a manner that the centres of all spheres shall be in the same transverse slice, at a 5.72 cm radius from the centre of the phantom, with the 1.7 cm sphere positioned along the horizontal axis of the phantom (Fig. 22). To simplify phantom preparation, it is recommended that the 4:1 and 8:1 measurements be performed on separate days.

The line source of the second phantom shall be filled with 116 MBq of ^{18}F to mimic an out-of-field radioactivity with an effective concentration equal to the background radioactivity concentration of 5.3 kBq/mL used in the body

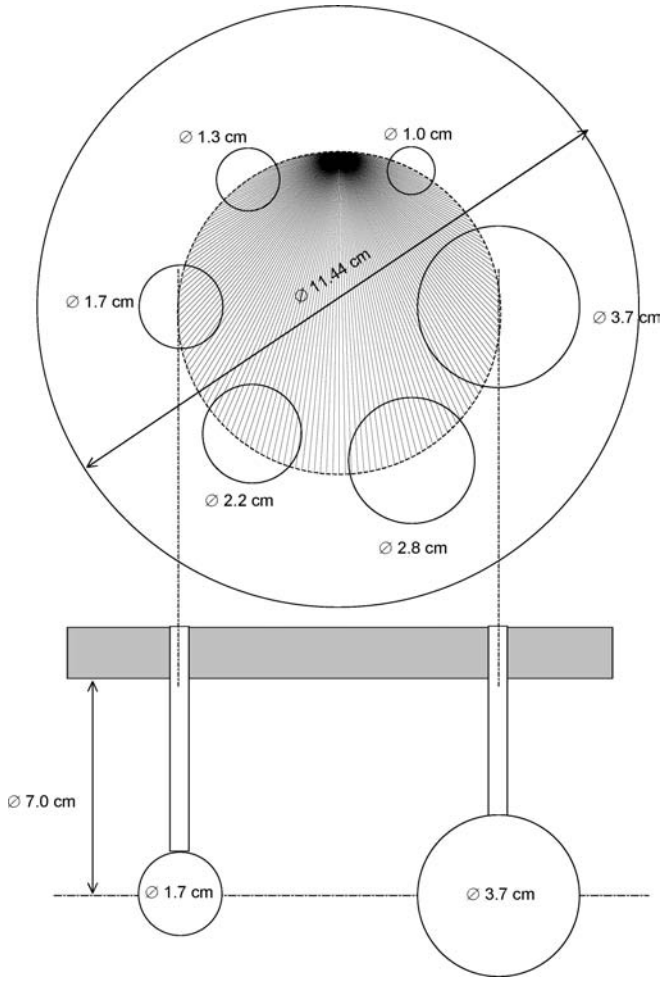

FIG. 22. Insert for hollow spheres in the image quality phantom [40]. The phantom material is polymethylmethacrylate. Note that all the spheres are in the same transverse slice. The diagram is not to scale.

compartment of the image quality phantom. Next, the line source shall be threaded through the 6.4 mm hole in the 70 cm phantom. If a uniform cylindrical phantom is used instead of the 70 cm phantom, then it should be filled with a uniform radioactivity concentration of 5.3 kBq/mL (≈33 MBq in a typical 20 cm diameter, 20 cm high cylindrical phantom).

The body phantom shall be positioned at the end of the table in a head first, supine position and shall be positioned axially in the scanner so that the

centre of the spheres is at the middle slice of the scanner and positioned transaxially so that the centre of the phantom is centred in the scanner. The phantom should also be aligned so that the plane through the centres of the spheres is coplanar to the middle slice of the scanner to within 3 mm throughout the length of the phantom. The 70 cm phantom is to be placed on the bed at the head end of the body phantom (the end nearest to the spheres) and abutting it, in order to best approximate out-of-field radioactivity, as is the case in a clinical situation.

5.1.5.5. Data acquisition

The duration of the acquisition shall be determined in such a way as to mimic a whole body scan that covers 100 cm in 60 min, where the bed is translated between positions for a distance less than the axial FOV. The scan duration will cover the time required for both emission and transmission studies, assuming that both are performed at each bed position. The total scan time for each bed position will therefore be equal to

60 min × (axial step/100 cm)

when the bed is moved by an axial step between two positions in a whole body scan. Additional scan durations that mimic different axial imaging lengths and scanning times may be used. The emission and transmission scan durations shall be reported along with the total axial imaging distance being simulated. On scanners capable of both 2-D and 3-D operation, the test should be performed in both modes.

5.1.5.6. Analysis

Whole body scans performed with lesion-to-background ratios of 4:1 and 8:1 shall be reconstructed in the manner recommended by the manufacturer for the standard whole body imaging protocol (e.g. pixel dimensions, slice thickness, acquisition and image matrix size, reconstruction algorithm with the appropriate filtering and smoothing). All acquisition and reconstruction parameters shall be reported.

(a) Image quality

One transverse slice shall be used in the image quality analysis. A transverse reconstructed image centred on the cold and hot spheres shall be used in the analysis. The appropriate slice can be determined by viewing the

individual slices and selecting the transverse image in which the hot and cold spheres are visualized with the highest contrast. The same slice shall be used for all spheres. Circular ROIs shall be drawn on each hot and cold sphere. The diameter of the ROI shall have a diameter that is as close as possible to the inner diameter of the sphere that is measured. The ROI analysis tool should take into account partial pixels and permit movement of the ROI in increments of 1 mm or less. Regions of interest of the same sizes as the ROIs drawn on the hot and cold spheres shall be drawn in the background of the phantom on the slice centred on the spheres. Twelve 37 mm diameter ROIs shall be drawn throughout the background at a distance of 15 mm from the edge of the phantom but no closer than 15 mm to any sphere (Fig. 23).

ROIs of the same sizes as the smaller spheres (10, 13, 17, 22 and 28 mm) should be drawn concentric to each of the 37 mm ROIs on the background region. The same set of background ROIs shall also be drawn on the slices as close as possible to +2 cm, +1 cm, –1 cm and –2 cm on either side of the central slice. A total of 60 background ROIs of each size, 12 ROIs on each of five slices, shall be drawn. The locations of the ROIs must be the same in each of the replicate scans. The average counts in each background ROI shall be recorded. The per cent contrast $Q_{H,j}$ for each hot sphere j is calculated as:

$$Q_{H,j} = 100[(C_{H,j} - C_{B,j})/C_{B,j}]/[(a_H - a_B)/a_B] \quad (28)$$

where $C_{H,j}$ is the average number of counts in the ROI for sphere j, $C_{B,j}$ is the average of the background ROI counts for sphere j, a_H is the radioactivity

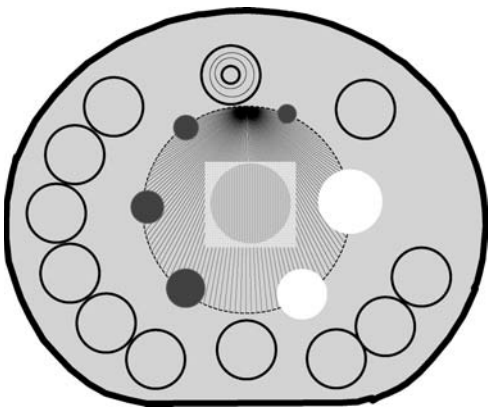

FIG. 23. Image quality analysis: Placement of background regions of interest. This diagram is not to scale.

concentration in the hot spheres and a_B is the radioactivity concentration in the background.

The per cent contrast $Q_{C,j}$ for each cold sphere j is computed as:

$$Q_{C,j} = 100(C_{B,j} - C_{C,j})/C_{B,j} \tag{29}$$

where $C_{C,j}$ is the average number of counts in the ROI for sphere j and $C_{B,j}$ is the average number of background ROI counts for sphere j.

The per cent background variability N_j for sphere j is calculated as:

$$N_j = 100\frac{\sqrt{\frac{1}{K-1}\sum_{k=1}^{K}\left(C_{B,j,k} - C_{B,j}\right)^2}}{C_{B,j}} \tag{30}$$

where K is equal to the number of background ROIs. If replicate scans are performed, the mean and standard deviation of the per cent contrast and of the per cent background variability over the replicates shall be reported.

(b) Accuracy of attenuation and scatter corrections

This part of the analysis allows assessment of the accuracy of the scattering and attenuation corrections by measuring the residual error in scattering and attenuation corrections in uniform regions. In order to do so, a circular ROI of 3.0 ± 0.2 cm in diameter shall be drawn, centred as precisely as possible, on the lung insert. The average pixel value within the ROI, $C_{lung,i}$, should be recorded for each slice i. With perfect corrections for scattering and attenuation, this value would be close to 0.

Next, 12 circular ROIs that are 3.0 ± 0.2 cm in diameter shall be placed on each slice i at the background locations specified in the previous section (Fig. 23), and the average pixel values within each ROI, $C_{B,i}$, should be recorded.

The accuracy of the scattering and attenuation corrections is assessed by measuring the average pixel value with the lung insert ROI as a percentage of the background, and expressing it as the per cent relative error $\Delta C_{lung,i}$ for each slice i as follows:

$$\Delta C_{lung,i} = 100 C_{lung,i}/C_{B,i} \tag{31}$$

where $C_{\text{lung},i}$ is the average counts in the ROI placed over the lung insert and $C_{\text{B},i}$ is the average of the twelve 3.7 cm background ROIs drawn for the image quality analysis.

(c) Accuracy of radioactivity quantitation

This part of the analysis allows assessment of the accuracy of quantitation of radioactivity concentration by the scanner after all corrections have been performed. The radioactivity concentration in the background compartment of the image quality phantom was specified at the beginning of this section to be 5.3 ± 0.27 kBq/mL. Therefore, the true radioactivity concentration is assumed to be known within 5% and should be denoted by A_{B}. Using the option provided by the manufacturer to display radioactivity concentration in MBq/mL, the average radioactivity $C_{\text{B},i}$, of the twelve 3.7 cm background ROIs drawn for the image quality analysis in slice i shall be recorded in MBq/mL as $A_{\text{B},i}$ and the quantitation error ΔA_i in slice i shall be calculated as:

$$\Delta A_i = 100(A_{\text{B},i} - A_{\text{B}})/A_{\text{B}} \tag{32}$$

5.1.5.7. Suggested tolerances

For both cases (lesion-to-background ratios of 4:1 and 8:1), the following data should be reported:

(a) The exact original background concentration in the phantom, the time when it was prepared and the injected radioactivity recommended by the manufacturer for a whole body scan.
(b) All acquisition parameters used in this study (which are expected to be the standard parameters recommended by the manufacturer for a whole body scan), such as the emission and transmission imaging times, axial step size and the total axial distance simulated in this imaging study.
(c) All reconstruction and correction parameters used in this study (which are expected to be the standard parameters recommended by the manufacturer for a whole body scan), such as the reconstruction parameters (the number of subsets and iterations if using an iterative reconstruction algorithm and the reconstruction filters) and other smoothing applied in the axial and transaxial directions, as well as the corrections applied for scattering, randoms, attenuation, decay, dead time, normalization, the image matrix size, and the corresponding pixel size and slice thickness.

(d) The per cent contrast and per cent background variability for each sphere size. If replicate scans were acquired, the average and standard deviation of the per cent contrast and per cent background variability over the replicates shall also be reported.
(e) The individual values of $\Delta C_{\text{lung},i}$ for each slice, as well as the average of these errors over all slices.
(f) A transverse reconstructed slice of the image quality phantom through the centre of all spheres, as well as a coronal image through the centre of the 1.7 cm sphere.
(g) The individual values of the quantitation error ΔA_i in each slice i, as well as the average error over all slices.

Since there are no manufacturer specifications, the user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). A 5% tolerance criterion with respect to the baseline established values for all image quality parameters, based on the three replicate measurements, is recommended.

5.1.5.8. Corrective action

If image artefacts are present in the reconstructed images, if lesion detectability is poor or if the tolerance level is exceeded, the daily QC should be rechecked, and recalibration of the system should be considered. If the problem persists, the manufacturer should be notified and corrective action taken.

5.1.6. Coincidence timing resolution for TOF positron emission tomography

5.1.6.1. Aim of the timing resolution test

This test applies only to PET scanners operating in the TOF mode. Characterization of timing resolution is an important test that determines the capability of the system to estimate the difference in time of arrival of the two coincidence photons, and hence obtain information about the likely location of the annihilation along the LOR.

5.1.6.2. Frequency

Measurement of frequency must be performed during acceptance testing and daily on a TOF scanner to ensure constancy of the timing resolution of the scanner, a key characteristic required for TOF scanners.

5.1.6.3. Materials

A point source of a long lived isotope such as ^{22}Na ($t_{1/2}$ = 2.6 a) or another radionuclide recommended by the manufacturer within a scattering material (e.g. steel or brass) is positioned exactly at the centre of the scanner.

5.1.6.4. Data acquisition

Follow the manufacturer's procedure for the estimation of the timing resolution. This would be done by acquiring coincidences with time of arrival, histogramming of differences in time of arrival, and estimation of a FWHM as a measure of the timing resolution.

5.1.6.5. Analysis

The manufacturer's procedure for timing resolution measurement should be used to obtain the timing FWHM.

5.1.6.6. Suggested tolerances

Measured values of timing resolution, R_T, should not exceed the specification given by the vendor.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criterion for timing FWHM is:

$$R_T^{\text{measured}} < 1.05 R_T^{\text{expected}}$$

5.1.6.7. Corrective action

The timing resolution is expected to be a highly constant parameter. If the tolerance criteria are exceeded, the results should be checked and the testing procedure repeated to confirm the finding. If the result is still outside the tolerance criteria, a recalibration of the system should be performed by appropriate service personnel.

5.2. COMPUTED TOMOGRAPHY ACCEPTANCE TESTING

5.2.1. Scattered radiation measurements and shielding verification

5.2.1.1. Aim

The aim of this test is to ensure that the radiation shielding specified in the original design complies with the local regulatory CT shielding requirements.

5.2.1.2. Frequency

Tests should be done at the time of acceptance, annually and at any time when the technical factors (e.g., kVp and mAs) or the workload of the scanner may have changed.

5.2.1.3. Materials

The materials required are professional reports [41–43], isodose maps provided by the CT manufacturer, the design of the shielding, a 32 cm diameter CT dose phantom and a sensitive radiation detector (e.g., a survey meter and a large volume ionization chamber).

5.2.1.4. Procedure

The amount of shielding material in the walls, floor and ceiling should be determined from the original shielding design. The ideal approach is for the qualified medical physicist to monitor construction and to visually ensure that the appropriate lead shielding is applied to the specified walls, and that all seams and penetrations are appropriately shielded. Otherwise, it is necessary at acceptance testing to verify, by making physical measurements if possible, the presence of the appropriate thickness of shielding material [42, 43].

The CT dose phantom should be placed on the table and scanned with a high kVp–mAs technique. The integrity of the shielding over the surface of the walls, floor and ceiling should be determined while the scanner is producing radiation, using a sensitive radiation detector, with particular attention being paid to potential gaps in the shielding around shielding joints, electrical switches and junction boxes.

On the basis of the measured attenuation of the shielding material and other required parameters [42, 43], including the occupancy factors of adjacent areas, the effective dose to individuals should be determined under various

workloads. If any of these effective doses approach the maximum permissible doses (based on local regulatory limits) to radiation workers or members of the public, as appropriate, then it may be necessary to add additional shielding, limit occupancy in adjacent areas or limit the number of patients scanned.

The same procedure applies at acceptance testing, annually or when technical factors or workloads may have been changed. However, evaluation of the integrity of the shielding, i.e. looking for gaps in the shielding, is only necessary during acceptance testing, unless subsequent construction or physical changes have occurred in the radiation barriers.

The detailed procedures discussed in Refs [41–43] should be followed.

5.2.1.5. Analysis

A survey meter may be used to quickly scan for potential gaps in the radiation shielding. If any gaps are located then a large volume ionization chamber must be used to determine the actual amount of radiation leakage through this area.

The attenuation of the walls, floor and ceiling should be determined. These data are then used to determine the effective dose to radiation workers and members of the public in adjacent areas.

5.2.1.6. Suggested tolerances

The effective doses determined on the basis of this procedure must not exceed the maximum permissible doses to radiation workers or members of the public, as appropriate. Ideally, the effective doses should be limited to some fraction of the maximum permissible dose, for example, one half, based on local regulatory requirements.

5.2.1.7. Corrective action

Any time that the effective doses, based upon measured attenuation of the shielding and calculations taking into consideration the workload of the CT scanner, approach the maximum permissible doses for radiation workers or members of the public, consideration must be given to how to improve the radiation shielding of the CT scan room or of how to limit the dose to these individuals. Area monitoring using thermoluminescent dosimeters (TLDs) or other dosimeters can also be considered if there is concern amongst staff about the appropriate shielding levels.

5.2.2. Computed tomography laser alignment

5.2.2.1. Aim

This test ensures that the gantry lasers and room alignment lasers (for therapy treatment purposes) are properly aligned with the CT gantry and table. This test is especially important if the images are being used for radiation therapy treatment planning.

5.2.2.2. Frequency

The CT laser alignment test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever a service is performed on the CT system which might have an impact on the laser alignment. If images are to be used for radiation therapy treatment planning, the test should be carried out daily by the technologist, or at least on those days prior to using the system for treatment planning purposes.

5.2.2.3. Materials

A laser alignment QC device is required, similar to that shown in Fig. 24 (Fig. 4 of Ref. [7]).

5.2.2.4. Procedure

The test device is placed on the tabletop (Fig. 24), aligned with the sides of the table, and attached to the tabletop. The device should be centred on the tabletop and aligned orthogonally with the long axis of the table.

The test device is positioned using all the gantry lasers. A CT scan with a 1 mm slice width is acquired.

With the test device still attached to the table and properly aligned with the gantry lasers, the alignment of the wall and ceiling lasers is tested relative to the position of the gantry lasers.

The procedure for alignment of the gantry and room lasers is described in detail in Appendix D, Sections 3–6, of Ref. [7].

5.2.2.5. Analysis

The CT image for gantry laser alignment should demonstrate the holes in the test device. In addition, these holes should appear similar from left to right

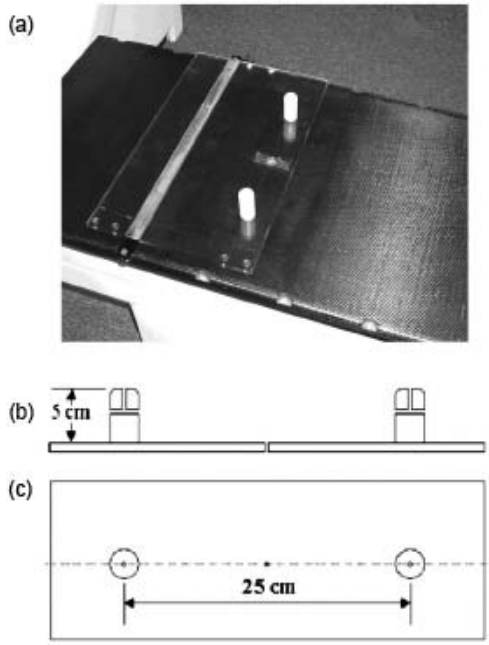

FIG. 24. (a) A CT laser QC device attached to the top of a table; (b) a diagram showing a side view of the device through the centre of pegs, with holes drilled inside the pegs; (c) a diagram showing the top of the device. Courtesy: S. Mutic [7].

and similar to those shown in Fig. 25 (Appendix D, Fig. 8, of Ref. [7]). A visual analysis of the images is described in detail in Appendix D, Sections 1 and 2, of Ref. [7].

5.2.2.6. Suggested tolerances

The results should demonstrate proper alignment within ±1 mm.

5.2.2.7. Corrective action

Necessary corrective action will depend on the specific application. If the PET/CT fused images are to be used for radiation therapy treatment planning, then ±1 mm is the maximum tolerance that should be accepted. Tolerances for other applications will be developed by the responsible medical physicist.

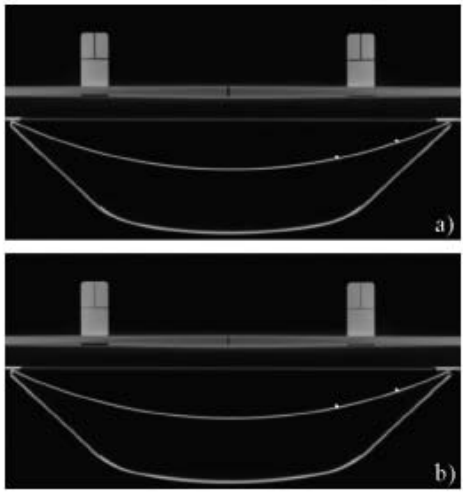

FIG. 25. A CT image of a laser alignment QC test device: (a) lasers aligned with an imaging plane; (b) centre of a test device offset by 1 mm from the imaging plane. Courtesy: S. Mutic [7].

5.2.3. Tabletop alignment and positional accuracy, and scout scan accuracy

5.2.3.1. Aim

This test needs to ensure that:

(a) The tabletop is level and orthogonal with respect to the image plane.
(b) The table and vertical motions according to digital indicators are accurate and reproducible.
(c) The table indexing and positioning under scanner control are accurate.

This test is especially important if the images are being used for radiation therapy treatment planning.

It should also be ensured that the scout scan image accurately indicates the position of the patient.

5.2.3.2. Frequency

This test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever a service is performed on the CT system that might have an impact on tabletop

alignment and positional accuracy and scout scan accuracy, including, but not limited to, system calibration, software changes or upgrades.

5.2.3.3. Materials

A laser alignment QC device similar to that shown in Fig. 24 (Fig. 4 of Mutic et al. [7]), ruler (1 m long), and 70 and 140 kg weights to be distributed on the table to simulate a patient are needed.

5.2.3.4. Procedure

For acceptance testing and annual checks, both 70 and 140 kg weights should be distributed on the table, representing a normal weight distribution of patients. For QC tests, as a minimum, the 70 kg weight should be used.

The laser alignment QC test tool is placed on the extreme head end of the table and aligned using the gantry lasers. A single 1 mm thick CT slice is acquired. The QC test tool is then placed on the extreme foot end of the table and aligned with the gantry lasers. A single 1 mm thick CT slice is acquired.

To determine the accuracy and reproducibility of the table indexing under manual and computer control, a ruler is placed on the centre of the table and accurately aligned with the long axis of the table. The table is then moved both manually and under computer control to determine the accuracy and reproducibility of motion. Details of this procedure are discussed in Appendix E of Ref. [7].

For scout scan imaging, the laser alignment QC test tool is placed with the pegs along the long axis of the table and centred on the centre line of the table. One peg is aligned with gantry lasers at the starting point of the scout scan image, and an image is acquired such that both pegs are included in the image.

This scout scan image should be used to locate a 1 mm slice directly over the hole in each peg. The 1 mm slices should be acquired.

The scout scan images should be analysed using the CT scanner display system.

5.2.3.5. Analysis

For the first part of this procedure, the positions of the horizontal holes in both pegs on the QC device are measured using the scanner cursor tool.

For the accuracy and reproducibility of the manual and computer controlled table positions, the measurements made on the ruler, using the laser marker, are recorded.

5.2.3.6. Suggested tolerances

The horizontal holes in the pegs of the test device should measure within ±1 mm for the first part of this procedure.

For the manual and computer controlled table positions to be accurate and reproducible, the measurements should be to within ±1 mm.

The centre of the first peg should be located at the start of the scout scan. The distance to the centre of the second peg should be 25 cm from the centre of the first peg within ±1 mm.

The two CT slices acquired of the pegs based on the scout scan image should be centred over the hole in each peg.

5.2.3.7. Corrective actions

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

5.2.4. Visual inspection and programme review

5.2.4.1. Aim

The aim of such an inspection and review is to ensure the adequacy of the physical and radiation safety environment for patients and staff.

5.2.4.2. Frequency

These inspections and reviews are carried out at the time of acceptance testing and annually thereafter.

5.2.4.3. Materials

This is a visual inspection. It is recommended that the medical physicist develop a checklist of items to evaluate during the visual inspection including, but not limited to, presence of auxiliary shielding, presence and adequacy of lead aprons, sharp corners, covers that do not close properly, potential fire safety hazards, presence of fire extinguishers, cleanliness of facility and equipment, and adequacy of air conditioning in the CT patient examination and operator rooms.

5.2.4.4. Procedure

The CT facility, including the control room and the equipment room, should be visually inspected, going through the checklist.

The programme review includes, but should not be limited to:

(a) Ensuring that the appropriate scan protocols are available in writing for the CT technologist and that those protocols have been programmed into the scanner;
(b) Reviewing policies and procedures, for example, those for assisting patients during a CT scan and other safety related issues.

The review of scan protocols should include evaluation of the adequacy of technical factors and patient dose based on patient size, the body part being imaged and the purpose of the examination (see also the test described in Section 4.2.8) [9].

5.2.4.5. Analysis

Analysis of patient doses from CT scans should include ensuring that doses are appropriate to the size of the patient (paediatric patients require significantly lower radiographic factors than adult patients), the body part being scanned (chest CTs should require significantly lower doses than abdomen CTs) and the purpose of the scan (follow-up CT scans can often be carried out at lower doses, for example, 50%, of the original diagnostic examinations), and PET/CT fusion imaging (it may be possible to reduce doses for CT studies for this application).

5.2.4.6. Suggested tolerances

All items on the checklist should meet the appropriate standards. The technical factors and patient doses for a CT scanner should be compared internally, to ensure consistency with other CT scanners in the facility. Patient doses should be compared with the DRLs published by appropriate professional organizations.

5.2.4.7. Corrective actions

Corrective actions should be taken as appropriate. Often issues uncovered during the visual inspection and programme review require

education of the responsible individuals, including the necessity to meet the appropriate standards.

5.2.5. Display profile and width

5.2.5.1. Aim

The aim of this test is to ensure that the volume of the patient being measured and displayed is similar to that selected on the CT scanner console.

5.2.5.2. Frequency

This test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced such that there might be an impact on the display profile and width, including, but not limited to, replacement of an X ray tube, system calibration and software changes or upgrades.

5.2.5.3. Materials

A phantom is designed for this purpose. For spiral scanning, this contains a thin (submillimetre) metal plate (Fig. 26) or a submillimetre sized air hole. Note that the thicknesses of these objects must be less than the nominal slice thickness.

5.2.5.4. Procedure

The test device is placed on the CT table and aligned such that the volume of interest is parallel to the acquisition plane. A series of images is acquired in

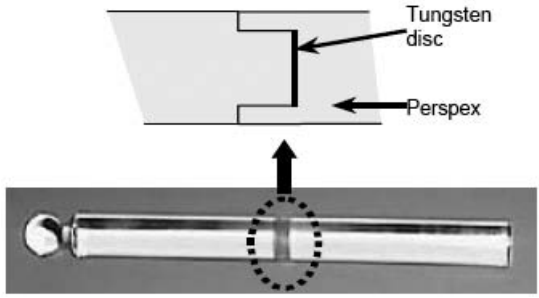

FIG. 26. ImPACT's z sensitivity phantom [44].

spiral mode using the nominal slice thicknesses to be evaluated by a typical clinical technique. The series is then reconstructed at the desired display slice width at intervals of about one tenth of the nominal display slice width [44].

5.2.5.5. Analysis

The CT number of the central region of each image is then recorded. A plot of CT number against the *z* axis allows the FWHM to be calculated.

5.2.5.6. Suggested tolerances

The width of the displayed volume should be within ±1 mm of the slice thickness selected on the CT operator's console.

Note that for extremely thin slices, especially if acquired in axial mode, the image thickness is typically significantly thicker than the nominal thickness; for example, a nominally 1 mm thick slice may actually be 3 mm thick or more. The manufacturer's specifications should be verified regarding slice thickness tolerances.

Differences in slice thickness measurements may be expected between axial and helical scan modes. The scan mode(s) used clinically should be evaluated.

5.2.5.7. Corrective actions

A service engineer will be required to correct issues relative to suggested tolerances.

5.2.6. High contrast modulation

5.2.6.1. Aim

The aim of this test is to ensure that images of high contrast objects have good modulation, i.e. that small details will be imaged with good fidelity.

5.2.6.2. Frequency

This test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced in a way that might have an impact on the high contrast resolution, including, but not limited to, replacement of an X ray tube, system calibration and software changes or upgrades.

5.2.6.3. Materials

A high contrast resolution phantom is required.

5.2.6.4. Procedure

Place the phantom on the CT scanner table at the isocentre and aligned with the *x*, *y* and *z* axes of the scanner. Align the bars of the resolution pattern so that they are at 45° to the *x* and *y* axes of the scanner. Select the FOV that covers the resolution phantom. Acquire one image of the phantom using a typical clinical technique.

5.2.6.5. Analysis

This analysis uses the technique of Droege and Morin [45]. An ROI should be placed inside a large high-contrast resolution pattern so that edge artefacts around the resolution pattern are not included in the ROI. The ROI should include at least five cycles of the resolution pattern in order to make a robust measurement, and it should not include edge or other artefacts. Measure the standard deviation of the pixels in the ROI and record this value as the high contrast modulation.

5.2.6.6. Suggested tolerances

The modulation of the selected resolution pattern should not change by more than ±15%.

5.2.6.7. Corrective actions

Low values of modulation will result if the phantom is not aligned parallel to the *x*, *y* and *z* axes of the scanner.

A service engineer will be required to provide service if reductions occur in the high contrast modulation.

5.2.7. The kVp value and the half-value layer

5.2.7.1. Aim

The aim of this test is to verify that the kVp value has been set correctly by the manufacturer's service engineer and to ensure that there is appropriate

filtration between the X ray source and the patient, thereby helping to minimize the radiation dose to the patient.

5.2.7.2. Frequency

This test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced in a way that might have an impact on the kVp value, half-value layer (HVL) or patient dose including, but not limited to, replacement of an X ray tube, system calibration, generator maintenance and software changes or upgrades.

5.2.7.3. Materials

The materials necessary are a type 1100 aluminium HVL filter set capable of adding 1–10 mm aluminium filtration to the beam, a pencil ionization chamber and an electrometer. Note that a service engineer from the manufacturer must be present, to assist the medical physicist with this evaluation.

5.2.7.4. Procedure

The service engineer should demonstrate to the medical physicist that the peak kilovoltage calibration meets the manufacturer's specifications. Consideration should be given to the impact of input power fluctuations on the kVp values applied to the X ray tube.

To measure the HVL, it will be necessary for the service engineer to stop the rotation of the X ray tube with the tube in the 12 o'clock position. (An alternative is to make the HVL measurement during a scout scan exposure.) The pencil ionization chamber is placed in the X ray beam, off the end of the patient table to minimize scattered radiation. The chamber must be placed in the centre of the X ray beam laterally. Displacing the chamber laterally from the centre of the beam will result in significant differences in the HVL measurement, due to the beam shaping filters used in most CT scanners.

The selected kVp and mAs values (120 kVp and 25 mAs are suggested) are sufficient to produce a dose of approximately 3 mGy. This same technique (kVp and mAs) will be used for all measurements to determine the HVL.

Measurements should be made using the selected technique with no added filtration in the beam, and the resultant dose recorded. A sheet of type 1100 aluminium filtration (e.g. 2 mm thick) should be placed in the beam approximately halfway between the X ray tube and the ionization chamber, an exposure made and the resultant dose recorded. This should be repeated with

increasing thicknesses of aluminium until the measured dose is about 40% or less of the dose measured with no filtration in the beam. Finally, one additional measurement should be made with no additional filters in the beam.

These measurements should be repeated for head and body mode, i.e. with head and body filters in place. Note that a detailed procedure for CT HVL measurement is available from ImPACT [46].

5.2.7.5. Analysis

The two measurements with no additional filtration in the beam are compared, to ensure consistency between the radiation doses produced by the CT scanner and those measured by the dosimeter system.

All of the measured doses are then used to determine the HVL, either by plotting the measured doses as a function of additional filtration thickness on log–linear graph paper or by using the approach described in Ref. [30].

5.2.7.6. Suggested tolerances

The two doses measured without additional beam filtration should be within ±2%. If not, sufficient variation exists in either the X ray tube output or the dosimeter system to preclude the use of the doses measured with the additional filters in the beam.

Half-value layers should be equal to or greater than those specified in the CT manufacturer's specifications, or in the appropriate radiation protection regulations. ImPACT indicates that HVLs are expected to be between 5 and 10 mm thickness of aluminium at 120 kVp, based on the American College of Radiology's extensive experience and measurements of CT scanners (these HVLs are higher than most regulatory limits, which are typically in the range of 3.2–4.3 mm aluminium, often with the same limits for general radiography as for CT). Further information regarding CT HVLs is available from ImPACT [46].

5.2.7.7. Corrective actions

If the two doses measured without additional beam filtration are not within ±1%, the source of the variation must be determined and rectified. If the variation is determined to be from the X ray generator, then the manufacturer's specifications should be reviewed to determine if this amount of variation is acceptable for the system in clinical use.

If the calculated HVLs are less than the minimum specified by the manufacturer or local regulations, additional filtration should be added to the beam so that the HVL meets or exceeds the minimum HVL.

5.2.8. Radiation doses, image noise and image uniformity

5.2.8.1. Aim

The aim of these tests is to ensure that:

(a) Appropriate radiation doses are being used for patient CT scans;
(b) Image noise is typical of what would be expected for the specific radiation doses;
(c) Computed tomography numbers (pixel values) are uniform over the image.

5.2.8.2. Frequency

This test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced in a way that might have an impact on radiation dose, image noise and image uniformity, including, but not limited to, replacement of an X ray tube, system calibration, generator maintenance and software changes or upgrades.

5.2.8.3. Materials

The materials required are:

— Head (16 cm diameter) and body (32 cm diameter) acrylic dosimetry phantoms;
— A pencil ion chamber and electrometer;
— DRLs for CT examinations from appropriate professional organizations.

Note that the noise measurements given by manufacturers are for specific phantoms with specific radiographic factors. It is *essential* that these phantoms and factors be used if the noise levels given by the manufacturer are to be used for evaluation purposes.

5.2.8.4. Procedure

Place the 16 cm diameter dosimetry phantom on the table centred at the isocentre of the scanner, with the long axis of the phantom aligned with the z axis of the scanner. Acquire a scout scan and a single 1 mm slice image of the phantom for alignment purposes.

Note that it is assumed that all the doses measured will be the CTDI and the DLP. See Refs [9, 47] for information on CT dosimetry.

For DRL and comparison purposes, $CTDI_w$ should be used, which is defined as follows:

$$CTDI_w = \frac{1}{3}CTDI_c + \frac{2}{3}CTDI_p \quad (33)$$

where $CTDI_w$ is the weighted CTDI, $CTDI_c$ is the CTDI measured in the centre of the phantom and $CTDI_p$ is the peripheral CTDI defined as the average of the four peripheral CTDI measurements.

Place the ion chamber in the centre of the phantom. Use a scout scan image to select the volume or slice to be imaged. Reset the dosimeter readout to zero. Make an exposure in axial mode employing a technique used for clinical head CT scans, and record the technique factors and measured doses. Place the chamber in each of the four peripheral holes of the phantom, take additional exposures and record the doses. The peripheral dose is defined as the average of these four peripheral CTDI measurements.

With the ion chamber in the centre of the 16 cm phantom, select one technique, for example, 120 kVp and 100 mAs, and take dose measurements with various slice thicknesses over the clinical range. Ensure that one set of measurements includes the factors that the manufacturer uses for the reference noise measurements for acceptance testing. Include a complete $CTDI_w$ measurement at one clinical head CT setting.

Place the 32 cm phantom in the scanner and repeat the above procedure. Make $CTDI_w$ dose measurements using typical adult chest, abdomen and pelvis techniques with this phantom.

If the scanner is used for paediatric patients, $CTDI_w$ doses should be measured for paediatric techniques, assuming a 20 kg patient, using the 16 cm phantom and clinical techniques for paediatric head, chest and abdomen examinations.

Save the images produced during the dosimetry scans, as these will be used for noise and uniformity measurements.

5.2.8.5. Analysis

For the analysis of radiation doses, image noise and image uniformity, proceed as follows:

(a) Develop a spreadsheet showing all of the data for easy analysis. Compare the measured $CTDI_w$ doses obtained under the various clinical scan conditions with those displayed on the scanner console. Compare these clinical doses with doses from other CT scanners at the same institution using the same techniques. In addition, compare the doses with national DRLs. Plot the doses for a given technique as a function of slice thickness.
(b) Select areas in both head and body phantoms that appear to be free of artefacts, and with relatively uniform CT numbers in the centre and periphery of the image. These areas should be between 4 and 10 cm^2. Measure the standard deviation and the average of the CT numbers in these areas. The noise is defined as the standard deviation of the CT numbers divided by the average of the CT numbers in each area.[2] Plot the noise as a function of dose on a semi-logarithmic graph.
(c) Determine the difference in the average CT number between the centre and periphery. This difference is the CT number uniformity value.

5.2.8.6. Suggested tolerances

The suggested tolerances for radiation doses, image noise and image uniformity are as follows:

(a) Computed tomography doses for various scanners at one institution should result in the same dose and noise levels for the same technical factors. Computed tomography doses should not exceed DRLs. Doses should be within ±20% of the manufacturer's specification for both $CTDI_w$ and DLP.
(b) The doses for chest CT images should be significantly lower than those for abdomen CT images for the same size patient (phantom) for both adult and paediatric imaging. Doses for paediatric patients, using the 16 cm phantom, should be significantly lower than those for adult patients.
(c) With respect to graphs showing dose as a function of slice thickness, doses should be linear with slice thickness. However, these values will typically show an increase in dose at thinner slice thicknesses due to the increased width of the pre-patient collimators relative to the nominal slice thickness (or overbeaming). Compare the results with the manufacturer's specifications for radiation slice thickness.

[2] Noise can also be defined using the contrast scale and attenuation coefficient of water [7].

(d) The image noise (the standard deviation divided by the average of the CT numbers) should be equal to or less than the values specified by the manufacturer. The average CT numbers should remain within ±5% of the values determined at acceptance testing.
(e) There should be a linear relationship between dose and noise when plotted on a semi-logarithmic graph. Noise levels should be comparable with those of other CT scanners at the facility for the same dose levels.
(f) The uniformity (difference between the average CT number in the centre compared with that at the periphery) should be between ±3 HU (CT numbers) in the head (16 cm) phantom and ±5 HU in the body (32 cm) phantom, where HU stands for Hounsfield unit.

5.2.8.7. Corrective actions

The corrective actions for radiation doses and image noise are as follows:

(a) Computed tomography doses that are above DRLs can be corrected by changing the techniques used, i.e. using the appropriate kVp value and reducing the mAs value. Patient doses for the same examinations with similar CT scanners should be similar.
(b) Increased image noise can result from several factors, including selecting inappropriately low kVp or mAs values, or both, or from malfunctioning of the scanner electronics. The sources of increased noise should be determined and corrected.
(c) Any non-linearities in the noise versus dose figures may be due to increased electronic noise. The source of this noise should be determined and corrected.

Non-uniformity in the CT numbers from the centre to the edge of the phantoms can result from several causes including, but not limited to, inappropriate selection of the beam shaping filter and an incorrect reconstruction algorithm.

5.2.9. Computed tomography number and electron density accuracy

5.2.9.1. Aim

The aim of these tests is to ensure that:

(a) The CT numbers and electron densities for various materials are within appropriate limits.

(b) The electron densities correspond to the values specified in the phantom manufacturer's specifications. (This portion applies only if images are being used for treatment planning purposes. It establishes the relationship between CT number and electron density.)

5.2.9.2. Frequency

The frequency with which these tests are carried out is as follows:

(a) Daily, or before patient scans, for water used in radiation therapy applications and monthly for other materials. These daily and monthly tests are the responsibility of the PET/CT technologist, working under the supervision of the qualified medical physicist.
(b) Annually for electron density.

Note that the annual tests must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced in a way that might have an impact on the CT number and electron density, including, but not limited to, replacement of an X ray tube, system calibration and software changes or upgrades.

5.2.9.3. Materials

The materials required for these tests are:

(a) A phantom with areas of different densities, with, at a minimum, water, polyethylene, acrylic polymers, Teflon and air.
(b) An electron density phantom.

Note that this portion of the evaluation should be carried out in cooperation with the radiation therapy physicist responsible for treatment planning.

5.2.9.4. Procedure

Place the phantom at the isocentre of the CT scanner, aligned with the *x*, *y* and *z* axes of the system, then:

(a) For CT number accuracy, acquire an image (for CT numbers) using the kVp and mAs values used in the acceptance testing specifications. Ensure

that the software reconstruction algorithm is the one used for clinical applications and that it is the same each time these measurements are made.

(b) For electron density purposes, acquire an image using the techniques that will be used for the images involved in the planning of radiation therapy treatment. Transfer the data to the radiation therapy treatment planning system for evaluation.

5.2.9.5. Analysis

An ROI sufficiently small to fit inside the area of interest (4–10 cm^2 for water) should be used to measure the average CT number of each material in the phantom image.

5.2.9.6. Suggested tolerances

Computed tomography number values should be within ±5 HU for values specified by the manufacturer.

5.2.9.7. Corrective actions

The cause of deviations from the suggested tolerances should be determined and corrected. This is usually accomplished by recalibration of the CT scanner by the service engineer.

5.3. PET/CT ACCEPTANCE TESTING

5.3.1. Accuracy of PET/CT image registration

5.3.1.1. Aim

The aim of this test is to assess qualitatively the accuracy of the registration of the images obtained with the PET and CT scanners [48]. Since the fusion of PET and CT images assumes perfect registration of the two modalities, it is crucial to ensure that the two studies are registered in different parts of the FOV (axial and transaxial) for a reasonable range of patient weights. The accuracy of PET/CT image registration becomes even more important when considering the scanner in conjunction with radiotherapy applications. However, due to the complex interplay of different factors affecting the acquisition (e.g. table deflection and patient weight distribution),

this test yields an insight into the accuracy of PET/CT image registration for the particular imaging situation.

This test is especially important if the images are to be used for the planning of radiation therapy treatment. In general, PET/CT systems are supplied with a special PET/CT offset procedure for the initial establishment, and subsequent checks, of the registration of the PET and CT fields of view. The procedure described here may be useful as an alternative means of verifying the accuracy of registration under the influence of the factors mentioned above.

5.3.1.2. Frequency

The PET/CT image registration accuracy test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, and whenever the CT system is serviced in a way that might have an impact on image registration accuracy, including, but not limited to, servicing of the table, after separating the PET and CT gantries for servicing, system calibration and software changes or upgrades.

5.3.1.3. Materials

The phantom needed for this test is the 'image quality phantom' described in IEC Standard 61675-1 [40], which is also used to assess image quality and accuracy of attenuation and scatter corrections during acceptance testing (Fig. 27). The phantom consists of:

(a) A 'body compartment' that is at least 18 cm in interior length in order to cover the whole axial FOV of the PET scanner;
(b) Six hollow spheres with internal diameters of 1.0, 1.3, 1.7, 2.2, 2.8 and 3.7 cm, and a wall thickness of no more that 1 mm;
(c) A cylindrical insert (5.0 ± 0.2 cm outside diameter) filled with a material of low atomic number that mimics lung attenuation (average density: 0.3 ± 0.1 g/mL), is centred inside the body compartment, and extends axially through the entire phantom.

In addition to the image quality phantom, this test requires the use of heavy weights (total weight of about 100 kg) to mimic the scanning of a heavy patient. Lead bricks or other heavy materials can be used for this purpose.

The body compartment shall be filled with an ^{18}F solution of 5.3 kBq/mL radioactivity concentration. The 2.8 and 3.7 cm spheres shall be filled with cold water to mimic cold lesion imaging. The 1.0, 1.3, 1.7 and 2.2 cm spheres shall be

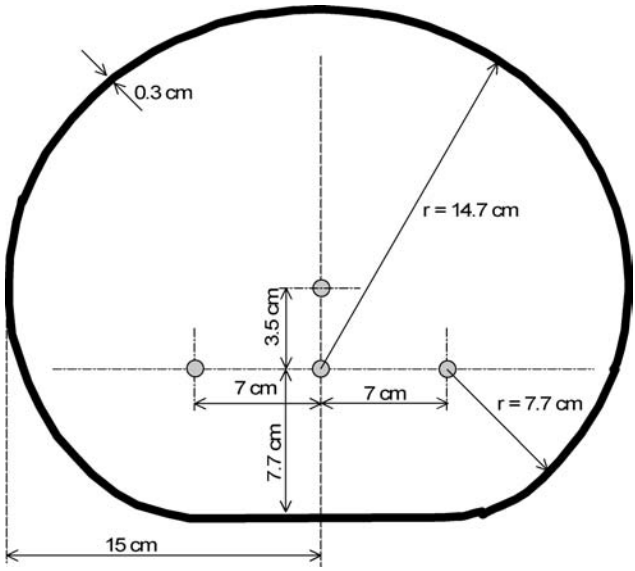

FIG. 27. Image quality phantom (IEC standard 61675-1 [40]). The phantom material is polymethylmethacrylate. This diagram is not to scale.

filled with an ^{18}F solution that is 8 times hotter than the background (sphere:background = 8:1), i.e. with a concentration of 42.4 kBq/mL. If a lower dose is recommended by the manufacturer for whole body scans, then a lower radioactivity concentration than 5.3 kBq/mL could be used and reported for this test. A practical approach to accurately achieve the 8:1 sphere-to-background ratio without spills is to fill the body compartment with an eighth of its total volume, add the radioactivity intended for the background compartment, use this solution to fill the spheres and then fill the body compartment with water and cover the phantom with the lid to which the spheres are attached. The spheres shall be positioned in such a manner that the centres of all spheres shall be in the same transverse slice, at a 5.72 cm radius from the centre of the phantom, with the 1.7 cm sphere positioned along the horizontal axis of the phantom.

The lead bricks (or equivalent heavy weights) shall be uniformly distributed over 1.5 m length of the table adjacent to the quality phantom. The phantom shall be positioned at the end of the table in a supine position. The phantom shall be positioned axially in the scanner so that the centre of the spheres is at the middle slice of the scanner, and positioned transaxially so that the centre of the phantom is centred in the scanner.

The phantom should also be aligned so that the plane through the centres of the spheres is coplanar to the middle slice of the scanner to within 3 mm throughout the length of the phantom. The lead bricks (or equivalent heavy weights) shall be uniformly distributed over a 1.5 m length on the table adjacent to the image quality phantom.

5.3.1.4. Data acquisition

The phantom shall be scanned on the PET and CT scanners using a modified version of the standard whole body protocol, in which the CT acquisition matrix is set to 512 × 512 and the PET acquisition matrix is set to 512 × 512, or, if those values are not available, to the largest values possible. Next, the lead bricks should be removed and a second whole body scan performed with the image quality phantom alone on the table.

5.3.1.5. Analysis

Both whole body scans shall be reconstructed in the manner recommended by the manufacturer for the standard protocol for whole body imaging, except for allowing for larger acquisition matrices (e.g. a PET reconstructed volume of 512 × 512 instead of the standard 128 × 128). The reconstructed CT and PET volumes shall be displayed simultaneously using the image fusion software provided by the manufacturer.

For both cases (in the presence and in the absence of heavy weights), the centres of all spheres shall be visually examined in all three directions on both PET and CT to ensure that they are adequately registered, spatially within 1 voxel. The edge of the phantom shall also be examined to ensure that the edge of the phantom, as seen on the PET scan, appropriately matches the phantom boundaries, as seen on the CT scan.

5.3.1.6. Suggested tolerances

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to place a call for maintenance). An appropriate tolerance criteria for the accuracy of PET/CT registration is a registration within ±1 pixel (or ±1 mm, whichever is smaller) when using a 512 × 512 matrix.

5.3.1.7. Corrective action

The accuracy of PET/CT registration is crucial for accurate attenuation correction and lesion localization. Therefore, this accuracy should not deviate

from the tolerance suggested. If tolerance criteria are exceeded, the problem should be reported to the manufacturer and a service is required.

5.3.2. Visual display and hard copy printing

5.3.2.1. Aim

The aim of this test is to ensure that:

(a) The visual display demonstrates all of the information in the digital image array.
(b) The printing device records an image similar in appearance to the visual display.

5.3.2.2. Frequency

The visual display and hard copy printing test must be performed by a qualified medical physicist at the time of acceptance testing, as part of the end-of-warranty testing, annually and whenever the CT system is serviced in a way that might affect either the visual display or the printing device.

5.3.2.3. Materials

A calibrated photometer with detectors designed to measure luminance (in units of candellas or nit·m^{-2}·sr^{-1}) and illuminance (in units of lux or lm/m^2). The luminance detector should be a small angle acceptance detector or, preferably, a detector with a fibre-optic probe, which allows for luminance measurements directly from the display surface.

Note that measuring luminance and illuminance requires two different detector configurations. Information on measuring these two quantities is available in an ACR manual [30], an AAPM report [49] and in Ref. [50].

5.3.2.4. Test patterns

There are two levels of test pattern available:

(1) Single, all inclusive, patterns such as the test pattern of the Society of Motion Pictures and Television Engineers (SMPTE) (Fig. 28) [52–55] and the AAPM Task Group 18 (Fig. 29) [50] Comprehensive QC Test Pattern;
(2) Specialized test patterns for detailed evaluation of displays [49].

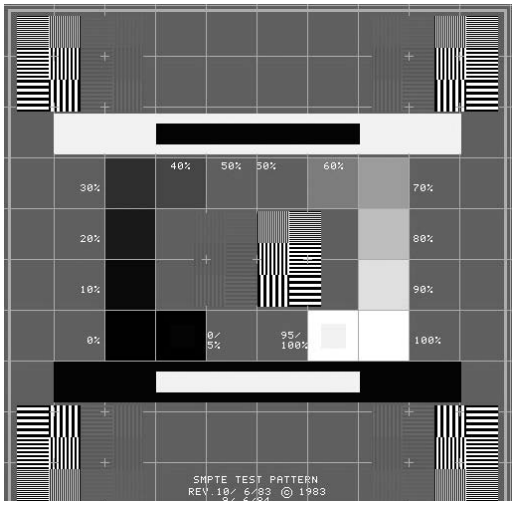

FIG. 28. The SMPTE test pattern for displays and hard copy imaging systems.

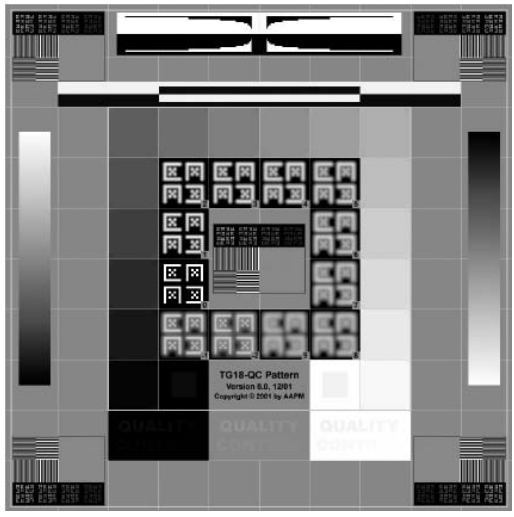

FIG. 29. The Comprehensive QC Test Pattern of AAPM's Task Group 18.

The SMPTE test pattern (Fig. 28) is recognized by most medical imaging companies worldwide and has been in use since the mid-1980s. It is described in Refs [51–54] and is suitable for both acceptance testing and QC. The test pattern is usually available on medical imaging systems or from the manufacturer's engineering department.

A task group of AAPM (Task Group 18) has published a report on evaluating medical imaging displays [49] and has developed several test patterns including the Task Group 18 Comprehensive QC Test Pattern (Fig. 29). Information regarding the use of this test pattern, as well as an extensive set of test patterns for evaluation of displays, is available on-line [49].

5.3.2.5. Procedures

(a) Optimization of display brightness and contrast

Optimization of brightness and contrast are carried out using the two smaller square patches that are inset into the 0 and 100% patches in the grey scale (Fig. 28 [55]). The inset patches are at 5 and 95% of the grey scale, so that they are 5% above pure black and 5% below pure white. If these two patches are visible, then one can be assured that all the information in the digital image is being displayed appropriately. The procedure is as follows:

(1) Display the SMPTE test pattern.
(2) Set the window width to encompass the range of values in the SMPTE test pattern image. This can be determined by displaying a histogram of the pixel values.
(3) Set the window level at either the centre of this range or the lower end of this range, depending on the design of the system. Note that this window width and window level will be referred to in this book as the 'standard window width and level', and should be used for all testing, including QC testing.
(4) Turn both the display brightness and the contrast controls to their minimum settings, i.e. completely counterclockwise.
(5) Adjust only the brightness control until the area outside of the image and the 0% patch are just slightly grey, as opposed to absolutely black.
(6) Adjust only the contrast control so that the 95% patch can be clearly seen.
(7) Adjust only the brightness control so that the 5% patch can be clearly seen.

(8) Confirm that the alphanumeric data in the display are clear and sharp. If they are not, readjust the contrast control until they are sharp. Note that the display is now optimized. The brightness and contrast controls should not be adjusted in the future by the clinical staff. If the clinical images appear to be unsatisfactory, it is now necessary to adjust the window and level settings, but not the brightness and contrast.

(b) Measurement of display luminance and room illuminance levels

Room illuminance is first measured using an illuminance meter. With the display turned off, the meter should be placed on the surface of the display with the detector facing away from the display. Note that the physicist making the measurement is in the FOV of the detector, which may affect the measurement — for example, if a white laboratory coat is worn or if the physicist is standing between the detector and a light source such as a window.

A second measurement of room illuminance should be made (with the display turned off) with the detector at the position of the physician looking at the images and pointed towards the display.

If either of these measurements exceeds the suggested tolerances, the room illuminance conditions must be corrected before proceeding. Note that the room illuminance levels have an impact on the quality and contrast of the image on the display. Consequently, it may be necessary to readjust the brightness and contrast controls of the monitor after measuring the room illuminance levels if these levels are higher than the suggested tolerances.

The display luminance should be measured at three different levels using the SMPTE test pattern and a fibre-optic probe with a photometer. Measurements should be made with the fibre-optic probe placed on the display surface and centred in the 0, 50 and 100% patches of the test pattern.

(c) Optimizing hard copy image quality

Once the display as been optimized as described above, it is possible to set up and optimize the hard copy imaging system. Ensure that the window width and level are set to the standard levels, i.e. the same values as those selected for optimizing the display, and proceed as follows:

(1) Display the SMPTE test pattern at the standard window width and level.
(2) Expose a film image of the SMPTE test pattern.
(3) Using a densitometer, measure the densities of the 0, 10, 40 and 90% patches of the film image.

(4) The densities should fall within the ranges given in Table 5. If they do not, it will be necessary to adjust the hard copy imaging device to obtain densities within the appropriate ranges. It may be necessary to obtain the assistance of a qualified service engineer to make internal adjustments to the imaging device or to adjust the lookup table.

(d) Colour display settings

All of the above procedures focus on the monochromic (black and white) display characteristics. These are the most critical characteristics in ensuring good image quality for PET/CT images. The colour portion of the display can be evaluated and adjusted using the SMPTE colour bar pattern, which is widely available. The image of this pattern on the display is compared with a standard image, usually in hard copy form.

5.3.2.6. Analysis

Measurements should be compared with suggested tolerances.

5.3.2.7. Suggested tolerances

Room illuminance levels must be in the range of 10–50 lx and ideally in the range of 2–10 lx.

Once the room illuminance levels have been optimized, the illuminance measurements made on the SMPTE test pattern should be as shown in Table 6.

The ranges of set-up optical densities for the film images of the SMPTE test pattern are given in Table 5.

TABLE 5. DENSITY RANGES FOR FILM IMAGES OF THE SMPTE TEST PATTERN

SMPTE area (%)	Density range (cd/m^2)
0	3.00–3.25
10	2.25–2.55
40	1.15–1.45
90	0.25–0.35

5.3.2.8. Corrective action

Adjustment of display brightness and contrast can usually be accomplished without the assistance of a service engineer. If these adjustments are not clearly visible on the front or back of the display, the service engineer should be contacted for assistance.

The room illuminance levels are critical to the image quality of displays. If the illuminance levels do not meet the suggested tolerances, the problem should be rapidly corrected. This may require changing the location of room lights, putting light-tight shades on windows and shielding the display from light arising from viewing boxes in other parts of the reading room. It is not unusual to have to make major changes in the lighting in a room to adjust the room illuminance to appropriate levels.

The film densities should fall within the ranges given in Table 6. If they do not, it will be necessary to adjust the hard copy imaging device to obtain densities within the appropriate ranges. It may be necessary to obtain the assistance of a qualified service engineer to make internal adjustments to the imaging device or to adjust the lookup table.

Adjustments for colour balance can be made using the appropriate adjustments, usually located near the brightness and contrast controls. On installation, a service engineer may be helpful in assisting and explaining the approach and terminology used by the manufacturer.

TABLE 6. SUGGESTED TOLERANCE LEVELS FOR DISPLAY LUMINANCE

SMPTE area (%)	Luminance (cd/m^2)
0	0–15
50	55–85
100	≥150

6. ROUTINE QUALITY CONTROL PROCEDURES

6.1. QUALITY CONTROL OF PET

State of the art PET imaging systems require periodic calibrations. The primary purpose of a QC programme is to verify that the images accurately reflect the distribution of radiopharmaceuticals within the patient. It has an important role in monitoring changes in performance so that service can be scheduled and performed before the need becomes critical and requires cancellation of patient studies. A comprehensive QC programme should maximize the quality of diagnostic information available to the physician.

Compared with stand-alone PET scanners, PET/CT systems require the monitoring of additional parameters pertaining to the performance of the CT scanner and the co-registration of the CT and PET data. The manufacturers of PET/CTs generally recommend procedures for routine QC of their equipment, and it is recommended that users follow these recommendations as a minimum. Each manufacturer defines procedures that are specific to its own products. In the absence of standards for routine QC procedures, the general minimum standards for routine PET/CT QC that all owners of PET/CT installations should carry out are recommended here. In some cases, the recommended procedures of a manufacturer may fully meet these standards, but users should implement supplementary procedures from these guidelines when this is not the case.

This section describes routine QC procedures that, when followed, should help to ensure continued optimal operation of PET/CT equipment in terms of image quality and accuracy, as well as the safety of both patients and operators. An effort has been made to ensure that the tests are simple to perform without compromising their ability to detect significant degradation of performance. Procedures are given for both PET and CT. The CT procedures in the following sections apply only to dual modality PET/CT systems.

Routine QC protocols should enable the identification of problems that could affect any of the following performance aspects:

— Image quality of PET;
— Image quality and patient dose of CT;
— Accuracy of CT based attenuation corrections;
— Accuracy of CT and PET co-registration.

Each recommended routine test procedure has a corresponding frequency, suggested tolerance and corrective action associated with it. Factors

that do not change rapidly, such as detector uniformity and PET/CT offset, are specified to be performed relatively infrequently, for example, quarterly and whenever an intervention, such as replacement of a detector module, that could affect them is performed. Other procedures are recommended to be performed daily. As the stability of different PET/CT systems will vary, users should monitor QC parameters and, if necessary, vary the frequency of QC checks as required.

Because of the differences in system architecture between scanners, it is not possible to give specific instructions applicable to every scanner for daily QC. Instead, it is recommended to carefully follow the procedures suggested by the manufacturer, as the latter should effectively assess the condition of the scanner, taking into account its specific characteristics (such as type of detector, geometry and sensitivity).

In PET/CT scanners, daily QC should include an evaluation of the performance of both the PET and CT components. The results of daily QC should be carefully evaluated against a standard operating procedure that defines acceptable limits for each QC parameter, and against the action that should be taken when parameters fall outside these limits. Since acquisition of PET/CT scans on equipment that is not performing to its specifications may compromise image quality, scan interpretation and safety, it is essential that daily QC parameters should be evaluated and that any necessary corrective actions have been completed prior to scanning.

6.1.1. Daily PET detector stability test

6.1.1.1. Aim

The aim of this test is to assess the constancy of the detector performance and to allow early detection of any sudden change, for example, failure of a detector module.

6.1.1.2. Frequency

This test should be performed daily by the technologist, prior to clinical use of the scanner.

6.1.1.3. Materials

Depending on the system, manufacturer and acquisition mode (2-D–3-D), the daily detector stability test will be accomplished using different test sources. Some of the most common sources are:

(a) A rotating ^{68}Ge line source;
(b) A uniform cylindrical ^{68}Ge phantom centred horizontally and vertically in the FOV of the PET;
(c) A ^{22}Na point source mounted on a centred plastic jig and placed approximately at the centre of the FOV (Fig. 30).

The activities of these sources should be as specified by the manufacturer of the PET system.

6.1.1.4. Data acquisition

Using the system's daily PET QC acquisition protocol, the detector stability scan or equivalent daily stability test should be set up and carried out.

6.1.1.5. Analysis

Sinograms (Figs 31 and 32) should be subject to a careful visual inspection for the presence of pronounced diagonal streak artefacts and then compared with previously acquired reference sinograms.

Some manufacturers supply quantitative tools for analysing and reporting daily QC procedures. In this case, the final report produced by the analysis tool

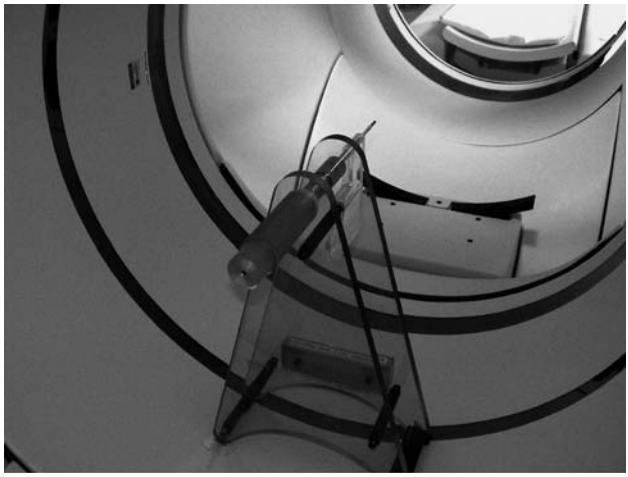

FIG. 30. A Philips Gemini TruFlight PET/CT, with a source holder in the position for daily QC; the same acquisition set-up is used to check the photomultipliers and the energy and time resolutions, as well as to acquire a sinogram for visual inspection.

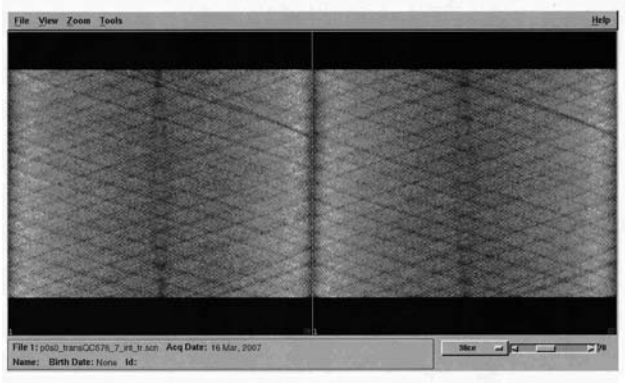

FIG. 31. An example of detector stability acquisition on a Philips Gemini PET/CT using a ^{137}Cs rotating source (courtesy: L. Indovina and A. Giordano, Catholic University, Rome).

FIG. 32. An example of a daily QC sinogram acquired on a Philips Gemini PET/CT using a ^{22}Na source (courtesy: L. Indovina and A. Giordano, Catholic University, Rome).

should be printed and/or digitally stored. The principal quantitative parameters calculated should be stored and QC charts produced in order to detect any trends.

6.1.1.6. Suggested tolerances

The tolerances for this test are usually provided by the manufacturer as a part of the daily QC software protocol. If appearance and/or results are

consistent with a sudden change in detector uniformity or other parameters, a warning message will usually be displayed.

6.1.1.7. Corrective action

Any results outside the allowed tolerance or the presence of artefacts in sinograms or reconstructed images should be carefully investigated prior to clinical scanning. The local field engineer or other staff involved with system maintenance and repair should be informed. Minor drifts in detectors may be corrected on some systems by repeating the detector normalizations until the detectors can be recalibrated. Major changes or pronounced streaks may require either recalibration of the detectors or replacement of faulty detector components.

6.1.2. Daily coincidence timing resolution tests in TOF PETs

Measurement of timing resolution in TOF PETs is described in Section 5.1.6. These measurements must be performed daily by the technologist on a TOF scanner to ensure the constancy of the timing resolution, a key characteristic required for TOF PET scanners.

6.1.3. Test of PET/CT scans in clinical mode

6.1.3.1. Aim

The aim of this test is to check the overall operation of the system in patient scan mode. This is a test of all the components involved in performing a clinical scan and is intended to identify problems with the PET and CT subsystems, including attenuation correction, bed motion, reconstruction and PET/CT registration.

6.1.3.2. Frequency

This is an optional daily test to be performed by the technologist prior to clinical use of the scanner.

6.1.3.3. Materials

The materials used consist of a uniform phantom of radioactivity of approximately 40 MBq, centred horizontally and vertically in the FOV of the PET.

6.1.3.4. Data acquisition

Perform a two bed PET/CT scan for 5 min at each bed position. Define bed positions to overlap at the centre of the phantom.

6.1.3.5. Analysis

Visually inspect reconstructed images for artefacts, and then confirm proper co-registration of PET and CT images.

6.1.3.6. Suggested tolerances

Inspect reconstructed images to check proper acquisition and reconstruction of CT and PET data at both bed positions. Reconstructed PET and CT images should appear uniform. Ensure that PET and CT data appear to be correctly co-registered. If a co-registration problem is suspected, perform a PET/CT co-registration procedure to update the system's co-registration parameters and repeat the test. If co-registration appears correct and image artefacts are seen in the CT images, the CT scanner will need to be serviced. If artefacts appear only in PET images, it may be possible to correct for them by renormalizing the detector. If this does not correct the problem, the PET scanner needs to be serviced.

6.1.3.7. Corrective action

Any faults with system operation and presence of artefacts in sinograms or reconstructed images should be carefully investigated prior to clinical scanning. The local field engineer or other staff involved with system maintenance and repair should be informed.

6.1.4. Uniformity of the reconstructed image

6.1.4.1. Aim of the test

Uniformity of the reconstructed image is a measure of the system response to a homogeneous radioactivity distribution in both the transverse and the axial FOV.

There is no general consensus on the testing methodology or analysis parameters such as quantitative indices of residual non-uniformity. The following procedure is based on the NEMA 1994 standard [56]. If the scanner

already has the protocol for the NEMA 1994 uniformity test, it may be used instead of this procedure.

6.1.4.2. Frequency

This test should be performed quarterly by the qualified medical physicist.

6.1.4.3. Materials

The phantom used for testing is composed of a plastic (polymethylmethacrylate) hollow cylinder, with an external diameter of 203 ± 3 mm, a length of the internal cavity of 190 ± 3 mm and a wall thickness of 3 ± 1 mm (IEC 61675-1 [40] and NEMA standard NU2-1994 [56]).

The cylinder (Fig. 33) is filled with a uniform solution of ^{18}F. The total radioactivity should be in the range 120–130 MBq; this gives a radioactivity concentration of approximately 21 kBq/mL, or about four times the mean whole body radioactivity concentration in a clinical setting (assuming a 370 MBq injection for a 70 kg patient).

The radioactivity, A_{cal}, should be carefully measured using a radionuclide radioactivity calibrator and the time of measurement, T_{cal}, recorded.

The source should be suspended in air to minimize the effect of scattered radiation and avoid any other attenuating material. The centre of the cylinder should be placed at the centre of the axial FOV, but displaced in the vertical direction by 25 mm.

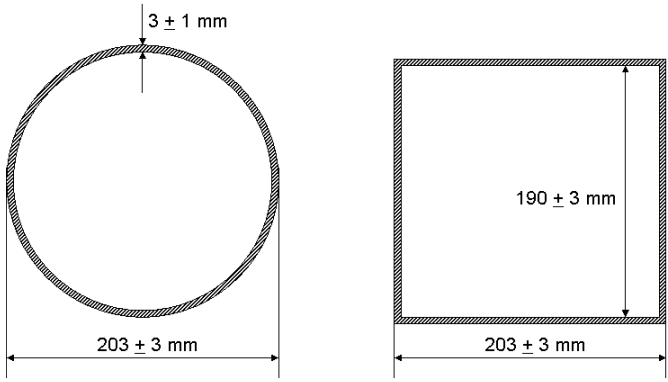

FIG. 33. Plan and elevation of a uniform cylindrical phantom.

Several models of scanner are supplied with an alternative source, a cylindrical uniform source of $^{68}Ge/^{68}Ga$, used for normalization or other routine QC procedures. If such a source is available, it can be used as an alternative to the fillable cylinder.

Commercial ready-to-use $^{68}Ge/^{68}Ga$ sources have dimensions that are slightly different from those of the fillable cylinder mentioned above, but, in this case, their ready availability and the fact that their use is suggested by the PET scanner manufacturers in several QC procedures also support their use for uniformity testing.

The radioactivity of the phantom and the reference time are in this case indicated by the manufacturer.

6.1.4.4. Data acquisition

For scanners capable of both 2-D and 3-D acquisition, measurements should be performed in each mode.

The emission acquisition time of the ^{18}F filled cylinder should be sufficient to collect not less than 20 million counts per transaxial plane.

In the case of use of a $^{68}Ge/^{68}Ga$ source, the cylinder acquisition time should be adapted to the current source radioactivity.

6.1.4.5. Analysis

Slices corresponding to the central 17 cm long active part of the phantom shall be reconstructed with all the corrections applied (for normalization, dead time, decay, randoms, sensitivity, scattering and attenuation). The calculated attenuation correction should be applied, to avoid noise propagation.

Reconstruction is performed according to the standard parameters (e.g. matrix size, pixel size, slice thickness, the reconstruction algorithm and filters) used in clinical scanning.

Reconstructed transaxial and sagittal slices should be displayed and carefully inspected visually for artefacts, and a subjective evaluation of uniformity should be made.

To obtain an approximate quantitative index of non-uniformity (Fig. 34), a circular area with a diameter of 175 mm should be centred inside each transaxial slice of the phantom. An orthogonal grid of square regions of interest, approximately 10 mm × 10 mm, should be drawn on each slice inside the circular area. Square regions that intersect with the circle of 175 mm diameter should be neglected.

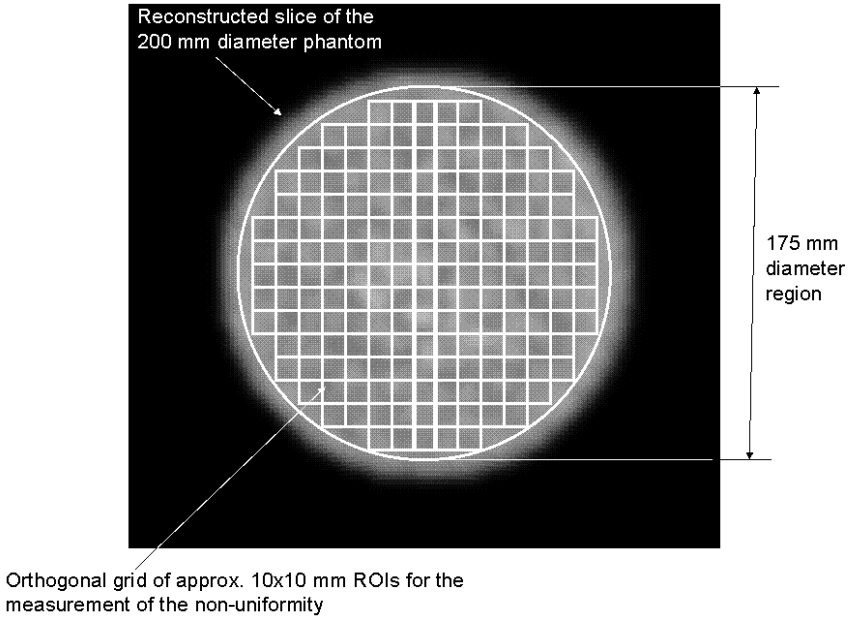

FIG. 34. Scheme of definition and positioning of ROIs for quantitation of non-uniformity.

We define $\mathrm{MAX}(C_k)$, $\mathrm{MIN}(C_k)$ and $\mathrm{AVE}(C_k)$ as the maximum, minimum and average number of counts, respectively, with respect to any square region k within a given slice i. Non-uniformity in each slice shall be evaluated as:

$$\mathrm{NU}_i = \mathrm{MAX}\left\{ \begin{array}{l} 100\dfrac{\mathrm{MAX}(C_k) - \mathrm{AVE}(C_k)}{\mathrm{AVE}(C_k)} \\ 100\dfrac{\mathrm{AVE}(C_k) - \mathrm{MIN}(C_k)}{\mathrm{AVE}(C_k)} \end{array} \right\} \tag{34}$$

and the maximum value of NU_i shall be reported as an index of intra-slice non-uniformity, together with the coefficient of variation:

$$\mathrm{CV}_i = 100\frac{\mathrm{SD}_i}{\mathrm{AVE}(C_k)} \tag{35}$$

where

$$\mathrm{SD}_i = \sqrt{\frac{1}{(N_{\mathrm{ROI,s}} - 1)} \sum_{k=1}^{k} (C_k - \mathrm{AVE}(C_k))^2} \tag{36}$$

in which $N_{\mathrm{ROI,s}}$ is the number of the square ROIs inside the 175 mm diameter circle.

To evaluate volume non-uniformity along the whole FOV, we define MAX(C_j), MIN(C_j) and AVE(C_j) as the maximum, minimum and average number of counts, respectively, with respect to all square regions in all slices.

Volume non-uniformity shall be reported as:

$$\mathrm{NU}_{\mathrm{vol}} = \mathrm{MAX}\left\{ \begin{array}{l} 100 \dfrac{\mathrm{MAX}(C_j) - \mathrm{AVE}(C_j)}{\mathrm{AVE}(C_j)} \\ 100 \dfrac{\mathrm{AVE}(C_j) - \mathrm{MIN}(C_j)}{\mathrm{AVE}(C_j)} \end{array} \right\} \tag{37}$$

together with the coefficient of variation,

$$\mathrm{CV}_{\mathrm{vol}} = 100 \frac{\mathrm{SD}_{\mathrm{vol}}}{\mathrm{AVE}(C_j)} \tag{38}$$

with

$$\mathrm{SD}_{\mathrm{vol}} = \sqrt{\frac{1}{(N_{\mathrm{ROI,v}} - 1)} \sum_{j=1}^{j} (C_j - \mathrm{AVE}(C_j))^2} \tag{39}$$

where $N_{\mathrm{ROI,v}}$ is the total number of ROIs within the volume.

6.1.4.6. Suggested tolerances

Visual inspection of the uniformity images should produce a final qualitative judgment of an image as being acceptable or non-acceptable.

The inter-slice and volume non-uniformity indices should be stable over time within a specified tolerance.

The user should set reference values, tolerances and action levels (i.e. to trigger the decision to make a call for maintenance). An appropriate tolerance criterion for the mean %NU is:

$$\%\mathrm{NU}_{\mathrm{measured}} < 1.05 \%\mathrm{NU}_{\mathrm{reference}}$$

6.1.4.7. Corrective action

If image artefacts are present or uniformity parameters are outside tolerance levels, daily QC should be rechecked and recalibration of the system considered. If the problem persists the manufacturer should be notified and maintenance scheduled.

6.1.5. PET normalization

6.1.5.1. Aim

The aim of this procedure is to acquire crystal efficiency data for use in correcting acquired sinograms for detector non-uniformities. The use of incorrect normalization data will compromise image quality.

6.1.5.2. Frequency

All manufacturers have a standard procedure for the acquisition of PET normalization data. Some manufacturers specify that this procedure should be performed monthly, but on certain systems where the procedure is more involved this is unnecessary and impractical. The frequency recommended by the manufacturer should be followed.

On systems where normalization is performed monthly, it can mask gradual deterioration of detector calibration. On these systems, it is recommended that detector calibration be performed on a quarterly basis, or prior to the monthly normalization if appreciable changes in QC values have been noticed over the preceding month.

The procedure should be performed by the qualified medical physicist, and additionally whenever the results of daily QC of PET indicate the need for renormalization or a service is carried out on the PET detector system.

6.1.5.3. Materials

Depending on the system and/or manufacturer and the acquisition mode (2-D–3-D), the normalization procedure can be accomplished using different sources and phantoms. The most widely used ones are:

(a) A rotating ^{68}Ge line source;
(b) A uniform cylindrical ^{68}Ge phantom centred horizontally and vertically in the FOV of the PET;
(c) A rotating ^{137}Cs point source.

The activities of these sources should be as specified by the manufacturer of the PET system.

6.1.5.4. Data acquisition

Before starting the acquisition, a backup copy of the previous normalization file should be made. Normalization data should be acquired following the instructions of the manufacturer.

6.1.5.5. Analysis

A visual inspection of the normalization sinograms should be made. If no major problems are observed, the new normalization data should be stored in a file, according to the flow chart established by the manufacturer.

6.1.5.6. Suggested tolerances

A visual inspection should be acceptable.

6.1.5.7. Corrective action

Recalibration of the system should be considered. If the problem persists, the manufacturer should be notified and maintenance scheduled.

6.1.6. 2-D–3-D Radioactivity concentration calibration

6.1.6.1. Aim

The aim of this calibration is to acquire scanner efficiency data for use in correcting acquired sinograms for detector non-uniformities. These factors are used in the calculation of radioactivity concentration and SUVs; inaccurate calibration factors will compromise accurate image based quantitation.

6.1.6.2. Frequency

All manufacturers have a standard procedure for the acquisition of 2-D–3-D radioactivity concentration calibration data; this procedure is referred to in different terms by different manufacturers (e.g. well-counter calibration, radioactivity calibration factors or SUV calibration). This procedure should be performed by the qualified medical physicist according to the manufacturer's specifications quarterly and when the PET detector system is serviced.

6.1.6.3. Materials

Depending on the system and the acquisition mode (2-D–3-D), the normalization procedure can be accomplished using different sources and/or phantoms. The most widely used ones are:

(a) A uniform cylindrical ^{68}Ge phantom centred horizontally and vertically in the FOV of the PET scanner;
(b) A fillable ^{18}F phantom (Fig. 35).

The activities of these sources should be as specified by the manufacturer of the PET system. Note that ^{68}Ge cylindrical phantoms cannot necessarily be relied on for scanner calibration as the stated radioactivity is sometimes substantially in error (up to 15–20%), and they do not allow cross-calibration with the dose calibrator as required for SUV calculation.

6.1.6.4. Data acquisition

Before starting the acquisition, a backup copy should be made of the previous calibration file. Calibration data should be acquired following the

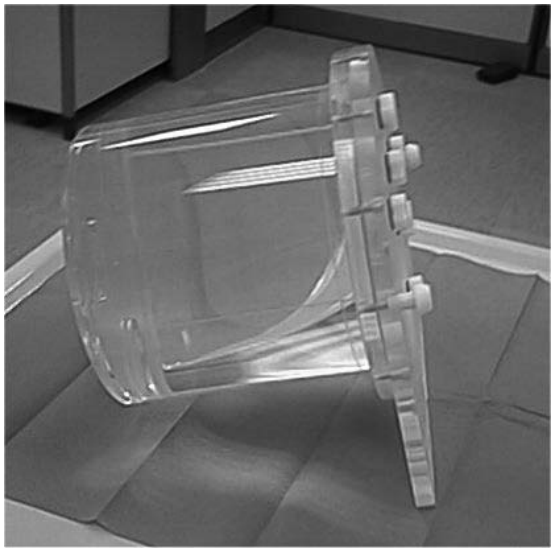

FIG. 35. A cylinder phantom to be filled with ^{18}F solution. Phantoms like this are used for normalization of the PET detector, for radioactivity concentration calibration and for uniformity testing.

instructions of the manufacturer. The acquisition should consist of an adequate number of total counts, in order to achieve good data statistics. When no indications are given or readily available, at least 20 million counts should be acquired.

6.1.6.5. Analysis

The calibration data obtained should be compared with previous results and the indications of manufacturers. If no major problems are observed, the new calibration data should be stored in a file, according to the flow chart established by the constructor. No visual inspection of data is usually needed.

6.1.6.6. Suggested tolerances

Comparison with previous results and the typical values given by manufacturers should be acceptable. Deviations greater than 5% from the expected results should result in corrective action being taken.

6.1.6.7. Corrective action

If significant deviations from expected results are observed, recalibration of the system should be considered. If the problem persists, the manufacturer should be notified and maintenance scheduled.

A useful optional procedure for checking SUV accuracy is to scan a phantom with a known amount of radioactivity and volume/weight using a multibed clinical protocol. If the radioactivity in the phantom, the calibration time of the radioactivity and the weight of the phantom volume are entered as part of the patient details, then the measured SUV should be 1.

6.1.7. Offset calibration for PET/CT

6.1.7.1. Aim of the calibration

To determine the x, y and z offsets required to register acquired PET and CT images, all PET/CT manufacturers have a standard procedure for the co-registration of PET and CT data. The use of incorrect offsets will result in misregistration errors in fused images and attenuation artefacts due to the use of misregistered CT data for attenuation correction.

6.1.7.2. Frequency

This test should be performed by the qualified medical physicist quarterly and whenever the PET and CT gantries are separated for service.

6.1.7.3. Materials

The materials required consist of:

(a) An alignment phantom comprising sources extending over the FOV of the PET (normally supplied with the system);
(b) Software to compute co-registration offsets from PET/CT images of the phantom.

6.1.7.4. Data acquisition

A PET/CT scan of the alignment phantom should be acquired according to the manufacturer's instructions.

6.1.7.5. Analysis

The x, y and z offsets required to register PET and CT images should be computed and stored in a file from where they can be accessed by PET/CT fusion software.

6.1.7.6. Suggested tolerances

Not applicable.

6.1.7.7. Corrective action

No corrective action is needed in this case.

6.1.8. Routine image quality test for PET/CT

6.1.8.1. Aim of the test

The aim is to monitor the consistency of image quality parameters using a widely available non-uniform phantom.

6.1.8.2. Frequency

The frequency test is optional and can be performed quarterly by the qualified medical physicist.

6.1.8.3. Materials

The materials required consist of a cylindrical phantom 20 cm in diameter and 20 cm in height with hollow spheres of differing diameters, as available from a variety of manufacturers, filled with 110–180 MBq ^{18}F and with all inserts installed. There should be no radioactivity in the spheres.

6.1.8.4. Data acquisition

Ensure that the phantom is centred horizontally and vertically in the FOV of the PET/CT. Acquire a PET/CT scan using a standard brain scanning protocol. Set up a PET scan for 20 million true events. Use the local standard clinical reconstruction protocol to reconstruct the PET data, including all corrections normally applied in clinical scans. The reconstructed image voxels should be in units of radioactivity/volume, for example, Bq/cm^3.

6.1.8.5. Analysis

Examine the reconstructed PET slices for any visible artefacts. Then perform the analyses described in the following subsections.

(a) Uniformity

In each of six equally spaced slices within the uniform section of the phantom, define a central circular ROI of diameter 20 mm smaller than the internal diameter of the phantom. Within the ROI calculate the integral uniformity, U, as:

$$U = 100\left(\frac{C_{\max} - C_{\min}}{C_{\max} + C_{\min}}\right) \tag{40}$$

where $C_{\max}$ and $C_{\min}$ are the maximum and minimum voxel values within the ROI, respectively.

(b) Radioactivity concentration

Within the same six slices and ROIs, compute the mean voxel value in Bq/cm^3. Compare this with the actual concentration at the time of the scan, and express the difference as a percentage of the known concentration.

(c) Spatial resolution

Visually determine the smallest diameter rods that can be distinguished in the reconstructed images. Record the diameter of the smallest diameter distinguishable rods.

6.1.8.6. Suggested tolerances

For each of the above parameters — uniformity, radioactivity concentration and minimum distinguishable rod diameter — if there is an apparent significant change in the parameter, check the daily QC results for possible signs of the cause. If the cause of the change is not apparent, perform further tests and if necessary schedule a service.

Note that the phantom *must* be positioned and oriented as reproducibly as possible each time this test is performed. It is *imperative* to keep records of all instances of this test, and *desirable* to have a facility to plot the parameter values over time.

6.1.8.7. Corrective action

If cold rods are not visible or if the uniformity exceeds established tolerances, daily QC should be rechecked and recalibration of the system considered. If the problem persists, the manufacturer should be notified and maintenance scheduled.

6.2. QUALITY CONTROL OF CT EQUIPMENT

6.2.1. CT laser alignment

6.2.1.1. Aim

This test ensures that the gantry lasers and room alignment lasers (for therapy treatment purposes) are properly aligned with the CT gantry and table.

This test is especially important if the images are being used for the planning of radiation therapy treatment.

6.2.1.2. Frequency

This test should be carried out at least monthly, and whenever the alignment lasers are serviced.

If images are to be used for radiation therapy treatment planning, this test should be carried out daily, or at least on those days prior to using the system for treatment planning purposes.

6.2.1.3. Materials, procedure, analysis and suggested tolerances

For details of the CT laser alignment procedure, refer to Section 5.2.2.

6.2.1.4. Corrective action

Necessary corrective action will depend on the specific application. If the PET/CT fused images are to be used for the planning of radiation therapy treatment, then ±1 mm is the maximum tolerance that should be accepted. Tolerances for other applications will be developed by the responsible medical physicist.

6.2.2. Tabletop alignment and positional accuracy, and scout scan accuracy

6.2.2.1. Aims

The aims of this test are to ensure that:

(a) The tabletop is level and orthogonal with respect to the image plane.
(b) Table and vertical motion according to digital indicators are accurate and reproducible.
(c) Table indexing and position under scanner control are accurate.

This test is especially important if the images are being used for the planning of radiation therapy treatment.

It should be ensured that the scout scan image accurately indicates the position of the patient.

6.2.2.2. Frequency

This test should be carried out at least monthly, and whenever the table or table gantry interlock system is serviced.

If the images are to be used for the planning of radiation therapy treatment, this test should be carried out daily, or at least on those days prior to using the system for treatment planning purposes.

6.2.2.3. Materials, procedure, analysis, suggested tolerances and corrective action

For details of the tabletop alignment and positional accuracy, as well as the scout scan accuracy procedure, refer to Section 5.2.3.

6.2.3. Computed tomography number and uniformity, image noise, and image artefacts

6.2.3.1. Aims

The aims of this test are to ensure that:

(a) The CT numbers for different materials are within the appropriate limits.
(b) The CT numbers are uniform over the image, the image noise is typical of what would be expected for the specific technique factors, and the images are free of artefacts.

6.2.3.2. Frequency

The frequency with which this test should be carried out is as follows:

(a) For CT numbers — monthly for different materials. For water, for radiation therapy applications — daily or before patient scans.
(b) For uniformity, noise and artefacts — monthly.

Note that in addition to the frequencies noted, these procedures must be carried out whenever an X ray tube is replaced, on system calibration, generator maintenance, software changes or upgrades, or on any other invasive service that may affect the CT number accuracy, uniformity, noise or image artefacts.

6.2.3.3. Materials

The materials required consist of a phantom filled with water and with regions of different densities, typically including polyethylene, acrylic polymers, Teflon and air.

6.2.3.4. Procedure

Place the phantom on the table centred at the isocentre of the scanner with the long axis of the phantom aligned with the z axis of the scanner[3]. Select a FOV that covers the phantom and about a 3 cm surrounding area. Acquire a scout scan and a single, 1 mm thick, image slice of the phantom for alignment purposes. Use these images to ensure that the phantom is centred in the FOV and aligned with the x, y and z axes.

Select the same standard clinical technique (typically 120 kVp and 200 mAs), acquire a single axial slice of 5 mm thickness, and ensure that the reconstruction algorithm is the same as that used for previous evaluations of CT number, uniformity, noise and image artefacts.

6.2.3.5. Analysis

Analysis proceeds as follows:

(1) Review the entire image for the presence of artefacts. If any artefacts are present, even subtle ones, compare this image with previous images to determine if these are new artefacts. If they are new, do not make any measurements of CT numbers, uniformity or noise before discussing the artefacts with the responsible medical physicist. If there are no new artefacts, or if told to do so by the responsible medical physicist, proceed with the measurements described in (2)–(4) below.
(2) Select an ROI sufficiently small to fit inside the regions of polyethylene, acrylic polymers, Teflon and air (4–10 cm^2), and measure the average CT number of each material in the phantom image, including air and water. Plot the values on the control chart.
(3) Use the same sized ROI to measure the average value and its standard deviation of the water in the centre of the phantom, and the average value of the water near the periphery of the phantom. The locations of

[3] Note that some scanners have special mounting brackets for reproducible phantom positioning.

these two areas should be the same as those measured previously for consistency of the results. Divide the standard deviation of the CT numbers of water by the average CT number of water, from the measurements made in the centre of the image. The result is the noise of the image.

(4) Subtract the average CT number of the water in the centre of the phantom from the average CT number of water obtained near the periphery. This is the uniformity value of the image.

6.2.3.6. Suggested tolerances

The suggested tolerances are as follows:

(a) Either there should be no artefacts in the image or the artefacts should be very subtle. Most importantly, there should be no new artefacts in the image compared with previous images of the same phantom.
(b) The values of CT numbers should be within ±5 HU of the values specified by the manufacturer.
(c) The image noise (the standard deviation divided by the average of the CT numbers of water in the centre of the phantom) should be less than ±10%.
(d) The uniformity (the difference between the average CT number at the centre and that at the periphery) should be within ±10 HU (CT numbers).

6.2.3.7. Corrective action

The responsible medical physicist should be contacted for consultation regarding how to proceed if it appears that corrective action is needed.

6.2.4. High contrast modulation

6.2.4.1. Aim

The aim of this test is to ensure that images with good modulation of high contrast objects, i.e. small details, will be imaged with good fidelity.

6.2.4.2. Frequency

This test should be carried out monthly or whenever the CT system is serviced in a way that might have an impact on the high contrast modulation.

6.2.4.3. Materials, procedure, analysis and suggested tolerances

Refer to Section 5.2.6 for details.

6.2.4.4. Corrective actions

Modulation can be increased by increased noise levels. If modulation has increased above the set tolerances, measure the image noise (Section 6.2.3) to ensure that the noise is within the tolerances set.

Low values of modulation will result if the phantom is not aligned parallel to the *x*, *y* and *z* axes of the scanner. If the modulation has fallen below the suggested tolerances, ensure that the phantom is at the scanner isocentre, aligned with the *x*, *y* and *z* axes, and that the bars are at 45º to the scanner's *x*–*y* axes.

Contact the responsible medical physicist for consultation regarding how to proceed with corrective actions.

6.2.5. Annual quality control tests

In addition to the QC tests described in Sections 6.2.1–6.2.4, there are additional QC tests that should be carried out on an annual basis by the responsible medical physicist. These are the same as the acceptance tests listed in Appendix II and include the tests described in Sections 5.2.1–5.2.9.

6.3. QUALITY CONTROL FOR PET/CT

6.3.1. Visual display and QC of hard copy image

6.3.1.1. Aim

The aim of this test is to ensure that:

(a) The visual display properly shows the clinical images.
(b) The quality of the film images is consistent over time.
(c) The quality of the film images matches the gray scale on the visual display.

6.3.1.2. Frequency

These tests should be carried out daily.

6.3.1.3. Materials

The materials required consist of:

— The SMPTE test image;
— A control chart for film density and contrast [47];
— Previous film images of the SMPTE test pattern.

6.3.1.4. Procedures

(a) Quality control of displays

The procedure for displays is as follows:

(1) Clean the front of the display with the appropriate cleaner and two soft cloths. Note that cleaner should not be sprayed directly on to the display. It should be sprayed on to a cloth, which is then used to clean the display. A dry soft cloth should then be used to dry the surface of the display. (Spray and solution on the display can result in liquids entering the electronics and causing problems.)
(2) Display the SMPTE test pattern on the display using the standard window and level settings.
(3) Ensure that the room lighting is the same as during acceptance testing in terms of the lights that are on.
(4) Examine the 5% patch inside the 0% patch and the 95% patch inside the 100% patch.

(b) Quality control of film images

The procedure for film images is as follows:

(1) Display the SMPTE test pattern on the visual display using the standard window width and level.
(2) Print the image on a film and process the film (if processing is required).
(3) View the film in a masked viewing box with a luminance of not less than 1500 nit.
(4) Measure the optical densities of the mid-density (MD) 0, 10 and 40% patches, and of the low density (LD) 90% patches.
(5) Determine the density difference (DD) by subtracting the density of the 40% patch from that of the 10% patch.

6.3.1.5. Analysis

Look closely at the visual display and review the film for artefacts. Ensure that the resolution patterns demonstrate acceptable detail and that the alphanumerical data are sharp and clear.

Review the control charts, ensuring that the MD, DD and LD values are within the control limits and that there are no indications of drift in any of the data, particularly over the past three to six measurements and in the longer term.

6.3.1.6. Suggested tolerances

The room light conditions should be consistent over time. No bright sources of light should be visible to the physician working at the visual display console. There should be no reflections on the display surface from lights behind or above the physician.

Both the 0 and 90% patches should be clearly visible on the visual display. The resolution in the centre and corners of the image should be the same as previously and similar to that seen on the film images. The high contrast resolution may be slightly less in the corners of the image than in the centre of the image.

Alphanumerical data should be sharp and clear.

The film should not exhibit any artefacts, and the resolution patterns, both low and high contrast, should be the same as those for previous films at all five locations (at the centre and four corners) of the image.

Verify that the MD, DD and LD values are within the control limits (Table 7) and that there are no apparent drifts in the values on the control chart.

6.3.1.7. Corrective actions

Any changes in the visual characteristics of the display or the room illumination conditions should be discussed with the responsible medical physicist.

TABLE 7. CONTROL LIMITS FOR FILM QUALITY CONTROL OF SMPTE TEST PATTERN IMAGES

Value	Patch (%)	Limits
MD	40	±0.15
DD	40–10	±0.15
LD	90	+0.05

The resolution and overall appearance of the film image should be similar to those in previous tests and similar to those of the visual display. The MD, DD and LD values should remain within the control limits, and no drifts should be apparent in the data. If any data points fall outside of the control limits, the test should be repeated immediately to rule out errors in technique or measurement. Any deviating results should be discussed with the responsible medical physicist.

6.3.1.8. References and suggested reading

For more in-depth discussions on these matters, see Refs [9, 44, 45].

6.4. CONSIDERATIONS FOR MOBILE PET/CT FACILITIES

Mobile PET/CT units are becoming increasingly widespread. They must meet the same standards of quality as fixed units. Therefore, all the procedures stated in this book are also required for mobile imaging systems. The qualified medical physicist responsible for QC (as stated in Section 4.2.5) must be clearly designated and recognized by both the equipment provider and the local organization hosting the mobile unit. The medical physicist should be available for immediate consultation about matters concerning QC of the equipment. Moreover, the medical physicist at each location should have access to a log of the current and recent QC data demonstrating the performance of the unit.

Appendix I

QUALITY CONTROL OF THE PET COMPONENT OF PET/CT

TABLE 8. RECOMMENDED FREQUENCIES FOR QC TESTS AND CALIBRATIONS FOR THE PET PART OF PET/CT SCANNER SYSTEMS

Test number	Procedure	At time of acceptance	Daily	Monthly	Quarterly	Annually	Other	Post-service
5.1.1	Spatial resolution	√					a	b
5.1.2	Sensitivity	√					a	b
5.1.3	Scatter fraction, count losses and randoms measurements	√						
5.1.4	Energy resolution	√[c]					a	b
5.1.5	Image quality and accuracy of attenuation and scatter correction	√				√	a	b
5.1.6	Coincidence timing resolution for TOF PET	√					a	

TABLE 8. RECOMMENDED FREQUENCIES FOR QC TESTS AND CALIBRATIONS FOR THE PET PART OF PET/CT SCANNER SYSTEMS (cont.)

Test number	Procedure	At time of acceptance	Daily	Monthly	Quarterly	Annually	Other	Post-service
5.3.1	Accuracy of PET/CT image registration	√					a	b
6.1.1	PET detector stability test		√					
6.1.2	Coincidence timing resolution tests		√					
6.1.3	Test of PET/CT scan in clinical mode		√[d]				a	
6.1.4	Uniformity	√			√		a	b
6.1.5	PET normalization	√		√			a	b
6.1.6	2-D–3-D Radioactivity concentration calibration	√		√			a	b

TABLE 8. RECOMMENDED FREQUENCIES FOR QC TESTS AND CALIBRATIONS FOR THE PET PART OF PET/CT SCANNER SYSTEMS (cont.)

Test number	Procedure	At time of acceptance	Daily	Monthly	Quarterly	Annually	Other	Post-service
6.1.7	PET/CT offset calibration	√			√		a	b
6.1.8	Routine image quality PET/CT test				√[d]			

[a] Whenever detector system performance is suspected to have changed significantly.
[b] This test should be performed whenever servicing is suspected to have affected the test results.
[c] If applicable.
[d] Optional procedure.

TABLE 9. TEST TOOLS AND CRITERIA FOR ACCEPTANCE AND QC TESTING OF PET SCANNERS AS PART OF PET/CT SYSTEMS

Test number	Procedure	Test tools	Tolerance criteria
5.1.1	Spatial resolution	Source holder; three point sources of F-18	$FWHM_{observed} < 1.05FWHM_{expected}$
5.1.2	Sensitivity	Phantom holder; 70 cm line source, five-sleeve aluminium phantom	$S_{TOT,observed} > 0.95S_{TOT,expected}$
5.1.3	Scatter fraction, count losses and randoms measurements	70 cm plastic cylinder with off-centre line source	$NEC_{observed} \geq NEC_{recommended}$ $SF_{observed} < 1.05SF_{expected}$
5.1.4	Energy resolution	Source holder; one point source of F-18	$RE_{observed} < 1.05RE_{expected}$
5.1.5	Image quality and accuracy of attenuation and scatter correction	IEC 61754 phantom [57]; 70 cm plastic cylinder with off-centre line source	Acceptable visual assessment
5.1.6	Coincidence timing resolution for TOF PET	Point source of a long lived isotope within a scattering material	$RT_{observed} < 1.05RT_{expected}$
5.3.1	Accuracy of PET/CT image registration	IEC 61754 phantom [57]; 100 kg load; ruler	±1 voxel
6.1.1	Daily PET detector stability test	Manufacturer supplied phantom	n.a.*
6.1.2	Coincidence timing resolution tests	Point source of a long lived isotope within a scattering material	$RT_{observed} < 1.05RT_{expected}$
6.1.3	Test of PET/CT scan in clinical mode	Manufacturer supplied phantom	n.a.*

TABLE 9. TEST TOOLS AND CRITERIA FOR ACCEPTANCE AND QC TESTING OF PET SCANNERS AS PART OF PET/CT SYSTEMS (cont.)

Test number	Procedure	Test tools	Tolerance criteria
6.1.4	Uniformity	20 cm cylindrical uniform phantom	$\%NU_{observed} < 1.05 \times (\%NU_{expected})$
6.1.5	PET normalization	Manufacturer supplied phantom	n.a.*
6.1.6	2-D–3-D radioactivity concentration calibration	20 cm cylindrical uniform phantom	n.a.*
6.1.7	PET/CT offset calibration	Manufacturer supplied phantom or IEC 61754 phantom [57]	n.a.*
6.1.8	Routine image quality PET/CT test	Laser alignment tool	n.a.*

* n.a.: not applicable.

Appendix II

QUALITY CONTROL OF THE CT COMPONENT OF PET/CT

TABLE 10. RECOMMENDED FREQUENCIES FOR QC TESTS AND CALIBRATIONS FOR THE CT PART OF PET/CT SCANNER SYSTEMS

Test number	Procedure	Responsible person	At time of acceptance[a]	Daily	Weekly	Monthly	Annually	Post-service
5.2.1	Scattered radiation measurements and shielding verification	Medical physicist	√				√	b
5.2.2	CT laser alignment	Medical physicist	√				√	b
5.2.3	Tabletop alignment and positional accuracy, and scout view accuracy	Medical physicist	√				√	b
5.2.4	Visual inspection and programme review	Medical physicist	√				√	
5.2.5	Display profile and width	Medical physicist	√				√	b
5.2.6	High contrast modulation	Medical physicist	√				√	b
5.2.7	kVp and half-value layer	Medical physicist	√				√	b
5.2.8	Radiation dose, image noise and image uniformity	Medical physicist	√				√	b

TABLE 10. RECOMMENDED FREQUENCIES FOR QC TESTS AND CALIBRATIONS FOR THE CT PART OF PET/CT SCANNER SYSTEMS (cont.)

Test number	Procedure	Responsible person	At time of acceptance[a]	Daily	Weekly	Monthly	Annually	Post-service
5.2.9	CT number and electron density accuracy	Medical physicist	√				√	b
6.2.1	CT laser alignment	Technologist		√	√			b
6.2.2	Tabletop alignment and positional accuracy, and scout scan accuracy	Technologist		√	√			b
6.2.3	CT number and uniformity, image noise, and image artefacts	Technologist		√	√			b
6.2.4	High contrast modulation	Technologist			√			b
6.2.5	QC tests	Technologist					√	b

[a] Acceptance testing occurs twice: firstly, before final payment and first patient use, and, secondly, four to six weeks before the end of the warranty period.

[b] Whenever there is a service that may affect the image quality, dose or other characteristics, and whenever the performance of the detector system is suspected to have changed significantly.

TABLE 11. TEST TOOLS AND CRITERIA FOR ACCEPTANCE AND QC TESTING OF CT SCANNERS AS PART OF PET/CT SYSTEMS

Test number	Procedure	Test tools	Tolerance criteria[a]
5.2.1	Scattered radiation measurements and shielding verification	NCRP Rep. 147 [42], AAPM Rep. 108 [41], isodose maps, shielding design, 32 cm CT dose phantom, sensitive radiation detector	Personnel doses less than maximum permissible dose; dose should be maintained ALARA[b], i.e. a fraction of the maximum permissible dose
5.2.2	CT laser alignment	Laser alignment device	±1 mm
5.2.3	Tabletop alignment and positional accuracy, and scout scan accuracy	Laser alignment device, 70 and 140 kg weights	±1 mm
5.2.4	Visual inspection and programme review	Checklist developed by responsible medical physicist, CT scan protocols	All items must meet local standards
5.2.5	Display profile and width	Phantom with thin metal plates, wires or an array of air holes at an angle to the scan plane	±1 mm
5.2.6	High contrast modulation	High contrast resolution phantom	±15%
5.2.7	kVp and half-value layer	Type 1100 filter set (up to 10 mm), pencil ionization chamber and electrometer. In cooperation with service engineer	HVL ≥ specified by manufacturer or radiation protection regulations

TABLE 11. TEST TOOLS AND CRITERIA FOR ACCEPTANCE AND QC TESTING OF CT SCANNERS AS PART OF PET/CT SYSTEMS (cont.)

Test number	Procedure	Test tools	Tolerance criteria[a]
5.2.8	Radiation dose, image noise and image uniformity	16 and 32 cm dosimetry phantoms, pencil ionization chamber and electrometer, DRLs for CT	Dose ±20% of manufacturer's specifications; noise: ±10%; CT number: ±5%; uniformity: ±10 HU for head, ±20 HU for body
5.2.9	CT number and electron density accuracy	CT phantom areas of different CT number materials, electron density phantom for radiotherapy	±5 HU for CT numbers specified by manufacturer
6.2.1	CT laser alignment		±1 mm
6.2.2	Tabletop alignment and positional accuracy, and scout scan accuracy		±1 mm
6.2.3	CT number and uniformity, image noise, and image artefacts	Laser alignment tool	CT number: ±5 HU; uniformity: ±10 HU; noise: ±10%; no artefacts
6.2.4	High contrast modulation	High contrast resolution phantom	±15%
6.2.5	Annual QC tests	See above procedures	See above procedures

[a] The tolerance criteria must be evaluated locally by responsible medical physicists based upon applications and needs.
[b] ALARA: as low as reasonably achievable.

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GLOSSARY

apodization. A technique in which the normal performance of an instrument is deliberately degraded in such a way that the instrument's performance is actually improved for one specific application.

attenuation correction. Correction of the loss of intensity of radiation as it passes through a medium. This may be caused by absorption or scattering. To correct the actual attenuation suffered in the body, the linear attenuation coefficient has to be measured. A transmission scan using radioactive sources or X rays is usually employed for its determination.

axial field-of-view. Dimensions of a slice through the tomographic volume, parallel to and including the system axis. In practice, it is specified only by its axial dimension, given by the distance between the centre of the outermost defined image planes plus the average of the measured axial slice width.

blank scan. A transmission scan done without the patient being on a couch. Such scans are used as part of the attenuation correction procedure.

coincidence detection. A method that checks whether two opposing detectors have each detected one photon simultaneously. By this method the two photons are treated as originating from the same annihilation event.

coincidence time window. The time interval during which two detected photons are considered to have been emitted simultaneously.

count rate. The number of detected counts per unit of time.

detector normalization. A calibration procedure by which the output of every detector is balanced in order to give the same response as for uniform irradiation.

effective dose. Effective dose is used in radiation protection, to compare the stochastic risk of a non-uniform exposure of ionizing radiation with the risks caused by a uniform exposure of the whole body. The stochastic risks are carcinogenesis and hereditary effects. It is not intended as a measure for acute or threshold effects of radiation exposure such as erythema, radiation sickness or death.

Effective dose equivalent is used to compare radiation doses on different

body parts on an equivalent basis, because radiation does not affect different parts in the same way. The effective dose, E, to an individual is found by calculating a weighted average of the equivalent doses, H_i, to different body tissues, with the weighting factors, W_i, designed to reflect the different radiosensitivities of the tissues: $E = \Sigma_i H_i W_i$.
The unit for effective dose is the sievert (Sv).

energy resolution. A parameter that denotes the ability of a system to differentiate two distinct energies. It is normally expressed by the FWHM (see below).

equivalent width (EW). The width of that rectangle having the same area and the same height as the response function, for example, the point spread function.

full width at half-maximum (FWHM). This term refers to resolution measurements (e.g. spatial and energy resolutions). Full width at half-maximum is usually measured from a profile through an image of a line or point source, or, in the case of energy, from the energy spectrum. The spread is due to resolution effects and is measured by the full width of the profile at a point that is half the maximum height of the profile.

full width at tenth maximum (FWTM). This is similar to the above mentioned term, using a profile level of one tenth of the maximum.

Hounsfield unit (HU). This is the numerical unit assigned electronically to each pixel in a computed tomography (CT) image, according to its X ray density. The fixed points on the scale are arbitrarily assigned as –1000 for air and 0 for water. The CT image is viewed in a 'window'. The range of Hounsfield units displayed (window width) and the centre point of the range of interest (window level) can be varied by the radiologist in order to observe specific tissues. The unit was named after Sir Godfrey Hounsfield (1919–2004), who developed CT scanning in the 1950s.

image artefact. (Also image artifact.) A term used generally in radiography to note the appearance on an image reflecting a problem with the radiographic technique rather than representing the actual anatomy/physiology of the patient. For example, a movement artefact is blurring of the image due to movement of the patient or organ during the exposure. All imaging techniques are susceptible to a range of artefacts.

image contrast. The contrast ratio is a metric of a display system, defined as the ratio of the luminosities of the brightest and the darkest colours the system is capable of producing. High contrast ratio is a desired aspect of any display, but with the various methods of measurement for a system or its parts, remarkably different values can be measured of the same subject. In CT or PET images, luminosity is equivalent to CT number or detected radioactivity, respectively.

image reconstruction. The process of obtaining a cross-sectional image from a set of projections. A reconstruction algorithm is a complex mathematical formula used by a computer to construct images from the data acquired by CT, MRI, PET or other scanners.

image registration. In computer vision, the sets of data acquired by sampling the same scene or object at different times, or from different perspectives, will be in different coordinate systems. Image registration is the process of transforming the different sets of data into one coordinate system. Registration is necessary in order to be able to compare or integrate the data obtained from different measurements.

Medical imaging registration (e.g. for data of the same patient taken at different points in time) often additionally involves elastic (or non-rigid) registration to cope with elastic deformations of the body parts imaged. Non-rigid registration of medical images can also be used to register a patient's data in an anatomical atlas.

image uniformity. A measure of how uniform the observed counts across the field of view are when the detector is irradiated by a uniform source. Integral uniformity is a measure of the maximum count deviation ((max – min)/(max + min)) over a given field of view. Differential uniformity is a measure of the maximum rate of change over a specified distance. Both shall be measured for the useful field of view (UFOV) and the centre field of view (CFOV).

line of response. The axis of the projection beam.

lower level energy discriminator (LLD). Threshold value used to reject pulses in an energy discriminator lower than the selected value.

noise equivalent count (NEC) rate. A quantity devised in order to estimate the useful count rate of a scanner, by taking into account, assuming Poisson statistics, the noise effects of scattering and randoms on true

coincidences. This quantity is typically estimated at different radioactivity concentrations.

point spread function. For tomography, a two dimensional point spread function in planes perpendicular to the projection beam at the specified distances from the detector.
Note that the physical point spread function characterizes the purely physical (intrinsic) imaging performance of the tomographic device and is independent of, for example, sampling, image reconstruction, and image reconstruction and image processing. A projection beam is characterized by the entirety of all the physical point functions of distance along its axis.

projection. Transformation of a 3-D object into its 2-D image or of a 2-D object into its 1-D image, by integration of the physical property that determines the image along the direction of the projection beam.

pulse pile-up. For imaging devices, false address calculation of an artificial event that passes the pulse amplitude analyser window, but is formed from two or more events by the pile-up effect. False measurement of the pulse amplitude, due to the absorption of two or more gamma rays, reaching the same radiation detector within the resolving time.

quality assurance (QA). The systematic process of checking to determine whether a product or service being developed is meeting specified requirements.

quality control (QC). A procedure or set of procedures intended to ensure that a manufactured product or performed service adheres to a defined set of performance criteria.

QMS. Quality management system.

radial resolution. Transverse resolution along a line passing through the system axis.

random coincidence. Result of coincidence detection in which both participating photons emerge from different positron annihilations.

rebinning. A mathematical procedure to reconstruct 3-D images from raw data, creating a stack of 2-D data sets. A 3-D image is reconstructed slice-by-slice from the data sets and the rebinned data are axially filtered to

reduce the blurring resulting from rebinning, the filtering being performed either before or after reconstruction of the 3-D image. The procedure used can be single slice or multislice rebinning.

scattered coincidence. A coincidence in which at least one participating photon was scattered before coincidence detection.

scatter fraction (SF). The ratio between the scattered true coincidences and the sum of scattered plus unscattered true coincidences for a given experimental set-up.

sensitivity profile. Also known as the axial sensitivity profile. Variation of the sensitivity of the PET scanner along its axial direction.

signal-to-noise ratio (SNR). This is often abbreviated SNR or S/N and is an electrical engineering concept defined as the ratio of a signal power to the noise power corrupting the signal. In image processing, the SNR of an image is usually defined as the ratio of the mean pixel value to the standard deviation of the pixel values. Related measures are the contrast ratio and the contrast-to-noise ratio.
In less technical terms, the SNR compares the level of a desired signal (such as music) to the level of background noise. The higher the ratio, the less obtrusive the background noise is.

singles rate. The count rate measured without coincidence detection, but with energy discrimination.

sinogram. A 2-D display of all 1-D projections of an object slice, as a function of the projection angle. The projection angle is displayed on the ordinate and the linear projection coordinate is displayed on the abscissa.

slice thickness. For tomography, the width of the axial point spread function.

spatial resolution. The ability to concentrate the count density distribution in the image of a point source at a point (tangential, radial and axial).

time-of-flight PET (TOF PET). A method for measuring the difference in time of detection of two photons, yielding information regarding the distance travelled by each of the photons from the location of positron annihilation in the field of view of a PET scanner. This information is used to improve the quality of a reconstructed image.

timing resolution. The resolution of the coincidence timing. On the basis of this property, a PET scanner can determine where an event originated between the two detectors, due to the different arrival times of the two photons. (See time of flight.)

tomography. This means literally 'drawing a body slice'. Tomography involves measurement from different angles around an object with the intention to 'reconstruct' an image of the internal distribution of some parameter (e.g., the radioactivity in SPECT or PET).

true coincidence. Result of coincidence detection of two gamma events originating from the same positron annihilation.

upper level energy discriminator (ULD). Threshold value used to reject pulses in an energy discriminator higher than the selected value.

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Consultants Meetings

Vienna, Austria: 21–25 November 2005,
6–12 June 2006, 5–9 March 2007

APPLYING RADIATION SAFETY STANDARDS IN NUCLEAR MEDICINE
Safety Reports Series No. 40
STI/PUB/1207 (124 pp.; 2005)
ISBN 92–0–111104–5
Price: €28.00

APPLYING RADIATION SAFETY STANDARDS IN DIAGNOSTIC RADIOLOGY AND INTERVENTIONAL PROCEDURES USING X RAYS
Safety Reports Series No. 39
STI/PUB/1206 (96 pp.; 2006)
ISBN 92–0–111004–9
Price: €28.00

APPLYING RADIATION SAFETY STANDARDS IN RADIOTHERAPY
Safety Reports Series No. 38
STI/PUB/1205 (120 pp.; 2006)
ISBN 92–0–110904–0
Price: 28.00

NUCLEAR MEDICINE RESOURCES MANUAL
STI/PUB/1198 (532 pp.; 2006)
ISBN 92–0–107504–9
Price: €65.00

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ISBN 978–92–0–103609–4
ISSN 2075-3772