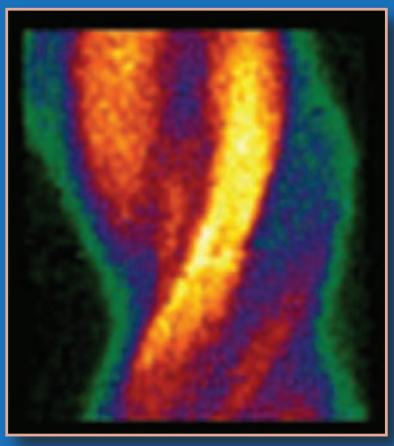

IAEA HUMAN HEALTH SERIES

No. 6

Quality Assurance for SPECT Systems

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QUALITY ASSURANCE FOR SPECT SYSTEMS

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IAEA HUMAN HEALTH SERIES No. 6

QUALITY ASSURANCE FOR SPECT SYSTEMS

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FOREWORD

Quality control is crucial to all aspects of nuclear medicine practice, including the measurement of radioactivity, the preparation of radiopharmaceuticals, the use of instrumentation to obtain images, computations to calculate functional parameters, and the interpretation of the results by the physician. It plays an integral part in fulfilling the regulatory requirement for establishing a comprehensive quality assurance programme as described in the International Basic Safety Standards for Protection against Ionizing Radiation and for the Safety of Radiation Sources. In 1984, the IAEA published IAEA-TECDOC-317, Quality Control of Nuclear Medicine Instruments, which addressed the quality control of radionuclide activity calibrators (also known as dose calibrators), gamma counters, and single and multiprobe counting systems, rectilinear scanners and scintillation cameras. An updated version of IAEA-TECDOC-317 was issued in 1991 as IAEA-TECDOC-602, and this included new chapters on scanner–computer systems and single photon emission computed tomography (SPECT) systems.

The rapidly increasing use of SPECT systems during the 1990s prompted the need for a further update of these publications with special emphasis on SPECT systems, planar scintillation cameras, camera–computer systems and whole body scanning systems. Since rectilinear scanners have already been, or will soon be, phased out in Member States, the current publication excludes them completely. Quality assurance and quality control aspects of instrumentation for radioactivity measurements in nuclear medicine are addressed in Technical Reports Series No. 454, Quality Assurance for Radioactivity Measurement in Nuclear Medicine.

The current publication is intended to be a resource for medical physicists, technologists and other healthcare professionals who are responsible for ensuring optimal performance of imaging instruments, particularly SPECT systems, in their respective institutions. It is intended for managers, clinicians and other decision makers who are responsible for implementing quality assurance/quality control programmes in nuclear medicine centres. It is hoped that it will play an important role in helping maintain image quality and lead to better utilization of nuclear medicine imaging instruments worldwide.

The IAEA Quality Control Atlas for Scintillation Camera Systems is intended to complement this publication. The Atlas provides many image examples of normal and abnormal quality control tests and should be consulted when performing the tests described here.

In the preparation of this publication the efforts of E. Busemann Sokole (Netherlands), R.Z. Stodilka (Canada), A.V. Wegst and R.E. Zimmerman (United States of America) are especially appreciated. The IAEA officers responsible for this publication were M. Dondi and S. Palm of the Division of Human Health.

EDITORIAL NOTE

CONTENTS

1. GENERAL CONSIDERATIONS

1.1. OBJECTIVE AND SCOPE

The objective of this publication is to provide professionals in nuclear medicine centres with detailed quality control test procedures for the scintillation camera and computer system. After studying this book, most qualified readers will be able to perform the three types of quality control tests, i.e. acceptance, reference and routine tests for the scintillation camera (also called the gamma camera) system for each of its imaging modes.

This publication focuses on the scintillation camera system, both in single and multiple head configurations, for obtaining images and quantitative data in planar imaging mode, whole body imaging mode and single photon emission computed tomography (SPECT). In addition, a section is devoted to the nuclear medicine computer of the camera system and quality assurance of nuclear medicine software. The final section addresses quality control of the digital image display.

Other nuclear medicine instruments, such as gamma counters and probes, are not discussed. Readers can find relevant information on gamma counters and probe systems in Ref. [1]. Tests and procedures for radionuclide activity calibrators (commonly known as dose calibrators) can be found in Ref. [2].

In addition to detailed descriptions of each quality control procedure, each respective section covers all tests for acceptance, reference and routine tests, recommended test frequency, test phantoms required and radiation sources, etc. Reference [3] complements this publication and assists the user with the evaluation of quality control test results.

Quality control is not a single action over a short period; instead, it is carried out through the whole life cycle of instruments, i.e. from planning and procurement to decommissioning. This process is described in general in this section and is applicable to all instruments of the nuclear medicine department.

1.2. QUALITY SYSTEM, QUALITY ASSURANCE AND QUALITY CONTROL IN NUCLEAR MEDICINE

It is widely recognized that the attainment of high standards of efficiency and reliability in the practice of nuclear medicine, as in other specialties based on advanced technology, requires an appropriate quality assurance programme.

The concept of quality in the term 'quality assurance' expresses the closeness with which the outcome of a given procedure approaches some ideal, free from all errors and artefacts. Quality assurance embraces all efforts made to this end. The term 'quality control' is used in reference to the specific measures taken to ensure that one particular aspect of the procedure is satisfactory (see Fig. 1). A clear distinction between these terms should be made.

A quality system in nuclear medicine should cover all aspects of clinical practice. It includes submission of requests for procedures; the preparation and dispensing of radiopharmaceuticals; the protection of patients, staff and the general public against radiation hazards and accidents caused by faulty equipment; the scheduling of patients; the setting-up, use and maintenance of electronic instruments; the methodology of the actual procedures; the analysis and interpretation of data; the reporting of results and, finally, the keeping of records.

This publication deals with a single, albeit highly important, component of such a comprehensive programme, namely, the quality control of instruments.

1.3. PRINCIPLES OF QUALITY CONTROL OF INSTRUMENTS

A fundamental principle in the quality control of nuclear medicine instruments is that the quality control should be undertaken as an integral part of the routine work of the nuclear medicine department and should be performed by members of the departmental staff themselves. However, some aspects must be carried out in collaboration with maintenance staff.

The quality control of each instrument should have as its starting point the selection and acquisition of the instrument itself, since instruments may differ widely in their characteristics and performance. The choice of an appropriate site for installation of the instrument should likewise be considered within the scope of quality control, since it may influence performance.

Once received and installed, an instrument should be submitted to a series of acceptance tests designed to establish whether its initial performance conforms to the manufacturer's specifications. At the same time, reference tests should be carried out to provide data against which its subsequent performance can be assessed by routine testing that is performed on a weekly, monthly, quarterly and annual basis. Finally, operational checks, carried out each day the instrument is used, should be put in force. Careful records of the results of all these tests should be kept and, if these reveal unsatisfactory performance,

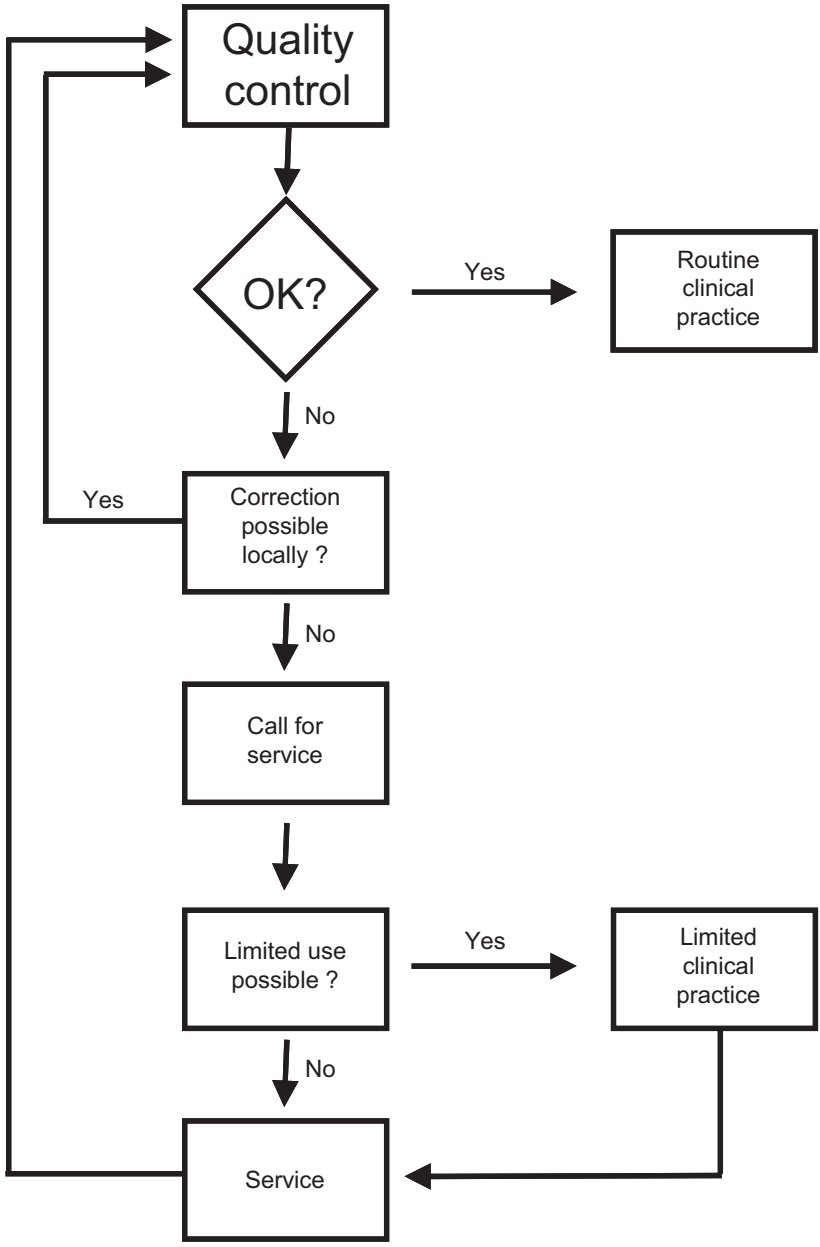

FIG. 1. Flowchart of a quality control programme for a scintillation camera.

appropriate corrective action should follow. These quality control procedures do not, of course, obviate the need for the usual preventive maintenance procedures, which should still be carried out on a regular basis. The success of such a scheme depends above all on the understanding and acceptance of all concerned. It further requires a clear definition of responsibilities and adherence to test schedules, protocols and proper procedures for the follow-up of test results.

1.4. SELECTION AND PROCUREMENT

The selection of an instrument with respect to manufacturer, model, etc., should be based not only on its suitability for the particular procedures to be carried out, as judged from its technical specifications, but also on such considerations as its ease of use (ergonomics), reliability and safety in operation, its compatibility with other instruments, the facilities and personnel available for its maintenance and the availability of spare parts. Technical advice on these points is often needed and the experience of other nuclear medicine centres can be valuable in this respect.

Considerable care is necessary in negotiating the purchase of an instrument. Full technical specifications should be solicited from manufacturers. Such specifications should cover all components in the instrument and all options, and should include power supply requirements; operational limitations as to temperature, humidity, etc.; size and weight bearing requirements; requirements for expendable items such as film and special paper for some specific printers as well as the availability of such items; and compliance with international and other standards. Quotations should indicate the price and terms; the date, mode and cost of delivery; the nature and duration of warranty; and the cost and specific coverage of service contracts. Also included in the quotations should be the manufacturer's arrangements for installation; the accessories, spare parts, manuals, test devices and expendables to be provided; the location and content of any training to be given to different categories of staff; the servicing facilities and personnel available; and the facilities for the supply of spare parts. Further, the quotation should detail the purchaser's arrangements for acceptance testing (perhaps in concert with the vendor), the minimum acceptable performance characteristics and the action to be taken if these are not met. Quotations should be compared with all these points in mind.

The wise purchaser will critically examine the servicing facilities and support personnel offered by different manufacturers or their representatives. An instrument with average performance characteristics but good servicing

facilities may well be preferred on grounds of reliability to one with outstanding performance characteristics but inadequate facilities for servicing. Maintenance of an instrument, including the supply of spare parts, has to be foreseen for its expected lifetime. This should be taken into account when costs are compared. Purchase price is an unreliable guide as to the total cost of an instrument, since it does not cover cost of repairs and regular contracted services to the nuclear medicine centre over the instrument's lifetime.

It is imperative that fully updated operation and service manuals, written in an appropriate language, accompany every instrument. Appropriate radiation sources, phantoms and other test devices needed for quality control should be provided or separately purchased at the time of instrument acquisition. It is also important to mention that the evaluation of offers and purchase orders should be jointly prepared by the responsible administrative and technical staff as a collaborative effort. This staff may involve physicists, physicians, technologists and administrators.

1.5. CARE, HANDLING AND PROTECTION OF EQUIPMENT

Owing to the sophistication and vulnerability of nuclear medicine imaging instruments, great attention and effort should be paid to preventive measures, namely, care, handling and protection against the following main environmental conditions:

(a) *Climatic environment of the instruments.* A good protection programme should provide effective air conditioning and humidity, dust and pollution control, etc.
(b) *Electrical environment of the instruments.* A good protection programme should provide effective AC line power conditioning against lightning, power line disturbances, electrostatic discharge and electromagnetic interference. The use of an uninterruptible power supply is advised to protect the system in the event of power failure.
(c) *Human environment of the instruments.* A good preventive programme should provide timely education of the operators, service engineers and technicians on the correct use and protection of the instruments. Only qualified and skilled service staff should be allowed to deal with the service and maintenance of sophisticated nuclear medicine equipment.
(d) *Magnetic environment of the instruments.* Nuclear medicine equipment is sensitive to magnetic fields and should not be located close to magnetic resonance imaging scanners or other strongly magnetic devices.

(e) *Background radiation.* A good protection programme would consider, during planning and installation, the location of major radiation sources, i.e. positron emission tomography facilities, X ray machines, linear accelerators and ^{60}Co devices for radiotherapy. Nuclear medicine instruments are extremely sensitive to these high energy sources and must be installed at appropriate distances from them. It is advisable to avoid location of radiopharmacy facilities close to the imaging rooms without proper attention to shielding. The storage and movement of radioactive materials, including patients, in the vicinity of nuclear medicine instrumentation should also be avoided.

It is imperative that all these protective measures be properly undertaken prior to installation and continue to be maintained during operation until decommissioning of the instrument, complying with current safety standards [4].

It is very important that all instructions from the operation manual and service manual for proper handling of the instrument should be carefully followed and that all the manufacturer's requirements for protection should be properly met before its installation.

1.6. PREVENTIVE MAINTENANCE

In contrast to the care, handling and protection programme, the preventive maintenance programme is designed and implemented against possible faults to the instruments. It should be periodically carried out and checked using the necessary quality control tests. A good preventive maintenance programme should include the following main procedures:

(a) Quality checks of parts, electronic circuits, components, connectors and cabling, etc.
(b) Inspection of detector/sensor condition.
(c) Checks of low and high voltage power supplies.
(d) Bias adjustment, preliminary adjustment of energy and position signals and preamplifier fine tuning, etc.
(e) Calibration of all correction circuits (e.g. energy, linearity, uniformity and attenuation corrections).
(f) Inspection of the integrity and stability of moving parts, with due consideration given to the lifetime and wear of components in frequent use, such as cables, relay switches, etc.

As in the case of corrective maintenance (repairs), preventive maintenance is machine dependent. As such, the protocols will differ from machine to machine. Usually, preventive maintenance is periodically carried out by qualified service engineers through contracted service. In addition, all documentation, including service manuals and circuit diagrams, necessary test tools and radiation sources, must be obtained at the procurement stage.

1.7. ACCEPTANCE AND REFERENCE TESTING

The acceptance of an instrument following its receipt and installation is a critical step towards the achievement of high quality performance and should be subject to correspondingly careful testing. Acceptance testing is undertaken to ensure that the performance of an instrument meets the technical and performance specifications quoted by the manufacturer. It should be carried out immediately after installation so that the supplier can be informed of any damage, deficiencies, or flaws before the warranty has expired. No instrument should be put into routine use unless it has been shown through acceptance testing to be performing optimally. An instrument that does not perform correctly at installation has a high likelihood of never doing so.

Acceptance testing is of concern to the maintenance staff, the manufacturer's agent and the eventual users of the instrument and all should be involved to some degree. As already indicated, it is important to establish during negotiations for purchase the manner in which such testing will be carried out and the minimum acceptable performance characteristics. Tests should be stringent and carried out according to clearly defined protocols. If they require specialized equipment, arrangements should be made for its provision. For acceptance testing of any major instrument, a representative of the manufacturer should always be present and should be able to initiate remedial action if specifications are not met. Otherwise, the onus for this falls on the purchaser. The practice of withholding payment of a part of the purchase price until acceptance testing has been satisfactorily completed is effective in many countries.

At the time of acceptance testing, reference tests should be carried out, from the results of which the subsequent performance of the instrument may be assessed during routine testing. These reference tests may be the acceptance tests themselves or less sophisticated versions of these that are more suitable for routine testing. Such tests should be repeated, as appropriate, to give a new set of reference data after major failure of the instrument and its subsequent repair, or when it is moved to a new site. Similarly, if for any reason an existing instrument did not undergo proper acceptance testing, the relevant tests should

be performed with the instrument in as good a working condition as possible at the time routine testing is initiated, in order to provide a set of reference data.

1.8. ROUTINE TESTING

Routine tests are those that should be carried out regularly on an instrument to ensure its optimum performance at all times and to determine the rate and extent of any deterioration in that performance with time. Such tests fall into two categories: the first includes tests that have been previously carried out as reference tests and that are repeated weekly, monthly, quarterly, yearly, etc.; and the second includes daily or operational checks that are to be carried out each day the instrument is used.

It is clear that routine tests should always be executed in a like manner if successive results are to be comparable. Therefore, they should be carried out according to clearly defined protocols. When appropriate, limits of acceptability for the results and courses of action to be taken if these limits are exceeded should be specified. Operational checks should be simple and so designed that they can be completed in an acceptably short time (e.g. 15 min for a scintillation camera), according to a defined sequence by an experienced person.

Unavoidably, test schedules constitute a compromise between what is desirable and what is feasible. The choice of tests and the frequencies with which they are carried out have to take account of the situation in the individual nuclear medicine department and the status of its instruments. It is important that staff in all categories develop an attitude of alertness to possible instrument malfunction and that all appropriate aspects of the nuclear medicine procedure are tested whenever clinical results are suspect. No schedule can be established for such occurrences.

1.9. INTERDEPARTMENTAL COMPARISONS, EXTERNAL ASSESSMENT AND ACCREDITATION

Interdepartmental (or interlaboratory) comparison studies are an integral part of a quality system. Such studies are based on acquiring an image of a test object with the scintillation camera system to be tested and then comparing the quality of this image and/or the quantitative results obtained with those from the other participating departments. An example is to evaluate an image by identifying potential ‘lesions’ in the image of the test object and assigning a confidence rating to each lesion suspected of being present. This

permits applying accurate and objective methods to measure the quality of an image. Such studies are useful in identifying malfunction of an individual imaging device and inadequate imaging practice.

Accreditation is the formal certification in relation to acceptability of a department to perform specific procedures, by an organization authorized to issue such a certification. In a rapidly increasing number of countries, success or failure of a department to achieve accreditation has a significant consequence on the reimbursement of nuclear medicine studies by the social security system. Quality control and the implementation of a quality system within the nuclear medicine department are requirements for successful accreditation. There are many components of the clinical service such as the training and competence of the staff members, the preparation and storage of the radiopharmaceuticals, referral policies and many others that contribute to the overall quality of the service [5]. The part of accreditation dealing specifically with nuclear medicine equipment addresses the calibration and quality control of equipment, validation procedures, calibration material, internal quality control, interdepartmental comparisons and how all these activities are documented.

1.10. QUALITY CONTROL RECORDS

Record keeping is of great importance in a quality management programme. The operational, quality control and maintenance records for each instrument should be assembled in appropriate log books and retained with the instrument. The records should include the results of the acceptance, reference and routine tests carried out for quality control, a record of preventive maintenance carried out and a record of failures, with details of their repair. The responsible person(s) should sign all entries. In addition, it is helpful to assemble and maintain a complete procedure manual detailing all clinical and test protocols. Indeed, such procedure manuals are required by accreditation organizations.

It is essential that all concerned appreciate the meaning of the records kept. Record sheets should be so designed that they are appropriate, easy to complete and easy to understand; explanatory notes should be provided, if necessary. Only essential data and results should be recorded; raw data can be kept in a separate book or file. Control charts and graphs displayed on the wall near the instrument are helpful in quickly ascertaining its long term stability and in stimulating regular testing. Images obtained in quality control testing should be kept in chronological order, either in a logbook or in electronic form, together with the relevant imaging parameters and the results of other quality control tests on the instrument. They should be frequently reviewed for

evidence of deterioration in performance, which may not be initially apparent. Records showing repeated failure and/or progressive degradation of performance provide unquestionable evidence for complete instrument overhaul or replacement. If the records or logbooks are set up as digital records, there must be a suitable backup of the records.

1.11. ORGANIZATIONAL ASPECTS

A basic requirement for the successful introduction of such a quality management system is that the head of the nuclear medicine department recognizes its necessity. The support of the administrative authorities is also required so that the means to carry it through can be secured. Detailed arrangements then have to be made, and responsibilities clearly defined, for acceptance and reference testing, routine testing, evaluation of test results and periodic review of results in relation to quality assurance as a whole. Regular meetings of all concerned, including both professional and technical staff, should be held for the latter purpose. Lack of adequate organization will foster a careless attitude in which tests are carried out irregularly, or only if malfunction is suspected. Proper quality control is impossible on such a basis.

A single person, generally designated the 'quality manager', should have supervisory responsibility for the entire quality management system and the authority to enforce it and act on its findings. This person should be responsible for overseeing all aspects of the quality system and should be involved in the evaluation and periodic review of the results. However, they need not actually undertake testing.

It is important that tests on a given instrument be carried out by a person or persons familiar with the instrument's use. Responsibility for daily and operational tests, at least, should rest with its regular users. This has the virtue of developing in the users an awareness of the principles of quality control.

If the results of a particular test do not fall within the specified limits of acceptability, a decision has to be taken whether or not to withdraw the instrument from operational use pending corrective action. It is also possible to limit its use to specified procedures, which may not be affected by the fault. Responsibility for such decisions should again be clearly defined. This is especially important when the test is carried out by a member of the paramedical staff. Normally, the responsibility lies with the physician head of the department, who is responsible for interpreting the clinical results from the instrument. The scheme should be sufficiently flexible to accommodate changes, based on accumulated experience, with respect to the tests included, their detailed protocols and the frequencies at which they are carried out.

The significance of such a scheme is not limited to the individual nuclear medicine department. In some countries, a comprehensive quality management system, including the quality control of instruments, is a prerequisite for hospitals to obtain accreditation. Links with national atomic energy and health authorities, professional associations and working groups are in any case desirable, as are contacts with manufacturers and their agents. Thus, certain tests scheduled relatively infrequently and requiring special test devices may be more conveniently organized on a national basis than within the individual department. (The routine control of accuracy of radionuclide calibrators, for example, may be undertaken in this manner by a central department having the necessary certified sources.)

Interdepartmental comparisons of instrument performance, often organized on a national, regional or even international basis, may be instructive and stimulating to participating departments, as well as being of considerable scientific interest. It should be realized, however, that such quality assessment or quality surveillance schemes, usually undertaken on an occasional basis and testing either the overall performance of instruments of a particular class (e.g. scintillation cameras) or even particular performance parameters of such instruments, are in no way substitutes for true quality management systems providing continuing control of all instruments in a department.

1.12. IMPLEMENTATION OF QUALITY CONTROL

The sections that follow contain recommended schedules and protocols for acceptance and routine testing of scintillation camera systems (planar, whole body scanning, single and multiple head SPECT systems), nuclear medicine computer systems and the digital display.

As previously indicated, the choice of tests and the frequencies at which they are carried out have to take account of the situation in the individual nuclear medicine department and the status of its instruments. Furthermore, it is not possible to draw up detailed test protocols applicable to all instruments in a particular class. Nuclear medicine departments should, therefore, modify the given protocols to suit their own individual instruments and test devices. What is indispensable is that once appropriate individualized schedules and protocols have been agreed upon, they should be strictly followed.

2. SCINTILLATION CAMERAS

2.1. INTRODUCTION

2.1.1. Basic principles, planar scintillation camera

The scintillation camera is an imaging device used in the practice of nuclear medicine. It utilizes a thin but large area thallium activated sodium iodide (NaI(Tl)) crystal as the radiation detector. The crystal is viewed by an array of photomultiplier tubes (PMTs). The design of scintillation cameras varies considerably, but to illustrate basic principles, the most common camera design will be described. Figure 2 depicts a section through the detector head and the key electronic components of a typical Anger type scintillation camera. Photons emitted by radionuclides in the patient or test source reach the crystal after first passing through a lead collimator. The collimator defines the direction of acceptance of the photons. Most collimators are of the parallel hole, diverging, converging, or pinhole type. More complex collimator designs, such as fan beam, are also used.

The crystal is viewed by photomultipliers from its back surface, either directly or through a light guide. The photomultipliers are all fed from a common high voltage supply and the voltage or gain is slightly adjustable at each tube. When a photon interacts with the crystal, it produces a light scintillation that spreads through the crystal and is detected by the PMTs. The fraction of the light that strikes the photocathode of each photomultiplier varies inversely with the distance of the photomultiplier from the point of interaction. The position of the photon interaction can be determined from the amplitude distribution of the pulses from the photomultipliers in the array caused by this single gamma ray interaction. This information is used to give a spatial location to the photon interaction defined in an X–Y coordinate system (see Fig. 3).

In an analogue camera, the pulses from all the photomultipliers are electronically processed after passing through a preamplification stage. At this point, the pulses are all simultaneously sent to X, Y and Z pulse arithmetic circuits. The X and Y circuits are networks that scale the pulse amplitudes in proportion to the X or Y position of the original interaction in the coordinate system. The result of this is two analogue signals: X and Y. The amplitudes are proportional to the spatial coordinates of the original interaction. In the Z circuit, the pulses are summed to provide a Z signal proportional to the total

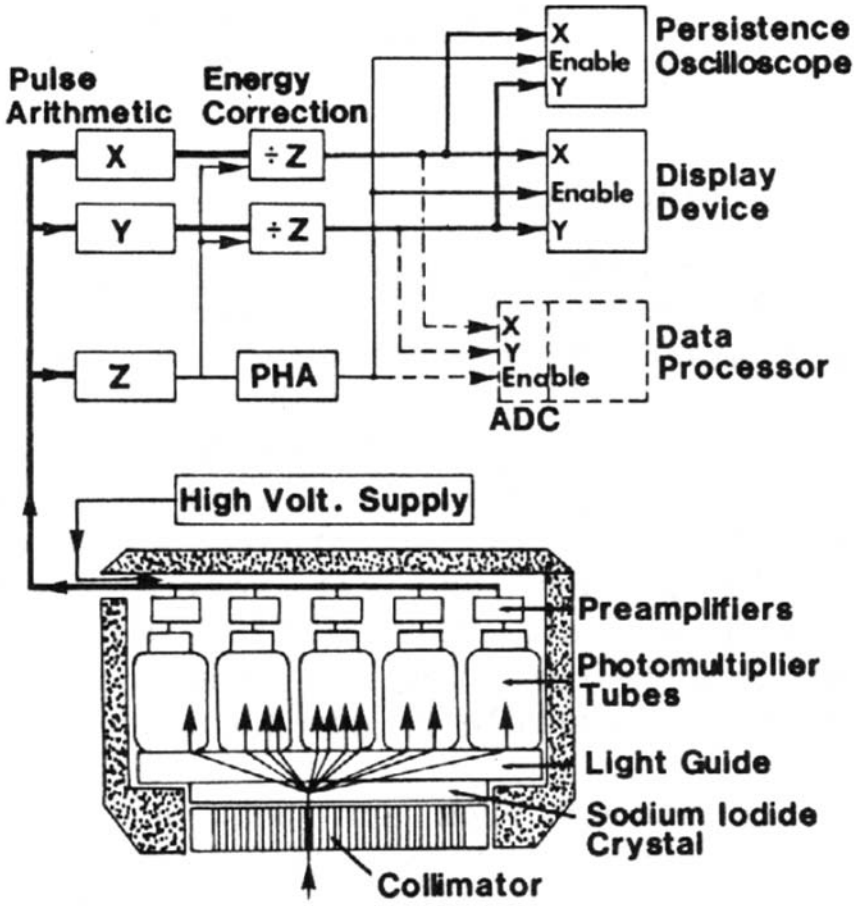

FIG. 2. Cutaway diagram of the detector head of an Anger type scintillation camera, with key electronic units.

energy deposited in the crystal by the photon interaction. Because the intensity of the scintillations increases with photon energy, and hence the photomultiplier output increases, the X and Y signals must be normalized so that the positional information is not dependent upon the photon energy. This is performed in the energy correction circuit where the X and Y signals are divided by the Z signal. Furthermore, the Z signal is sent to the pulse height analyser (PHA). If the Z signal falls within the PHA window set for the radionuclide in use, the PHA enables the X–Z and Y–Z signals to be recorded. In an analogue camera, this is usually achieved in a cathode ray oscilloscope.

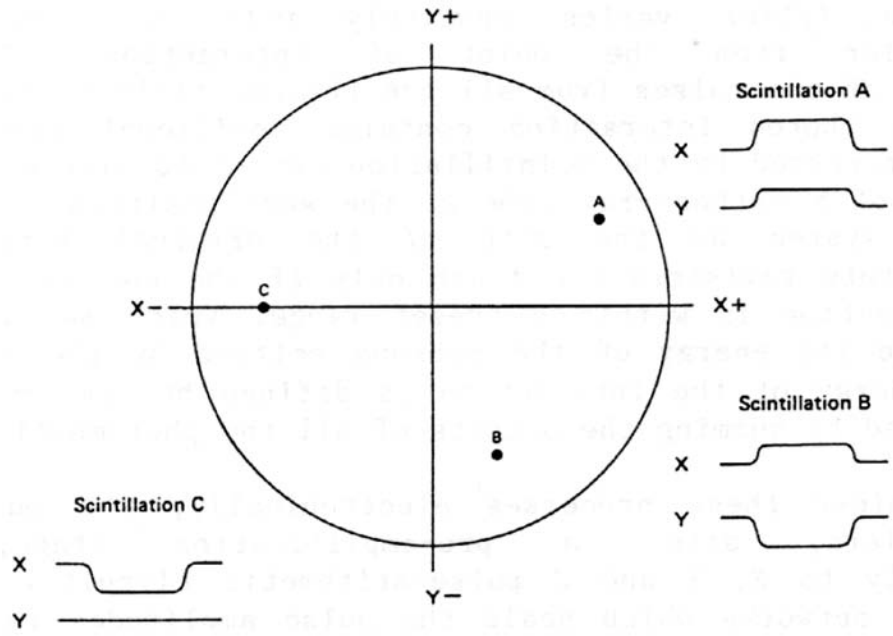

FIG. 3. The X–Y coordinate system of a scintillation camera, shown superimposed on the crystal face. Outside the coordinate system are shown examples of the X and Y signals (short duration voltage pulses) resulting from scintillation events occurring in different parts of the crystal.

The face of the oscilloscope is normally kept dark. This is achieved by blocking the electron beam from the oscilloscope face with a negatively biased grid. When the amplitude of the Z signal falls within the preset PHA window, an unblanking signal is generated, which causes the grid to become positive and allows the beam to pass. At the same time, the X–Z and Y–Z signals are used to deflect the beam, causing a brief flash to appear on the oscilloscope face at a position corresponding to that of the original scintillation. If a persistence oscilloscope is used, the flashes remain visible sufficiently long to form an image on the persistent phosphor screen. If a conventional oscilloscope is used, or an image formatting device incorporating such an oscilloscope, a permanent record is obtained by recording the flashes on film for a preset count or a preset time. The X–Z and Y–Z signals may also be digitized by analogue to digital converters (ADCs) for storage and later processing on a computer that is directly interfaced to one or more scintillation cameras. The Z pulse is used to start the digitization of the position pulses.

In ‘all digital’ scintillation camera designs, digitization is accomplished at each PMT and the pulse position is calculated by computer. This allows versatility not possible with analogue features.

2.1.2. Components of a planar scintillation camera

2.1.2.1. NaI(Tl) scintillation crystal

NaI(Tl) crystals are available in both circular and rectangular shapes. Mobile cameras and special cardiac systems have small field of view (300 mm) crystals. Large field of view (400 mm) crystals are available as multipurpose systems. Rectangular crystals with sizes as large as 400 mm × 500 mm and available in several thicknesses ranging from 3.2 to 25.4 mm are used in many single and dual headed systems. The crystal size determines, in part, the area of the patient viewed in a single image. The crystal thickness influences several performance parameters, in particular spatial resolution and sensitivity. Thin crystals yield better spatial resolution. However, their sensitivity is significantly reduced for photon energies over 140 keV. For general use, a thickness of 9.5 mm is often selected.

Any damage to the crystal results in an inoperable scintillation camera and requires costly replacement of the crystal. The large surface area as well as the hygroscopic and brittle nature of the crystal requires constant care to avoid puncturing the housing or otherwise damaging the crystal, especially in the process of changing collimators. Leaving a collimator on the scintillation camera head when it is not in use protects the crystal from mechanical shock and any rapid fluctuation in temperature. Nevertheless, sudden or gradual damage may occur unwittingly. For this reason, monitoring of the crystal is an important feature of quality control.

2.1.2.2. Photomultiplier array

All photomultipliers in the array, which may contain 37, 61 or even more tubes, must have matched amplification (gain) characteristics in order to provide a uniform count density (flood field uniformity) when the crystal is ‘flooded’ with a spatially uniform flux of photons. If one photomultiplier has a markedly lower gain than those surrounding it, the area of the image corresponding to the location of that tube will appear as one of lower sensitivity and if the tube fails, zero sensitivity. Such conditions are unacceptable in diagnostic imaging. Prior to the installation of tubes in a new instrument, the gains are carefully matched. However, each tube ages at its own rate, thus the gains must be periodically rematched by slight adjustment of the

high voltage to each tube. This is termed 'tuning the head' and is usually performed by a service representative of the manufacturer: the more photomultipliers, the more difficult the task. Daily quality control is necessary to alert the user to the need for this maintenance service.

The width of the photopeak is highly dependent upon the precise adjustment of the gains of the photomultipliers. Each photomultiplier produces a unique photopeak and when these are summed to form the Z signal, all of the photopeaks should coincide. However, because of small gain differences between individual photomultipliers, this is rarely the case; photomultipliers with gains lower than the average contribute information to the low side of the composite photopeak and those with gains higher than the average contribute to the high side (see Fig. 4).

In order to achieve a uniform flood field image, the window width of the PHA must encompass the contributions of all photomultipliers. For this reason, typically a 20% window is used. This window, centred on the 140 keV photopeak of ^{99m}Tc, would have a width of 28 keV, ranging from 126 to 154 keV. Such a window includes a significant amount of scattered radiation originating from photon interactions within the patient and leads to a loss of image resolution and contrast. Modern cameras allow the use of a 15% energy window because of linearity and energy correction circuits. If the window is offset to the high side (asymmetric high energy window) of the photopeak, the information contributed by the lower gain photomultipliers will be progressively eliminated and the image areas corresponding to these tubes will have a lower count density. Correspondingly, if the window is offset to the low side of the peak (asymmetric low energy window), the information contributed by the higher gain photomultipliers will be progressively eliminated and the areas corresponding to these tubes will have a higher count density (see Fig. 5).

If the window width is narrowed but remains centred on the photopeak, areas corresponding to photomultipliers both of lower gain and of higher gain will be progressively eliminated. Thus, uniformity across the field of view is a function of proper placement of the PHA window, which can only be achieved by daily calibration. Uniformity is also a function of the window width, photon energy and the proper tuning of all photomultipliers.

For a more detailed discussion and additional image examples, refer to Sections 2.2.1 and 2.2.2 of Ref. [3].

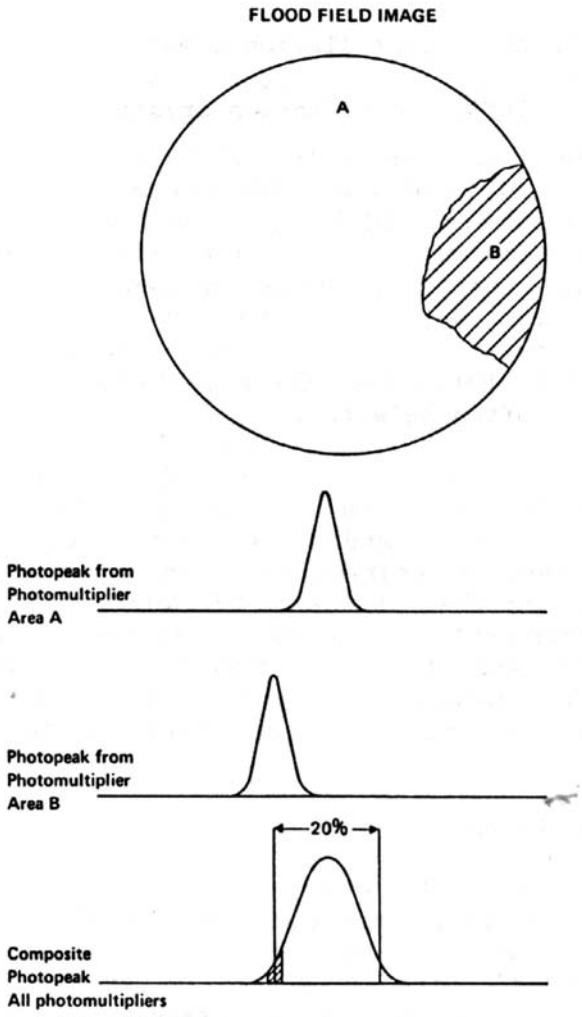

FIG. 4. Relationship between photomultiplier gain and flood field uniformity. In the flood field image, area A appears as one of uniform count density. Area B has perceptibly lower count density. The upper pulse height spectrum shows the photopeak from a photomultiplier within area A. The middle pulse height spectrum shows the photopeak from a photomultiplier at the centre of area B. The lower pulse height spectrum shows the photopeak of the Z signal, which is the composite of those from all the photomultipliers in the detector head, with the corresponding 20% PHA window and (cross-hatched) the position of the contributions from the photomultiplier at the centre of area B. A significant proportion of the pulses from the latter fall below the window and, hence, are rejected. This is the reason for the lower count density in the area in question.

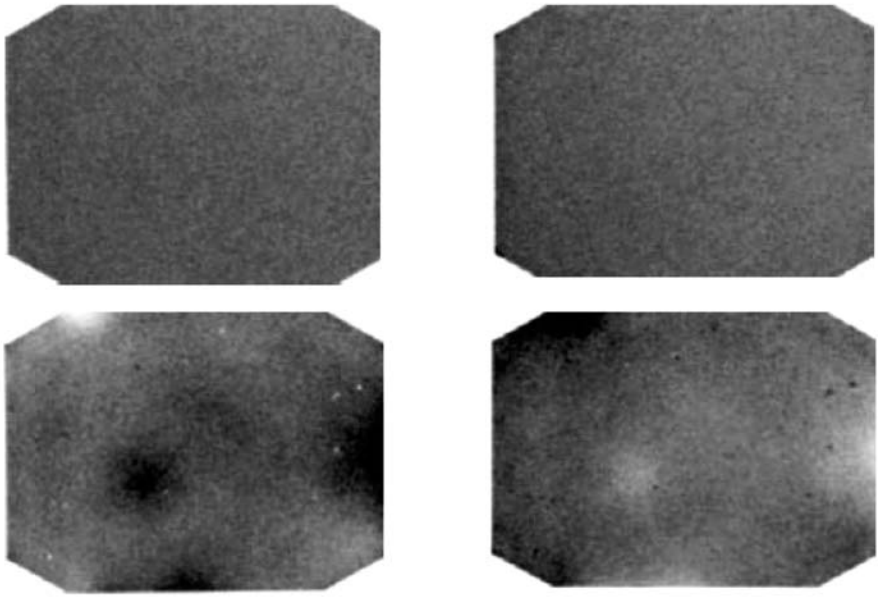

FIG. 5. Symmetric and asymmetric energy windows — poor PMT gain balance. Four million counts were acquired in each of the four intrinsic (collimator removed) quality control images. Images with asymmetric windows were acquired in order to check PMT balance. Top left: 20% symmetric energy window, approximately 30 000 counts/s. Top right: 20% symmetric energy window, approximately 75 000 counts/s. Bottom left: 10% asymmetric high energy window. Bottom right: 10% asymmetric low energy window. Results: The top left and top right images show that intrinsic uniformity with symmetric windows at moderate and high count rates is generally satisfactory. The images obtained with asymmetric energy windows reveal that some PMTs are out of balance. In the asymmetric high image, there are some regions with higher counts that appear darker, indicating that some tubes have higher gain than others. Tubes with lower gain appear as regions of reduced intensity. In addition, there are small 'cold' spots indicating crystal hydration. In the asymmetric low image (bottom right), the tubes that are hot or cold in the asymmetric high window image now appear as cold or hot, respectively. Crystal hydration now appears as small 'hot' spots (see Ref. [3]).

2.1.2.3. Pulse arithmetic circuits

In fully analogue scintillation cameras, the X and Y position circuits are separate but identical and contain amplifiers that, if properly adjusted, ensure equal amplification in both X and Y directions, i.e. a round object will give a round image. A drift of one amplifier will cause a round object to give an oval image. For this reason, the measurement of any object-to-image parameter should be performed in both X and Y directions. Object-to-image relationships may also be affected by non-linearity in the Z signal. This is of consequence only if the outputs of more than one PHA are used simultaneously to produce a composite image, for example, in ^{67}Ga imaging in which photons of two or three energies may be summed (see Fig. 6), or to produce a corrected image in which photons of a single energy are subtracted from those of another. If non-linearity exists in the Z signal, when the X and Y signals are divided by the Z signal, the spatial amplifications for different Z signals will differ. The superposition of several images will then result in a loss of resolution.

In cameras of more recent design, such as all digital cameras, this problem is minimized.

For a more detailed discussion and additional image examples, refer to Section 2.4 of Ref. [3].

2.1.2.4. Pulse timing circuits

The duration of the X, Y and Z pulses has a significant effect upon the count rate capabilities of the scintillation camera. The pulses resulting from each scintillation must be processed through the camera electronics, whether or not the event is finally selected for display, because of the acceptance of its Z signal by the PHA. Within this processing period, there is a time, the pulse-pair resolving time, *T*, largely determined by the duration of the pulses, during which the camera electronics are not capable of responding to further scintillations. At high count rates, the camera behaves largely as a 'paralysable' system; that is, every subsequent scintillation that occurs during this dead time extends it. Thus, if the intensity of the incident photons and the input count rate (the count rate that would be observed if there were no count loss) increase, the observed count rate increases to a maximum (the maximum count rate) and then decreases as an increasingly larger proportion of the scintillations occur during the extended dead time. This phenomenon is accompanied by a marked decrease in image quality (see Fig. 7). Digital camera systems handle pulses somewhat differently, but all exhibit a maximum counting rate.

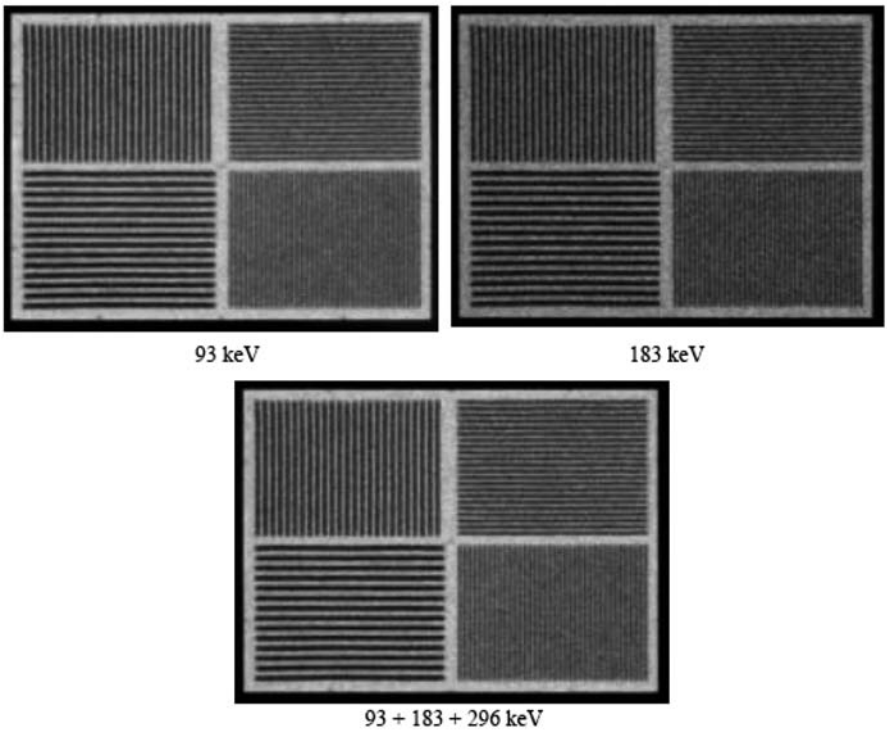

FIG. 6. Gallium-67 multiple window spatial registration: visual method — acceptable. Intrinsic quadrant bar pattern images obtained with a ^{67}Ga point source, 20% energy window over each photopeak, 512 × 512 matrix, bar widths of 3, 4, 5 and 6 mm, acceptance testing. All three images are similar and there is no loss of image quality from superimposing images from the three energy windows. The results are acceptable (see Ref. [3]).

The measurement of the count rate performance of a scintillation camera can be performed in several ways, depending upon the age and the design of the electronics of the system. For digital systems, the count rate performance is best tested using a decaying source and by repeatedly determining the observed count rate from a decreasing input count rate. The observed range must start beyond the maximum count rate and continue until the observed count rate reaches a low count rate (<4000 counts/s). From these data, the observed count rate at which 20% of the counts are lost can be determined ($C_{-20\%}$). This can be done with or without scatter. An alternative method is to use calibrated copper filters to reduce the count rate progressively. In older systems, T can be measured under conditions of only moderate count loss and with no radiation

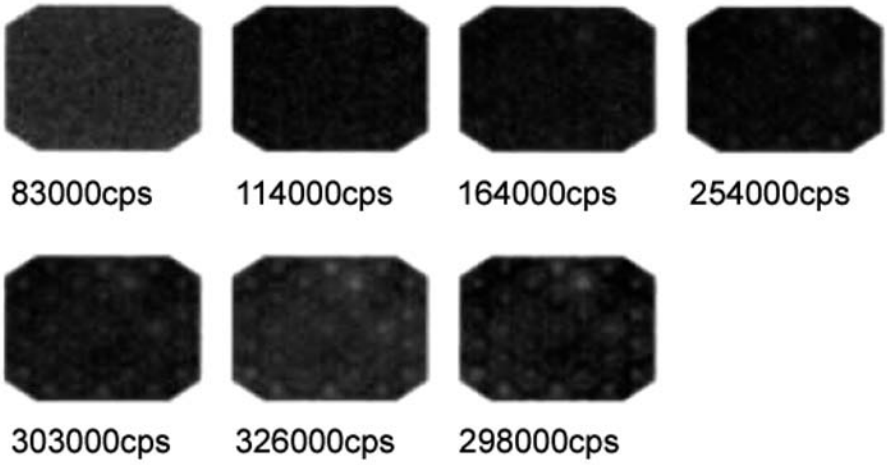

FIG. 7. Changes in uniformity at high count rates. Intrinsic ^{99m}Tc images of 4 million counts were acquired with a 20% energy window for different source activities, producing different count rates, given under each image. Top row from left to right: 83 000, 114 000, 164 000 and 254 000 counts/s. Bottom row from left to right: 303 000, 326 000 and 298 000 counts/s. Results: With increasing count rate the uniformity deteriorates. Above 114 000 counts/s the PMTs were visualized as cold areas. Note that in the bottom row the maximum count rate of the scintillation camera has been reached. In the last image, the maximum count rate has been exceeded, resulting in a reduced count rate in response to increased source activity. In practice, clinical work is unlikely to produce count rates higher than the lowest value shown here (see Ref. [3]).

scatter, using a two source method. From its value under these conditions, it is possible to deduce the relationship between input and observed count rates and to calculate, for example, the input count rates, $R_{-20\%}$ for a 20% count loss, and the corresponding observed count rates, $C_{-20\%}$. These methods constitute useful acceptance tests since $R_{-20\%}$ measured with no radiation scatter is a performance index specified by many camera manufacturers. The method used must be determined by consulting the product literature or the manufacturer's representative. (It must be noted that the 1994 version [6] of the United States of America's National Electrical Manufacturers Association (NEMA) test protocols does not include the two source method to assess count rate performance. It is included as an acceptance test in this publication since many camera specifications still include this parameter.)

Parameter $R_{-20\%}$ measured with no radiation scatter is not, however, relevant to clinical situations. In clinical imaging, scintillations due to lower energy scattered radiation arising from the patient, while not themselves displayed, may significantly increase the effective value of *T*. Both $R_{-20\%}$ and $C_{-20\%}$ measured with radiation scatter are lower than those measured without scatter: $C_{-20\%}$ measured with scatter should not be exceeded in any clinical study. Operating the camera at higher observed count rates may compromise its spatial resolution and will only give a small increase in observed count rate for a large increase in administered radioactivity and, hence, radiation dose to the patient.

To test the count rate performance of the scintillation camera further, the intrinsic flood field uniformity and spatial resolution should be measured at count rates of approximately 75 000 counts/s.

For a more detailed discussion and additional image examples, refer to Section 2.5 of Ref. [3].

2.1.2.5. Energy, linearity and uniformity correction circuits

Several schemes have been introduced to improve the uniformity across the field of view by microprocessor techniques. The first to be introduced was based upon either adding or subtracting counts to each of the approximately 4000 elements (pixels) of a 64 × 64 matrix. The number added or subtracted is derived from the sensitivity of that pixel relative to the mean of all pixels in a previously stored flood field image. These methods are no longer used.

Scintillation cameras of newer design use a multistage process. First, to take account of photomultiplier gain variations, a small correction is applied to each Z signal, dependent upon its specific X–Y location, so that the photopeaks for all locations coincide exactly (energy correction). This results in a narrower composite photopeak and allows the use of a narrower PHA window. The second stage is the application of a small correction to each X and Y pulse, dependent upon its specific location, to eliminate spatial non-linearity (linearity correction). The correction is often derived by using an image of a series of line sources, in both X and Y directions, and computing the deviation of the image from the actual lines over the face of the crystal. A third stage may utilize a count normalization based on a previously acquired high count calibration flood (uniformity or sensitivity correction). The final image is uniform to approximately 2–3% and is essentially free of spatial non-linearity.

Some cameras use gain stabilization to compensate for PMT drift using a pulsed light source, fibre optically fed to the crystal near each photomultiplier so that the individual gains can be adjusted every few milliseconds. This technique was developed for rotating cameras to stabilize photomultiplier gain

with the orientation of the camera relative to the Earth's magnetic field and with temperature changes.

2.1.2.6. Display devices

A scintillation camera may be equipped with several types of display device for the purpose of visualizing the radioactive concentrations as detected by the camera. Older analogue systems use an oscilloscope that produces a flash of light on the face of a cathode ray tube at the same position on a similar X–Y coordinate system as the site of the original interaction in the crystal, or uses a multiformat imager that can place 1, 2, 4 or more images on one sheet of film by a moving lens system. Digitized scintillation camera images are displayed on a computer monitor. The quality control of display systems is presented in Section 7.

2.1.3. Basic principles, camera–computer systems

Scintillation camera–computer systems are designed to allow the collection, digital analysis and display of the image data from a scintillation camera. In fully integrated systems, the camera bed motion, uniformity correction and image collection parameters are controlled at the computer console. The components of the computer in such a system are essentially the same as those of a computer used in any other application, i.e. a central processing unit (CPU), memory and magnetic storage device. Additional hardware items necessary for nuclear medicine applications include ADCs, which convert the analogue signals: from the camera to digital numbers that the computer is able to manipulate and display as a graph or image.

The analogue image information produced by the scintillation camera normally consists of three signals: the X and Y signals, which represent the position of the photon interaction in the crystal, and the unblanking signal, which indicates that the energy of the interaction falls within the PHA energy window set for the radionuclide in use. In some cameras, the energy signal, Z, is also provided so that complete energy spectra as well as images can be collected from the camera. If the camera is all digital, the image data may be transferred to the computer through a direct digital interface.

2.1.4. Components of a camera–computer system

2.1.4.1. Analogue input

Special line driver circuits are commonly used to drive the low power scintillation camera signals to the computer. The use of the drivers not only ensures that the signals are not distorted but also protects the camera circuits from being damaged by the extra electronic load. The line drivers may also be used to alter the voltage levels of the signals so that they are of the magnitude required by the computer interface. Most systems have sample-and-hold circuits that retain the values of the position signals during the time that the computer is processing a detected event, even if the camera removes the signals from the line. Failures in these circuits may produce artefacts in the digital image, but usually will not affect the analogue operation of the camera. If the analogue and digital images differ, these circuits should be considered as potential sources of the problem.

2.1.4.2. Analogue to digital conversion

The X and Y position signals must be converted to digital numbers to be processed by the computer. There are several types of ADC found in camera–computer systems. The most common is the successive approximation converter, which makes sequential estimates of the required numbers. Starting with the bit representing the largest power of two, the converter sets the bit and then converts the binary number to an analogue signal through a digital to analogue converter. The amplitude of this analogue signal is compared with that of the signal being converted (see Fig. 8). If the signal being converted is smaller, the bit is turned off; if it is larger, the bit is left on. The ADC steps through each of the bits in the digital word, performing this process each time. For an 8 bit digital word (256 position values), the conversion takes eight cycles.

For a more detailed discussion and image examples, refer to Sections 2.2.2I–2.2.2L and 5.1 of Ref. [3].

2.1.4.3. Data processing

The data processor in the context of this publication includes the CPU and the memory of the computer. The CPU in a conventional computer is the section that controls the timing and operation of the overall system. It also includes the arithmetic processing unit that performs the calculations and

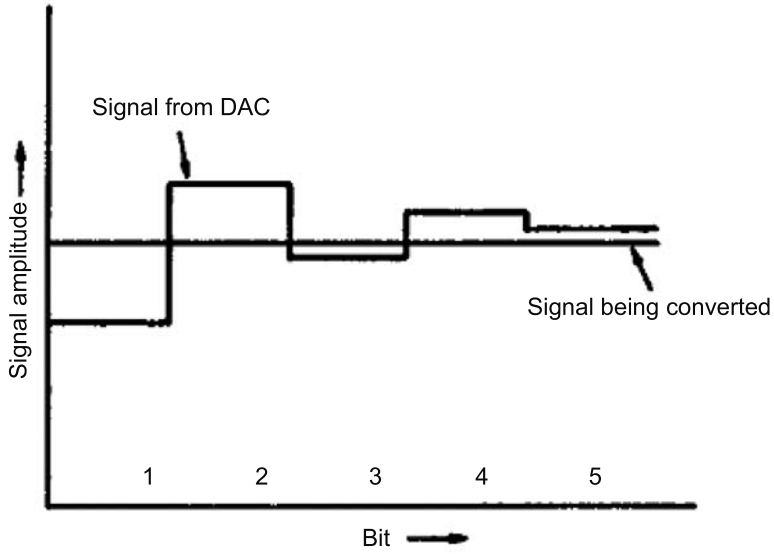

FIG. 8. Successive approximation of analogue to digital conversion. The binary number corresponding to the analogue signal being converted is approximated bit by bit. At each step, the resulting analogue signal from a digital to analogue converter is compared with the signal being converted.

makes logical decisions. In newer computers, the boundaries of the CPU are less clear, as the single CPU is being replaced by distributed microprocessors.

Although this is important to the system designer and to a certain extent to the user, it is not important for the understanding or execution of the quality control tests to be discussed.

The computer memory consists of a series of storage locations, or bins, into which data can be placed as words for later retrieval and manipulation. Memory is characterized by the number of storage locations and the size of the individual word. The number of locations determines the amount of data and the size of programs that can be present at any given moment. The size of the memory word determines the magnitude of the number that can be stored at a given location as a binary number. Some word sizes have been given special names. The most common is the byte, which refers to a group of 8 binary digits or bits. In general, the size of the memory word determines the counts that can be collected in a digital image. Some computers allow the user to select the size that will be used for image collection. Use of an 8 bit (byte mode) storage element allows the collection of a count of only 255 per image element (or pixel). Use of a 16 bit (word mode) storage element accommodates numbers of

up to 65 535 or ±32 767 per pixel, depending on the particular computer. Computers may use other word sizes, some using up to 32 bits.

The use of an 8 bit storage element for nuclear medicine imaging may represent a limitation and a potential source of error. In imaging procedures in which the radiopharmaceutical is concentrated in a small anatomical area, the pixels corresponding to this area quickly become saturated. This is also true for test procedures that require imaging small point or line sources. Depending on the particular computer, the computer may: (i) stop collecting, (ii) continue collecting in the non-saturated areas while holding the saturated pixels at 255, thus severely distorting the quantitative data, or (iii) continue counting and allow the saturated pixel to 'roll over' and lose multiples of 256 counts. Each of these may cause distortion of the quantitative data unless the system is capable of performing a suitable correction. It is important for the user to understand the clinical significance of such limitations and to choose the data collection mode appropriate to the clinical study to be performed. With the speed and capacity of modern computer systems, the use of an 8 bit storage element is no longer necessary.

2.1.4.4. Image formation

The output from the ADC is used in one of two ways by the computer during data acquisition: list mode and frame mode. In list mode (see Fig. 9), the digital data representing the coordinates of photon interactions in the crystal are simply stored as lists in memory. This is analogous to a person recording

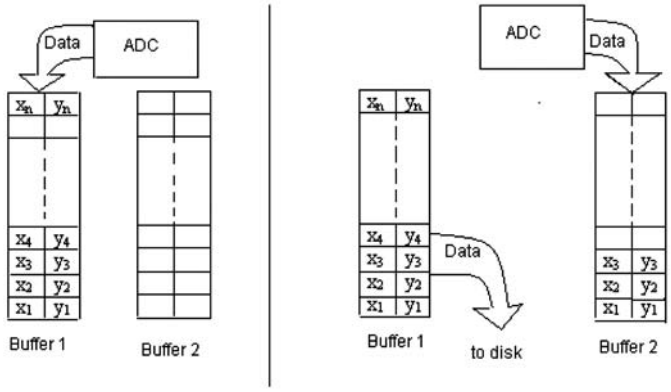

FIG. 9. List mode acquisition. The digital data from the ADC are stored as a list in a memory buffer and are subsequently written to disk for the construction of an image.

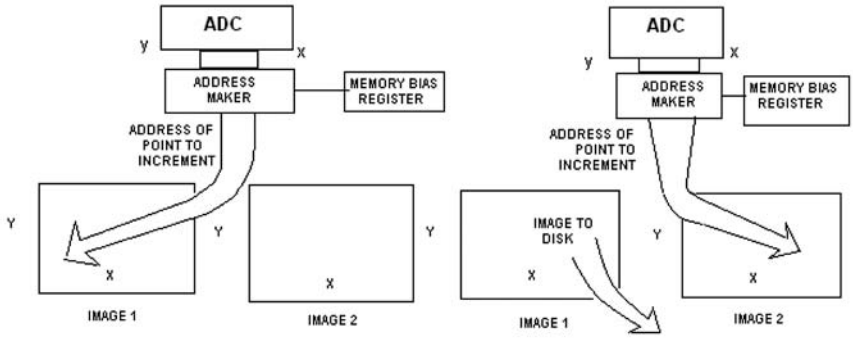

FIG. 10. Frame mode acquisition. The digital data from the ADC are used directly to construct an image in the memory by successively increasing, incrementally, specific memory locations. The image data may then be written directly to disk.

numbers on a sheet of paper. In frame or histogram mode (see Fig. 10), the digital data are used to identify the address of a specific memory location corresponding to the location of the interaction. The contents of this memory location are then increased incrementally by one. Frame mode collection constructs an image in memory buffers during collection while list mode only generates a list of interaction coordinates. Dynamic flow studies can be performed in frame mode by periodically writing the images to disk and restarting the collection in memory. A modified form of frame mode, termed electrocardiogram (ECG) gated acquisition, is often used for cardiology studies. In this mode, the data acquisition is synchronized by the patient's ECG. In such gated acquisition, a series of frames are generated, each one representing a small segment of the cardiac cycle as shown in Fig. 11.

The number of pixels in the array or matrix into which the digital image is divided determines the capability of the computer to retain the spatial resolution provided by the scintillation camera. A camera with a larger field of view requires a larger matrix to provide the same spatial resolution in the final digital image. The choice of matrix size for a particular clinical study should be based on the analytical requirements of the study. A study that is performed primarily to perceive fine detail requires a finer matrix than one performed simply for the generation of time–activity curves from large regions of interest. The relationship between matrix size and field of view is given in Table 1, in which the size of the area represented by a single pixel is given in millimetres.

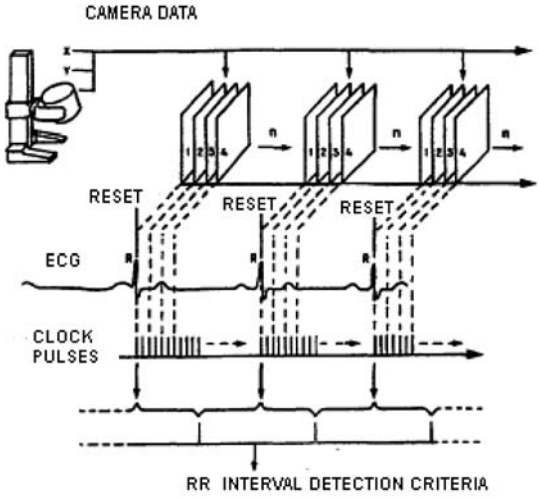

FIG. 11. ECG gated acquisition. The digital data acquired over a large number of cardiac cycles are used in conjunction with the R-wave of the ECG to construct a series of images in memory, each accumulated during a small segment of successive cycles.

TABLE 1. RELATIONSHIP BETWEEN SCINTILLATION CAMERA FIELD OF VIEW, MATRIX SIZE AND PIXEL SIZE

Field of view (cm)	Approximate size of a single pixel (mm)			
	64 × 64 matrix	128 × 128 matrix	256 × 256 matrix	512 × 512 matrix
10	1.6	0.8	0.4	0.2
20	3.1	1.6	0.8	0.4
30	4.7	2.3	1.2	0.6
40	6.2	3.1	1.6	0.8

Aside from the question of spatial resolution, the choice of matrix size has an impact on the expected counts per pixel. For a given imaging situation, a change from one matrix size to the next higher, e.g. from 64 × 64 to 128 × 128, reduces the count per pixel by a factor of four, since the image is distributed over four times as many pixels. Thus, a finer matrix can sometimes be used to prevent pixel saturation, although with the use of 16 bit storage elements this should not be a problem.

2.1.4.5. Data storage and transfer

It is necessary to provide supplementary storage in addition to that provided by the memory of the computer, for two reasons. The first is that data and information, i.e. programs and operating systems, must be transferred between computers. The second reason is that nuclear medicine imaging procedures generate a significant amount of data that must be stored for later retrieval and analysis. Magnetic storage is achieved by the use of two types of media: magnetic disk and magnetic tape. Disks are used for rapid storage and retrieval, while tape, which is slower, is used more often for long term storage and exchange between dissimilar systems. Other media that are used in nuclear medicine include optical disks: compact disk recordable, digital versatile disk and magneto optical disk.

Data are recorded on magnetic disk by read–write heads that pass over the surface of the disk in prescribed circular tracks creating small magnetized zones. The disk surface is divided into a number of storage blocks onto which the image data and programs are placed by the computer. The number and size of the data blocks are dependent on the particular disk design. In modern disk systems, the total storage capacity is extremely large and data transfer rates are very fast. Such high transfer rates may be required in high count rate studies in which counts are written to disk during collection. It is important to understand that the modern computer disk unit is a precision electromechanical device that must be properly cared for. Without appropriate preventive maintenance and careful handling, the disk unit will fail long before it should and thus prevent the rest of the computer system from operating. Failure can also result in the loss of clinical data files and software.

Optical disks write information to light sensitive disks using a modulated laser beam. Properly cared for, optical media remain stable for many years and are frequently used for archiving important data. Bulk storage and access to a number of disks through a device known as a 'jukebox' allows quick storage and retrieval of large quantities of data.

The transfer of data from one manufacturer to another and from one digital medium to another, i.e. from computed tomography (CT) scanner to scintillation camera–computer, is now possible through standards developed for formatting, transferring and archiving files. Interfile allows transfer of nuclear medicine images from one manufacturer to another and a NEMA standard, DICOM [7], allows transfer from one modality to another across most manufacturers that are implementing the standard. The appeal of viewing and fusing images from different modalities is generating strong support for DICOM.

2.1.4.6. Image display and hard copy

The image display is usually presented to the user on a high resolution monitor. Points on the screen have an intensity or colour related to the count of the corresponding pixel in the array. Most displays have their own dedicated image memories. A smaller secondary memory, sometimes termed a transformation table, colour table, or lookup table, is also used to map the count information of the image at the desired intensity levels of the display. The use of this table makes it possible to alter the contrast, brightness, or grey scale of the display without modifying the actual image data.

The computer display can be transferred to film or paper by a multiformatter that views a monitor. Colour paper, transparency printers and laser image processors are also available which will reproduce the information displayed on a computer screen by video image capture. In addition to the images, this allows high quality reproduction of graphs, 3-D displays and alpha numeric information. A full discussion of the quality control of display devices can be found in Section 7.

2.1.4.7. User interaction

The user may interact with the display through devices such as a light pen, a joystick, a track ball, a touch pad or a 'mouse'. A light pen is a light sensitive pointer aimed directly by the user at the selected part of the image. A joystick is a small resistive device adjusted by the user. The computer continually monitors the position of the joystick in both the X and Y directions and places a cursor on the display screen at a point having coordinates corresponding to the position indicated by the joystick. A mouse similarly places a cursor on the display screen at a point under its control. These devices may be used to indicate regions of interest (ROI), single points, or anatomical landmarks.

2.1.5. Basis of schemes for testing scintillation camera performance

Various levels of performance testing are required in the life of any scintillation camera. Initially, manufacturers perform a set of tests on each camera in the factory to determine if published specifications are met. In the USA, factory testing is done according to protocols developed by NEMA. Results for each camera are compared with the published specifications before the shipment is authorized. The NEMA performance standards [8] are recognized throughout the world. Hence, for new cameras, one manufacturer's specification is directly comparable to another's. The tests involved are mostly intrinsic tests, that is, they are tests on the camera without a collimator or other

accessories. These tests reflect only the camera's characteristics, not necessarily its operating performance under clinical conditions. Alternative test protocols are those of the International Electrotechnical Commission (IEC) [9–12]. The user should be aware of which test protocols have been followed by the manufacturer to provide the specifications for a particular system.

The camera should be acceptance tested by the user after installation to determine if, once installed, it performs according to the specifications of the manufacturer. This testing must be rigorous and similar enough to the NEMA or IEC protocols so that comparable results are obtained. When comparing test results with specifications provided by the manufacturer as measured according to NEMA or IEC protocols, the user must be aware of the energy window width applied.

While performing acceptance tests, reference tests should be initiated. Reference tests reflect operating performance under clinical conditions, can be repeated in routine testing, and are often system tests performed with collimator mounted, added accessories and a variety of clinically used radionuclides. Reference tests are more suited to being carried out by the user and can be adapted to local conditions and requirements. A number of organizations have developed test protocols for reference tests, among them the American Association of Physicists in Medicine [13–15] and the Institute of Physics and Engineering in Medicine [16]. These tests, along with some acceptance tests, provide the basis of routine testing. Last, but most important, operational checks must be initiated which are to be performed each day that the instrument is used.

2.1.6. Performance characteristics

Only proper testing can determine whether a scintillation camera is operating as it should. The performance characteristics evaluated in acceptance and routine testing will now be identified, along with the major design and operational factors that influence them.

2.1.6.1. Spatial resolution

Spatial resolution is a performance characteristic of a scintillation camera that describes its capability to resolve two separate point or line sources of radiation as separate entities. Spatial resolution is conventionally quantified either as the full width at half maximum (FWHM) (see Fig. 12) of the response to a thin line source perpendicular to the long axis of the source, or as the minimum separation of two sources that can just be distinguished from each

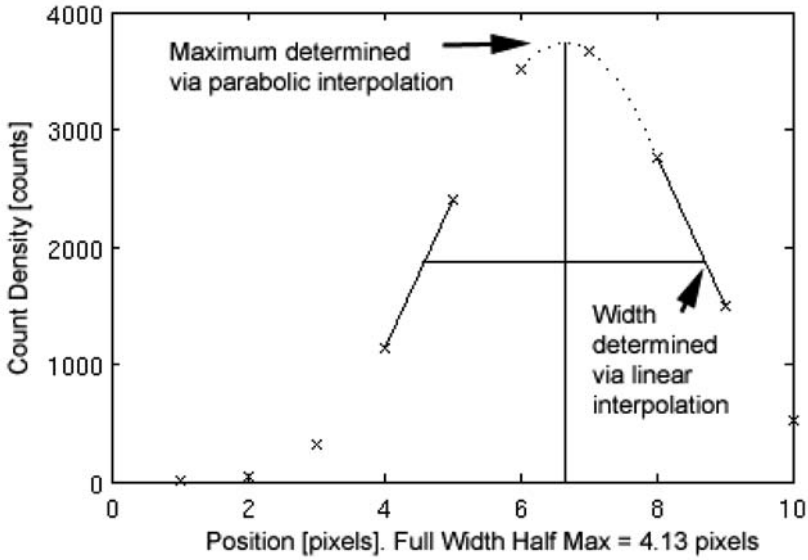

FIG. 12. Example of FWHM calculation, adapted from NEMA NU-1 2001 protocols. Given a profile through a line spread function (crosses indicate measured values), the maximum is determined via parabolic fit (dotted curve) through the largest measured value and its two nearest neighbours. The maximum interpolated value is then determined from the local maximum value of the fitted parabola. The half maximum locations are determined by linear interpolations from the nearest two neighbours of the half maximum value (see Ref. [8]).

other. (Thus, a small width or separation corresponds to good or 'high' spatial resolution.)

The spatial resolution exhibited by the detector alone is called the intrinsic resolution, R_i. The collimator alone exhibits a spatial resolution, R_c, which is best when the source is located at the surface of the collimator and deteriorates as its distance from the collimator increases. The system resolution, R_s, of the detector with the collimator mounted can be estimated for a source positioned at any stated distance from the collimator by:

$$R_s = \sqrt{R_i^2 + R_c^2} \tag{1}$$

In general, intrinsic spatial resolution improves with an increase in the number of photomultipliers for the same crystal diameter (implying a decrease in the diameter of each tube) or the energy of incoming photons and with a decrease in the thickness of the crystal or light guide, the width of PHA window, the proportion of scattered photons and the count rate. Collimator resolution improves with an increase in the number or length of holes and a decrease in the diameter of holes or thickness of septa.

The major factors that degrade intrinsic spatial resolution are electronic component failure; poor alignment of the gains of the photomultipliers; defects in, or deterioration of, the crystal and high count rate. In some cameras, switching to the high count rate mode decreases spatial resolution. System resolution is affected by the collimator used and degrades as the distance from the radiation source to the collimator surface increases.

Another major factor that affects the spatial resolution in a digital image is the sampling of the image, i.e. the number of digital picture elements or pixels. Increasing the area of the camera face corresponding to the digital image without a corresponding increase in the matrix size degrades the spatial resolution (see Table 1). This is an operational characteristic of digital systems and should not be considered a system failure.

2.1.6.2. Energy resolution

Energy resolution describes the capability of the scintillation camera to distinguish between photons of different energies, in particular between primary and scattered radiation. It is conventionally quantified as the FWHM of the photopeak, measured in energy units. This is expressed as a percentage of the gamma radiation energy (%FWHM). (Thus, a small %FWHM corresponds to good or 'high' energy resolution.)

The major factors that degrade energy resolution are poor alignment of the gains of the photomultipliers; failure of one or more photomultipliers; defects in, or deterioration of, the crystal; physical separation of the photomultiplier–light guide assembly from the crystal and high count rate.

Most scintillation cameras have electronic methods of energy correction that align the photopeaks produced by each of the PMTs to improve the overall energy resolution. This is usually done by collecting an energy spectrum in each pixel of a 64 × 64 or finer digital array. For each spatial location, the shift in the Z pulse necessary to align all photopeaks is calculated and stored in a lookup table. These energy correction tables are periodically recollected over the life of the camera, ensuring improved operation.

2.1.6.3. Response to uniform irradiation (flood field uniformity)

The response to uniform irradiation (flood field uniformity) is a performance characteristic of a scintillation camera that describes the degree of uniformity of count density in the image when the detector is flooded with a spatially uniform flux of incident photons (see Fig. 13). It may also describe the degree of constancy of count rate from a collimated point source when the source is moved over the field of view.

Flood field uniformity may be quantified as the degree of uniformity exhibited by the detector itself (intrinsic uniformity) or by the detector with collimator mounted (system uniformity). It may also be quantified in terms of the maximum variation in count density over the entire field of view (integral uniformity) or in terms of the maximum rate of change of count density over a specified distance (differential uniformity). (Thus, a small variation or rate of change corresponds to good or 'high' uniformity.)

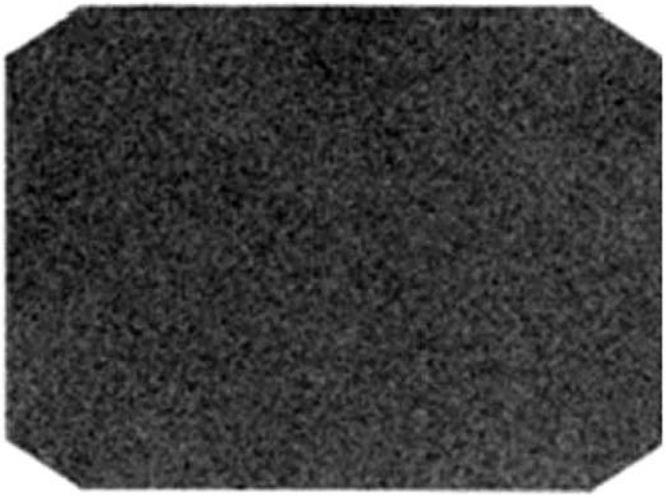

FIG. 13. Routine intrinsic uniformity image, ^{99m}Tc, 3 million counts, 20% energy window set symmetrically over the 140 keV photopeak of ^{99m}Tc. The image shows good uniformity. The most basic and sensitive routine quality control test of a scintillation camera is that of uniformity. This must be performed carefully (preferably daily before using the camera for clinical studies), it must be critically evaluated and any necessary action must be undertaken before further imaging takes place (see Ref. [3]).

FIG. 14. Defective PMT. Daily quality control image of flood field uniformity, ^{99m}Tc, 15% energy window, 4 million counts. Results: The image shows a large, circular cold area that was due to a defective PMT. Note the inner halo of lower counts and the outer halo of higher counts at the edge of the defect. Service is required (see Ref. [3]).

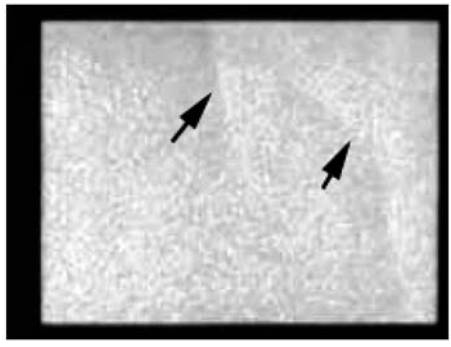

FIG. 15. Loss of optical coupling. A 2.5 million count intrinsic flood image acquired with ^{99m}Tc. Regions of altered intensity with clearly defined borders (arrows) are seen in the image. This was caused by a decoupling between the PMTs and the glass exit window of the crystal housing (see Ref. [3]).

The major factors that degrade intrinsic uniformity are poor alignment of the gains of the photomultipliers; failure of one or more photomultipliers (see Fig. 14); spatial non-linearities; defects in, or deterioration of, the crystal; physical separation of the photomultiplier–light guide assembly from the crystal (see Fig. 15); incorrect setting of the position or width of the PHA window and high count rate. Additional factors that degrade system uniformity are defects in, or damage to, the collimator (see Fig. 16).

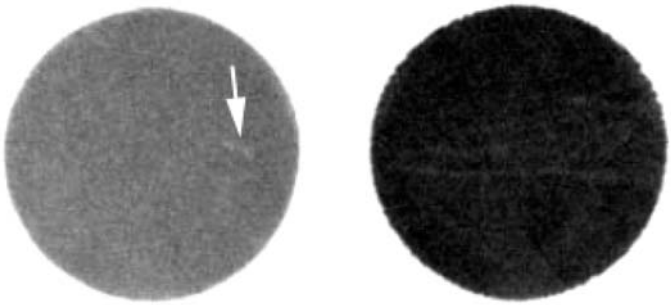

FIG. 16. System uniformity — collimator septa damage — scraped. Routine system uniformity, ^{99m}Tc flood source, 20% energy window, 3 million counts. Low energy high resolution (left) and high energy (right) collimators on the same scintillation camera. The flood field image from the low energy high resolution collimator (left) shows a short diagonal line of decreased count density (white arrow) due to septa that were damaged when the collimator was removed from the detector head. The results of the damage are occluded septa and diminished sensitivity. On the flood image from the high energy collimator (right), various irregular lines of reduced intensity are visible in the image. The septa were deformed by the collimator having been dragged across the mounting hardware on the detector (see Ref. [3]).

For a more detailed discussion and additional image examples, refer to Section 2.2 of Ref. [3].

2.1.6.4. Spatial distortion (spatial non-linearity)

Spatial distortion is a performance characteristic of a scintillation camera that describes the amount of spatial distortion of the image with respect to the object. Spatial non-linearity describes the degree of non-linearity in the image of a linear object. Spatial non-linearity may be quantified as the maximum spatial displacement in the image over the field of view and can be estimated by inspecting the image of a linear object. (Thus, a small displacement corresponds to good or high linearity.)

Spatial distortion and flood field uniformity are closely related. If severe spatial displacements occur, the uniformity will be poor in the same areas. The major factors that degrade spatial distortion are the same as those listed for flood field uniformity.

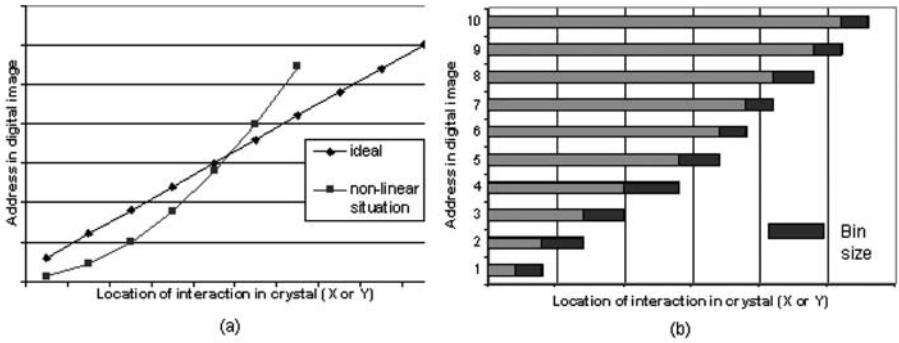

FIG. 17. (a) Integral non-linearity. The relationship between distance on the scintillation camera face and distance on the digital image varies across the image. (b) Differential non-linearity. The bin size varies in an irregular manner over the image.

2.1.6.5. Integral and differential ADC linearity

ADC linearity describes the capability of an ADC to convert accurately an analogue position signal to a digital address or location and is directly related to spatial distortion and flood field uniformity.

An ideal system should give a linear relationship between the location of an interaction in the crystal and the corresponding address in the digital image (see Fig. 17(a)). This should be true for both the X and the Y directions. Poor integral linearity in an ADC causes the relationship between distance on the camera face and distance on the digital image to vary across the image. It is difficult to detect without precise quantitative measurements.

In an ideal system, the sizes of all the bins are equal (see Fig. 17(b)). In an ADC with poor differential linearity, bin size varies in an irregular manner over the image. Poor differential linearity in an ADC causes stripes and lines to appear in the digital image. The effect is usually seen in both the X and the Y directions, i.e. both horizontal and vertical lines appear, but in some instances only one axis is affected (see Fig. 18).

The major factors that degrade integral linearity in a camera–computer system are a poorly calibrated analogue amplifier or a failure in the camera itself. Differential non-linearity may be present in an ADC as the result of faulty power supplies, which allow transients to affect the conversion process. Another possible cause of differential non-linearity is improper matching of circuits in the analogue part of the camera–computer interface.

For a more detailed discussion and additional image examples, refer to Section 5.1 of Ref. [3].

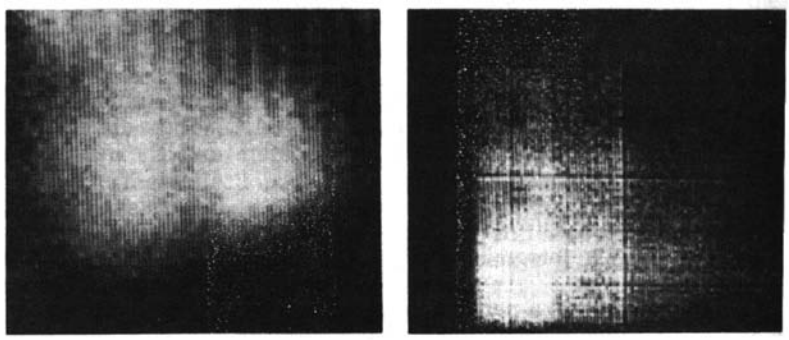

FIG. 18. Effects of differential non-linearity in ADCs: (left) non-linearity in both X and Y ADCs, (right) non-linearity in X ADC only.

2.1.6.6. Count rate performance

The count rate performance of a scintillation camera describes the non-linearity in the relationship between the count rate and the intensity of incident gamma radiation on the crystal surface. Resolution can also degrade from spatial displacements in the image that occur at high count rates.

The complexity of the camera–computer system and its response to a changing count rate make it impossible to quantify the count rate response by a single parameter. Several measurements are required. The intrinsic count rate performance with a decreasing flux of incident gamma radiation caused by a decaying short lived source can be measured with the source positioned so that no scattered radiation reaches the camera. This may be achieved by suspending the radiation source in air, away from material objects (see Fig. 19), or preferably, by using copper absorbers to filter out the scatter component (see Fig. 20).

Similar tests can be performed to quantify the count rate response under differing conditions. For example, to examine the response under clinical conditions, the test can then be repeated with the decaying source in a carefully controlled scattering medium. These measurements can be repeated again as system measurements with a collimator mounted and all peripherals enabled. From the resulting graphs, the count rate for a 20% count loss and the maximum count rate under the respective measurement conditions may be determined. From a practical standpoint, only the first test with no scatter present is usually made, although the additional tests provide results that are more clinically relevant. Alternatively, in non-fully digital systems, the count rate performance may be characterized by determining the pulse pair resolving

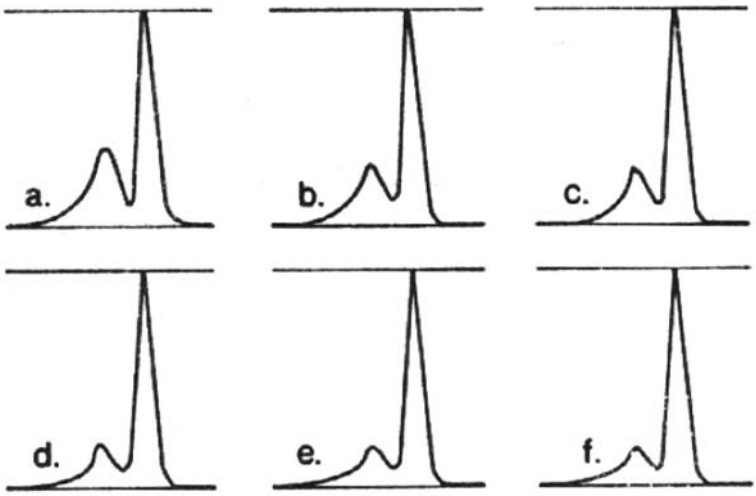

FIG. 19. ^{99m}Tc spectra observed with the multichannel analyser of a scintillation camera: (a) open source on floor, (b) open source on plaster wall, (c) open source 10 cm from plaster wall, (d) source on light foam pad 22 cm above floor and 36 cm from plaster wall, (e) source on light foam pad 22 cm above wood tray table in open doorway, (f) source suspended on tape in open doorway. Significant scatter is evident at all positions.

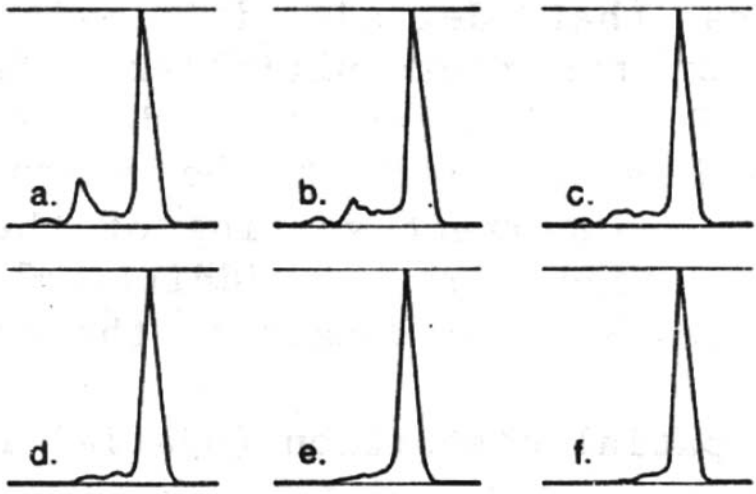

FIG. 20. ^{99m}Tc spectra observed with the multichannel analyser of a scintillation camera with copper absorbers covering source: (a) no absorber, (b) 1 absorber, (c) 2 absorbers, (d) 3 absorbers, (e) 4 absorbers and (f) 5 absorbers. Each copper absorber has a thickness of 1.28 mm. Use of 6 mm or more of filtration by copper produces a clean, scatter free spectrum, which is not altered by additional thickness of copper.

time at a specified count rate. Finally, the spatial resolution and flood field uniformity should be measured at some specified high count rate.

The major factor that degrades count rate performance is a decrease in the ratio of observed to detected events. This may be caused by a narrowed PHA window, collection in a zoom mode, or by an increase in the scattered photon component. In particular, it is important to be aware of the presence of high energy photons in the decay scheme of a radionuclide such as ^{57}Co or ^{123}I. All of these factors will increase the fraction of non-detected events.

Degradation in count rate response can also be caused by extended ADC conversion time, poorly adjusted sample-and-hold circuits and delays due to other data processing that the computer may be performing at the time of data collection. In systems that are able to perform simultaneous acquisition and analysis, or simultaneous acquisition from two or more camera heads, the possibility of degraded count rate response due to delays originating in software, especially at high count rates or high frame rates, must be seriously considered by the user. Newer cameras are designed to process and acquire a larger amount of data more quickly. The speed of each system component is also increased. One mismatched component will slow the entire system.

2.1.6.7. Timing accuracy of data collection

The timing accuracy of data collection describes the capability of the camera–computer system to partition data accurately into the desired temporal segments or frames. Timing accuracy may be quantified by performing a simulated clinical study. This is done first by comparing the apparent frame time, as deduced from the count in each frame, with the requested frame time; and second, by comparing the apparent collection time, as deduced from the sum of the apparent frame times, with the individual frame duration with the requested collection time. Another timing consideration is the capability of the system to perform an ECG gated study properly and to divide the cardiac cycle accurately into the desired number of segments without undesired delays or variations in collection times.

In systems offering simultaneous acquisition and analysis (or simultaneous collection), the major factors that degrade the timing accuracy of data collection in framed dynamic studies are delays in disk response and in originating software response. Two major factors degrade accuracy in ECG gated studies: the first is uncertainty in the timing of signals generated by the ECG and the second is a delay between generation of the gate signal and its receipt by the computer due to intervening electronics, e.g. tape recorders that relay the signal by reading it from the recorded tape rather than passing it directly to the computer. A further important factor for accurate registration of

a volume curve is the synchronization of the initial positive (or negative) slope of the R wave with the response of the computer.

2.1.6.8. Planar sensitivity

Planar sensitivity describes the probability of detecting a photon incident on the detector. Planar sensitivity is conventionally quantified as the count rate per unit of activity for a flat radioactive source of defined diameter at a defined distance from the exposed face of the crystal housing of the uncollimated camera (intrinsic sensitivity), or from the exposed face of the collimator (system sensitivity).

In general, intrinsic sensitivity is directly related to the thickness of the crystal and width of the PHA window and inversely related to the photon energy. System sensitivity is also directly related to the ratio of the crystal area not covered by the collimator septa to the total area of the crystal and inversely related to the collimator thickness.

The major factors that degrade intrinsic planar sensitivity are: count loss due to high count rate; poor alignment of the gains of the photomultipliers; failure of one or more photomultipliers; spatial non-linearity; defects in, or deterioration of, the crystal; physical separation of the photomultiplier–light guide assembly from the crystal and incorrect setting of the position or width of the PHA window. Defects in, or damage to, the collimator are additional factors that degrade system sensitivity.

2.1.6.9. Detector head shielding leakage

Detector head shielding leakage is a measure of the adequacy of the lead shielding incorporated in the detector head. The purpose of this shielding is to eliminate background and other stray radiation.

Detector head shielding leakage is evaluated by measuring the count rate from gamma emitting radiation sources of various energies positioned at different sites around the detector head. Detectors are designed to be used with a given range of photon energies and they should not be used or tested with any energy greater than the specified maximum.

2.1.7. Operational considerations

2.1.7.1. General operating conditions

In view of the complexity of scintillation cameras, special care and attention must be paid to their operation. Adoption of the following practices will help to maintain stable operating conditions.

(1) The high voltage supply to the photomultipliers should be interrupted as little as possible. As a minimum, a dropout relay should be fitted in the electrical supply so that the full operating voltage is not reapplied immediately after an interruption of the electrical supply. Preferably, an uninterruptible power supply should be used, which will continue to supply power for a given length of time. Automatic changeover to a battery source will maintain the high voltage on the PMTs and retain the correction tables, if available.
(2) The temperature and humidity ranges within which a computer will operate are very limited and care must be taken that these are not exceeded. High temperature and humidity can result in expensive failures. As the temperature is lowered by air conditioning, the humidity rises, which may result in the need for a dehumidifier. Manufacturers' specifications should be followed with regard to environmental conditions.
(3) Oscilloscopes, monitors, computers and display devices should be switched off overnight and for longer periods of disuse. If the camera is configured to a network, care must be taken when switching off any component of the system.
(4) When left for long periods, the camera is best left positioned with the crystal face horizontal and directed downward. This helps to prevent separation of the photomultiplier–light guide assembly from the crystal.
(5) A collimator should be attached to the detector head at all times to provide physical and thermal protection of the crystal.
(6) Whenever collimators are changed, the detector head, collimators and collimator mountings should be checked for damage.
(7) To avoid crystal fracture, the room temperature should not be allowed to change rapidly (see Fig. 21). A rule of thumb is that a change in temperature should be gradual and less than 5°C/h.

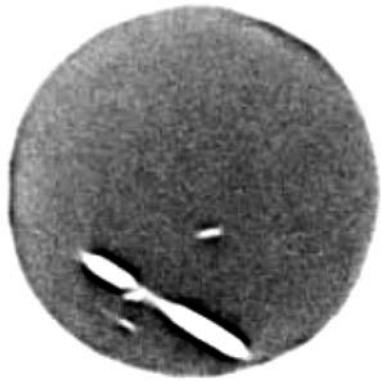

FIG. 21. Cracked crystal. Routine intrinsic uniformity testing revealed various cold areas with hot edges. These cracks are due to thermal changes. The hot edges are due to light reflection at the crack of the crystal (see Ref. [3]).

(8) Radioactive contamination of the collimators, the detector head and the bed should be avoided. It is good practice when setting radioactive materials on the face of the crystal housing, collimator, or bed to always place them on plastic sheeting.
(9) Film cassettes should be kept clean and handled with care to prevent light leakage.
(10) Basic quality control procedures for film processors and laser imagers should be regularly carried out. Film processors should be maintained and cleaned as recommended by the manufacturer's representative. Only fresh chemicals should be used. The chemicals should be replaced at recommended intervals and the temperature should be monitored on a daily basis.
(11) Strict adherence to radiation safety practices should be maintained, especially during the filling of phantoms and when making sources.

For a more detailed discussion and image examples, refer to Sections 2.2.7, 6 and 7 of Ref. [3].

2.1.7.2. Documentation

The complexity and versatility of a camera–computer system make it imperative that adequate documentation be obtained at the time of installation. Documentation for both the hardware and the software should be provided. It is strongly recommended that the buyer obtain sufficient

documentation about the hardware to allow repairs to be made by a competent electronics technician. It may be desirable to obtain two copies of all documentation so that one complete set can be stored in a safe location away from the system itself.

2.1.7.3. Preventive maintenance

The room in which a camera–computer system is installed should be kept scrupulously clean. Even when protected by air filters, computer disks can be destroyed by high levels of dust and smoke. The filters should be cleaned at regular intervals. Where appropriate, manufacturers should also specify regular cleaning of disk packs, disk heads and magnetic tape heads. Such cleaning, although advisable, should be carried out only by properly trained staff. Improper cleaning of disks can be much more damaging than no cleaning at all. More specialized computer maintenance should be performed by a qualified service engineer at regular intervals.

2.1.7.4. Software

The original manufacturer's software disks, or other computer media, must always be safely stored in a secure location away from the computer. These should never be used for the routine operation of the computer. The original software disks should be copied and these backup copies should also be stored in a secure location. The importance of this policy cannot be overemphasized. If the contents of the distribution media are accidentally destroyed in the absence of backup copies, the entire system will be rendered useless in the event of failure until new copies are obtained from the manufacturer. This may entail significant expense in both time and money.

Each system will also have various configuration files, such as digital image correction files, that must be stored as part of the routine system data backups and must be available for use as required. The service engineer should be consulted for advice on system software and data backup.

2.1.7.5. Record keeping

It is essential that a logbook, in written or digital form, be kept with the system at all times. Unexpected events tend to happen and should be recorded in the logbook in as much detail as possible. The user should also try to find out why the unexpected event occurred. Examination of small, seemingly inconsequential failures may allow the prevention of major failures in the future. It is also useful to find out where similar systems are in use (preferably

before the system is purchased), so that when problems arise, other users can be contacted and advice obtained. The events recorded should note both hardware and software failures. Regular backups of a digital logbook must be implemented.

2.1.7.6. Test conditions

Specific test conditions applicable to acceptance, reference and routine testing of a scintillation camera are described and should be followed during all testing procedures:

(1) No electrical or mechanical modifications to the instrument should be made prior to testing.
(2) The PHA should be adjusted before any tests are carried out so that the specified window is used and centred on the appropriate photopeak. If the camera is equipped with automatic energy selection, the pulse height spectrum display should always be checked.
(3) Background radiation levels should be reduced to a minimum by removing extraneous radiation sources, including patients to whom radiopharmaceuticals have been administered. Adjacent rooms should be checked as well.
(4) The count rate in any test, unless otherwise specified, should not exceed 10 000 counts/s in cameras manufactured before 1980 and 30 000 counts/s in newer cameras.
(5) The radionuclide, source configuration, amount of radioactivity, collimator, instrument settings, imaging parameters and test results should be recorded in the instrument logbook and accompanied by the images whenever possible.
(6) A representative of the manufacturer should be present during the acceptance testing procedures. The manufacturer should be contacted before acceptance testing in order to procure any resolution test phantoms that are used only during acceptance testing. These terms should be specified in the contract with the manufacturer.
(7) The camera–computer interface should be adjusted so that the detector's useful field of view (UFOV) without zoom, i.e. the field of view defined by the collimator, is entirely contained in the digital image.
(8) The image orientation should be properly set on the camera head or in the software before commencing the test procedure.

2.1.7.7. Tests to be carried out

This section describes a series of tests to be carried out on a scintillation camera, starting with the acceptance tests. Each test, along with the frequency at which it is to be carried out, is noted in Section 2.2: Test schedule.

Each day of use, simple operational checks are suggested to determine that the entire system is in good working order. These include checks of collimator and detector head mountings, energy calibration of the PHA, flood field uniformity, sensitivity and background count rate, in addition to checks that image and film processing devices are in good working order. The choice of a daily system or intrinsic uniformity check should be made according to local conditions. A quantitative evaluation of a high count flood field uniformity test should be performed at least on a weekly basis. Regular quality control of spatial resolution and spatial linearity should be carried out on a weekly basis.

The manufacturer's quality control test methods for a particular camera system, and recommended amounts of radioactivity to be used, should always be adhered to. This is especially important with respect to quantification of the resulting acquired test data using the manufacturer's software and applying action thresholds for decision making.

The tests described in this section provide a basic evaluation of the camera–computer system relevant to nuclear medicine applications. They do not represent a complete and exhaustive test of the computer.

Tests on multiple head scintillation cameras should be carried out on each individual detector head and on its associated electronics, as appropriate. Additional tests for these systems can be found in Section 5.

2.1.7.8. Radiation sources, test phantoms and other items required

A number of items are required for more than one testing procedure and are described to avoid repetition:

(1) Unsealed radionuclides in solution, e.g. ^{99m}Tc, for point, flood and line sources.
(2) A long lived radionuclide flood source, in the form of an extended sheet of plastic, with photon energy similar to that of the radionuclide most commonly used, e.g. ^{57}Co flood source (122 keV) for ^{99m}Tc (140 keV).
(3) Point source containers, e.g. 1 mL disposable plastic syringes into which the radionuclide solution can be drawn.

(4) A source mounting for a point source in its container, on the central axis of the detector at a distance from its face equal to five times the largest dimension of the UFOV (defined by the lead mask, if available) (see Fig. 22).

(5) A lead mask conforming to the shape of the crystal and at least 3 mm thick, masking the crystal to the dimensions of the UFOV (if possible) (see Fig. 22).

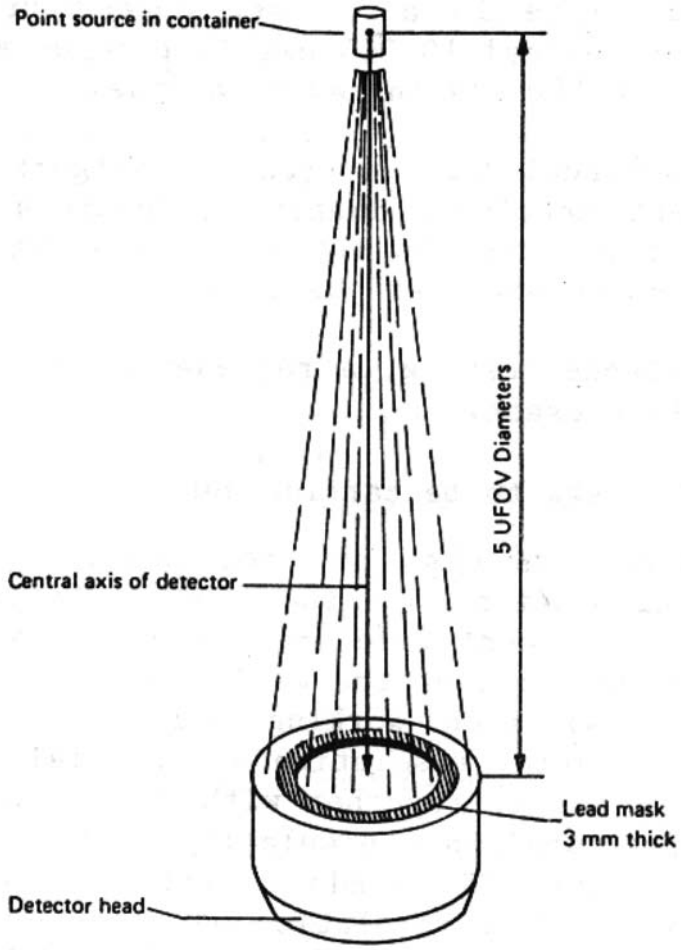

FIG. 22. Mounting of point source in its container on the central axis of the detector at a distance from its face equal to five times the diameter of the UFOV as defined by the lead mask.

(6) An open lead pot at least 4 mm thick and with a well depth of 50 mm.

(7) A flood field uniformity phantom (flood phantom) (see Fig. 23). When filling the flood field uniformity phantom, follow the manufacturer's instructions, or the following suggested instructions. Using a syringe with attached needle, introduce an appropriate volume of radionuclide solution with a given activity concentration. Fill the phantom with water, making sure to leave an air bubble at the top. Mix the contents well. Top off the phantom with water, taking care not to overfill the phantom, which may cause bulging at the centre and thus a non-uniform source. (The total volume of water should be equivalent to the cavity volume of the phantom.) Insert the sealing plugs and tighten. At the same time, check the sealing plugs to be sure they are not leaking. Before emptying, allow the radionuclide to decay (approximately 10 half-lives). Indeed, if the phantom is in regular use, only partial emptying may be necessary between tests. Periodically, however, it should be emptied and washed

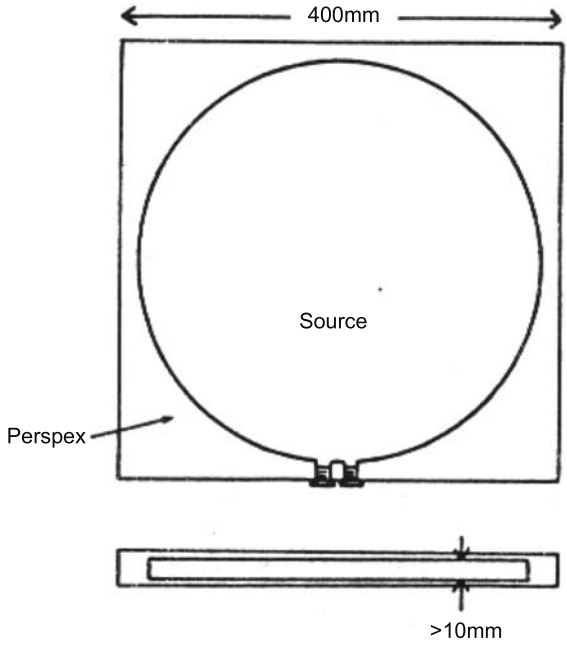

FIG. 23. Flood phantom, fabricated in plastic (e.g. Lucite, Perspex), which provides a uniform flood source when filled with ^{99m}Tc in solution. The diameter of the liquid filled area should be 5 cm greater than the UFOV.

with clean water, then with dilute sodium hypochlorite solution to discourage growth of algae. The phantom must always be stored in a shielded location.

(8) Spatial resolution phantoms of differing design (see Figs 24 and 25). These are transmission phantoms used in conjunction with a point source or flood source.

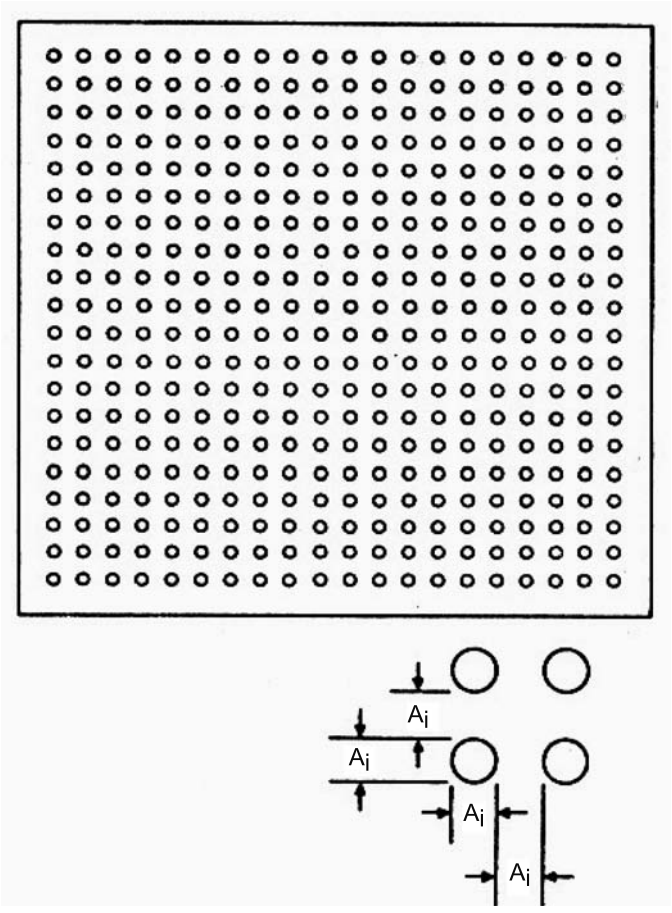

FIG. 24. Orthogonal hole transmission pattern (OHTP) phantom. The phantom consists of a sheet of lead about 3 mm thick with a regular pattern of circular holes, sandwiched between two sheets of plastic. The minimal lead spacings, A_i, are equal to the hole diameters, A_i.

(9) Intrinsic spatial resolution phantoms (see Fig. 26). Lead phantoms with slits of 1 mm width at 30 mm spacings, one phantom with the slits aligned in the X direction and one in the Y direction. (These phantoms are available from each manufacturer with appropriate software to calculate intrinsic resolution in the X and Y directions. Prior to acceptance testing, the manufacturer should be requested to provide the phantoms designed for the particular system to be tested, for the duration of the testing, along with the appropriate software.)

(10) A system spatial resolution phantom (see Fig. 27). Each of 4 capillary tubes with an internal diameter of less than 1 mm is filled with a radionuclide solution (~0.3 GBq/mL (8 mCi/mL) of ^{99m}Tc) at an appropriate activity concentration using a syringe fitted with a small bore needle. After filling, each tube must be plugged with sealing clay or wax at both ends to prevent leakage.

(11) A planar sensitivity phantom. A circular, flat bottomed plastic container 10 cm in diameter and 1 cm deep (e.g. Petri dish) is suitable.

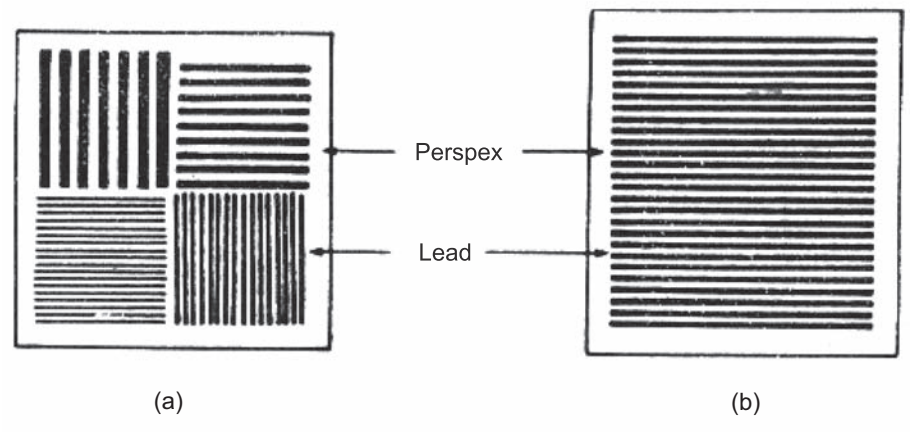

FIG. 25. Spatial resolution phantoms: (a) quadrant bar phantom, (b) parallel line equal spacing (PLES) phantom.

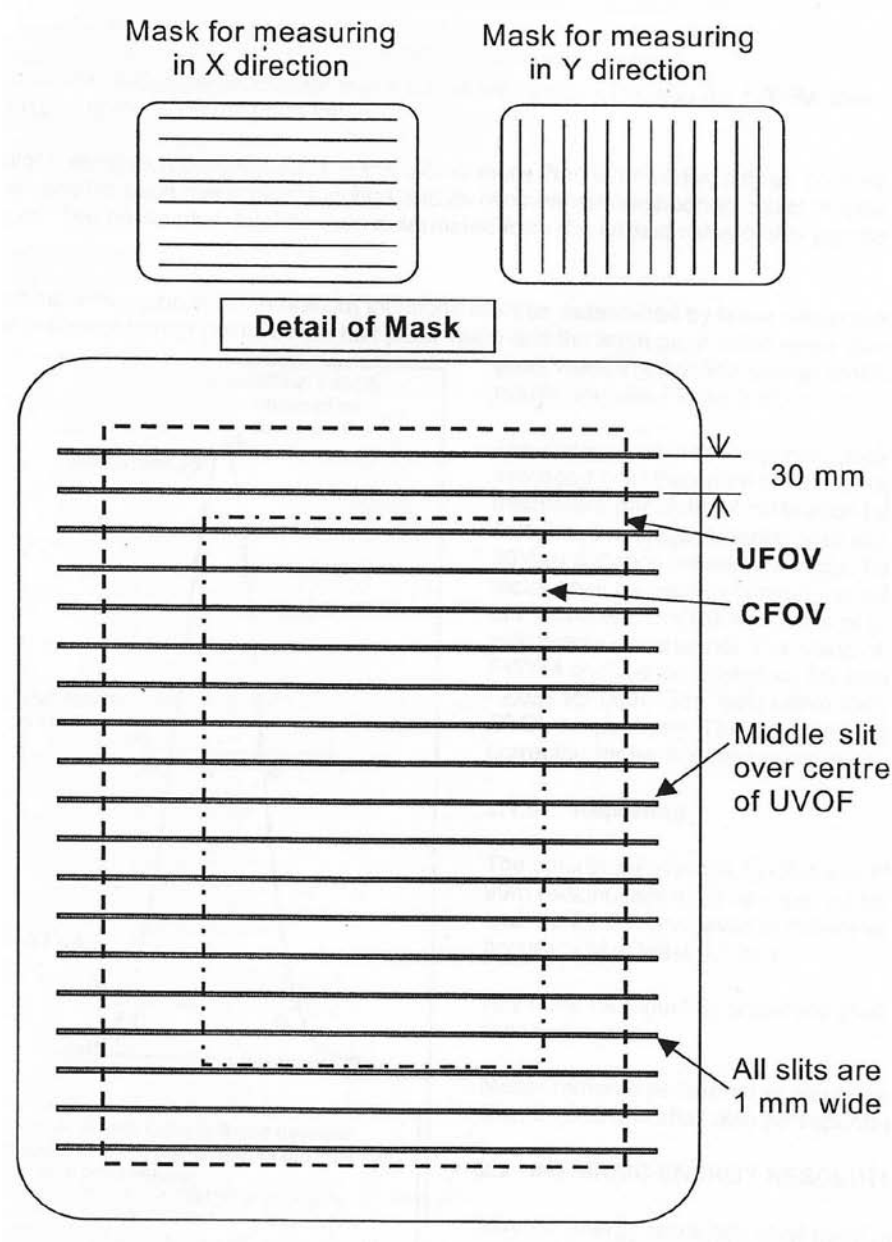

FIG. 26. Intrinsic spatial resolution and linearity phantom. The phantom is constructed of a lead sheet at least 3 mm in thickness with 1 mm slits at 30 mm spacing. The phantom is available from every manufacturer, with prior arrangement, for the duration of the acceptance testing (see Ref. [9]).

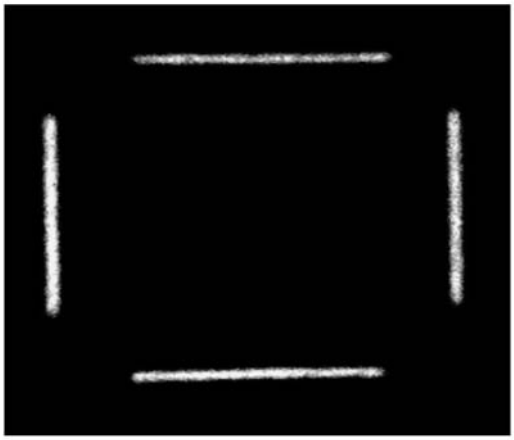

FIG. 27. Image of a system spatial resolution phantom consisting of two sets of parallel line sources filled with ^{99m}Tc in solution. The solution is contained in glass microcapillary tubes, about 0.5 mm in internal diameter, plugged at each end with sealing clay or wax and mounted on the edges of a 10 cm square drawn on a sheet of rigid plastic or heavy cardboard. Note that the resolution in the X and Y directions differ.

2.2. TEST SCHEDULE

Table 2 lists the recommended quality control tests for a scintillation camera, with suggested frequencies for testing. The operational checks should be carried out each day the instrument is used.

2.3. ACCEPTANCE AND REFERENCE TESTS

2.3.1. Physical inspection

Purpose of test

To inspect a scintillation camera, control console, computer, and data storage and display devices for shipping damage and production and design flaws.

TABLE 2. TEST SCHEDULE FOR SCINTILLATION CAMERAS

Section no.	Test	Acceptance	Reference	Weekly	Half-yearly
	Acceptance and reference tests				
2.3.1	Physical inspection	X			
2.3.2	Centring of PHA window settings		X	X	
2.3.3	Intrinsic flood field uniformity for ^{99m}Tc	X	X	X	
2.3.4	Intrinsic flood field uniformity through narrowed and asymmetric PHA windows		X		X
2.3.5	Intrinsic flood field uniformity for radionuclides other than ^{99m}Tc		X		X
2.3.6	Intrinsic spatial resolution: —visual —quantitative	 X X	X		X
2.3.7	System flood field uniformity		X		X
2.3.8	System spatial resolution and linearity	X			
2.3.9	System planar sensitivity	X	X		X
2.3.10	Collimator hole alignment		X		
2.3.11	Intrinsic count rate performance	X	X		X
2.3.11.1	Alternative 1	X	X		X
2.3.11.2	Alternative 2	X	X		X
2.3.11.3	Alternative 3	X	X		X
2.3.11.4	Alternative 4	X	X		X
2.3.12	Basic computer timing		X		X

TABLE 2. TEST SCHEDULE FOR SCINTILLATION CAMERAS (cont.)

Section no.	Test	Acceptance	Reference	Weekly	Half-yearly
2.3.13	Computer timing in dynamic acquisition		X		X
2.3.14	ECG gated acquisition		X		X
2.3.15	Multiple window spatial registration	X	X		X
2.3.16	Detector head shielding leakage	X			
2.3.17.	Intrinsic or system spatial resolution and spatial linearity (routine method)		X	X	
2.3.17.1	Method 1: Flood source method		X	X	
2.3.17.2	Method 2: Point source method		X	X	
2.3.17.3	Method 3: Digital image method		X	X	
	Operational checks (to be performed each day the camera is used)				
2.4.1.	Collimator and detector head mountings and collimator damage				
2.4.2.	Energy calibration of PHA				
2.4.3.	Flood field uniformity and sensitivity				
2.4.4.	Background count rate				
2.4.5	Film handling and processing				

Procedure

(1) Detector housing and support assembly: Inspect the aluminium casing surrounding the NaI(Tl) crystal for signs of indentation or puncture and the gantry for loose parts or mechanical difficulties. Move the gantry, bed and detector head through all possible motions and to the fullest extent of travel, noting any grinding noises, loose parts, inability to move, or improperly functioning controls.

(2) Control panels: Inspect the switches and other controls for loose or broken parts. Check for switches that do not throw securely. Check computer keyboards and accessories for proper operation.
(3) Image display devices: Inspect the display screen for scratches, fingerprints, dust or other debris. Inspect the image monitors for any interference patterns, rolling, lines or other signs of improper operation or electrical interference.
(4) Image recording devices: See Section 7.
(5) Hand control: Inspect the hand control for proper mechanical operation and confirm that the cable has acceptable strain relief at maximum extension.
(6) Emergency devices: Test each emergency button to ensure that all gantry motion ceases when each button is activated.
(7) Mobile cameras: Test the driving mechanism and all emergency stopping devices to ensure proper operation.
(8) Collimators: Inspect the collimators for damage. Load each onto the camera head and ensure that the collimator mounting mechanism is aligned and is working properly. While mounted, activate each motion sensor device to ensure that all gantry motion ceases.
(9) Electrical connections, fuses and cables: Inspect for any loose or broken cable connectors and pinched or damaged cables. Locate all fuses and circuit breakers to enable prompt checking during equipment failure. Ensure that cables are housed, wherever possible, in conduits and are not loose on the floor. Also, ensure that they were placed to allow maximum patient access.
(10) Data storage and display devices: Steps (2)–(4) above are applicable.
(11) Operation and service manuals: Check that all appropriate documentation is available, including performance specifications.

Observations

This test is intended to be performed as an acceptance test. Physical inspection should be carried out immediately on receipt and installation of an instrument. This will allow the supplier to be informed immediately of any damage, deficiencies, or flaws and to arrange for expeditious repair. In the event of major damage, acceptance testing must usually be halted until this is rectified. If only an isolated component (i.e. a collimator) is involved, acceptance testing may proceed after notification of the damage. If performance specifications are not available, they should be requested and obtained from the manufacturer's representative before acceptance testing.

2.3.2. Test of centring of PHA window settings

Purpose of test

To test that all preset PHA windows for clinical imaging are properly centred for every radionuclide to be used with the scintillation camera.

Materials

- Point sources (see Section 2.1.7.8) consisting of the radionuclides concerned, with activities of about 10 MBq (250 μCi), in suitable containers;
- Source mounting for point source (see Section 2.1.7.8);
- Lead mask (see Section 2.1.7.8).

Procedure

If a pulse height spectrum display is available:

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the lead mask centrally on the crystal housing.

For each radionuclide in turn:

(3) Mount the source in the source mounting.
(4) In the acquisition mode, select the default energy setting for the radionuclide concerned, which sets the energy and window width to be used for clinical imaging.
(5) Observe the display to ensure that the respective photopeaks are centred in the window settings. If they are not centred, manually adjust each photopeak so that it is centred. Record the peak value that properly centres the photopeak.
(6) Remove the source.
(7) Repeat steps (3)–(6) for each radionuclide in turn. When finished, remove the lead mask and replace the collimator.

If no pulse height spectrum display is available:

(1) Set the manual settings for the energy of the photopeak.
(2) Set a narrow PHA window (5%, if possible).

(3) Vary the energy setting about the correct energy of the radionuclide tested, taking a count at each setting.
(4) Check that the maximum count occurs at the correct energy setting. If it does not, record the energy at which the maximum count occurred.
(5) Repeat steps (1)–(4) for each radionuclide in turn. When finished, remove the lead mask and replace the collimator.

Observations

This test is intended as a reference test to be performed at the time of acceptance and at weekly intervals. After the completion of the acceptance testing procedure, the PHA windows should be checked daily until the stability of the preset windows is confirmed.

If, at acceptance or anytime thereafter, the preset PHA windows do not correspond to the centre of the appropriate photopeaks, the manufacturer's representative should be immediately notified so that the preset windows can be properly adjusted. Until the adjustment can be done, the energy windows must be set manually. It is possible to have one radionuclide properly centred and another not, thus each must be checked.

Interpretation of results

If any or all preset energy windows appear maladjusted, this would suggest an incorrect energy calibration of the system. If, at acceptance or anytime thereafter, the preset PHA windows do not correspond to the centre of the appropriate photopeaks, the manufacturer's representative should be immediately notified so that the preset windows can be properly adjusted.

If the photopeaks are not properly centred in the preset energy windows, then the centring must be repeated each time the radionuclides are used, until the windows are properly adjusted by the manufacturer's representative.

Limits of acceptability

All photopeaks must be properly centred. If any or all are not centred, corrective action must be taken.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.3. Test of intrinsic flood field uniformity

Purpose of test

To test the intrinsic response of a scintillation camera to a spatially uniform flux of incident photons over the field of view using a symmetric (centred) energy window over the photopeak.

Materials

- Point source (see Section 2.1.7.8) consisting of 10–20 MBq (0.3–0.5 mCi) of ^{99m}Tc in solution in a suitable container mounted in the source holder. The count rate should not be greater than 30 000 counts/s with the manufacturer's default PHA window.
- Source mounting for point source (see Section 2.1.7.8).
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the lead mask centrally on the crystal housing.
(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(5) Acquire an image with a preset count of approximately 3×10^7. Use a matrix size that produces pixel sizes with a linear dimension of 6.4 mm ± 30%. A 64 × 64 matrix size can be used for any camera with the largest dimension of the UFOV under 400 mm. If the UFOV is larger, a 128 × 128 matrix should be used. The chosen preset count should result in about 10 000 counts in the centre pixel.
(6) Obtain a hard copy of the image with the normal contrast settings and then adjust the image contrast so that low contrast defects are visible (a high contrast image). Record the background and contrast settings. Obtain a second hard copy image with the high contrast settings. Store the digital image data for further analysis.
(7) Remove the source and lead mask. Replace the collimator.

Data analysis

Method 1: Visual image

Visually inspect the image for variations in brightness or density. Look carefully at the high contrast image. Note any areas that clearly stand out or any areas indicating an out-of-tune PMT.

Method 2: Digital analysis (NEMA method [7, 9])

(1) The central field of view (CFOV), which is the area defined by scaling all linear dimensions of the UFOV by a factor of 75%, will include any pixel which has at least 50% of its area within the CFOV. The pixels to be included in the uniformity analysis calculations are determined in one sequential pass over the image data with the following two rules. First, set any pixels at the edge of the UFOV that contain less than 75% of the mean counts per pixel to zero. Second, set the values of all CFOV pixels to zero when at least one of the four abutting pixels has a value of zero. (The diagonal neighbours are ignored.)

(2) Smooth the image data resulting from step one in the image processor once. Use a nine point smoothing function with the following pattern of weightings:

1	2	1
2	4	2
1	2	1

The weighting factor for a pixel outside the analysed area in the nine point filter function shall be zero. The smoothed value shall be normalized by dividing by the sum of non-zero weighting factors.

(3) Determine the maximum (max) and minimum (min) counts in pixels lying within the UFOV and the CFOV. The integral uniformity is then given by:

$$\text{integral uniformity} = 100[(\max - \min)/\max + \min)] \quad (2)$$

(4) Determine, for each row or column of pixels in the X and Y directions within the UFOV and the CFOV, the maximum count difference in any five contiguous pixels. This is performed by examining the first set of five pixels, then stepping forward one pixel and analysing the next set. Determine the highest value of this maximum count difference in the sets of rows and columns. The differential uniformity is then given by:

$$\text{differential uniformity} = 100[(\text{high} - \text{low})/(\text{high} + \text{low})] \quad (3)$$

where high and low are the pixel counts giving the highest value of the maximum count difference.

Observations

This test is intended to be performed as an acceptance and reference test and at weekly intervals. If the scintillation camera is fitted with a uniformity correction circuit, the test should, if possible, be performed with and without the circuit enabled.

Most modern scintillation camera–computer systems have installed the appropriate software to perform this digital analysis automatically. It can be accomplished, more laboriously, with a printout of the counts in each pixel of the matrix.

Method 3: Digital analysis

Follow the manufacturer's recommendations.

Interpretation of results

Method 1: Visual image

Compare the image obtained at acceptance testing with the image acquired by the manufacturer at the factory or by his representative at the installation. Compare the images acquired at routine testing with the reference images acquired at acceptance.

Methods 2 and 3: Digital analysis

The values of the NEMA integral and differential uniformities for the UFOV and CFOV obtained at acceptance testing should be compared with the manufacturer's published specifications for the system tested and the worst case values.

The values obtained at routine testing should be compared with the reference values. The uniformity of most scintillation cameras with a uniformity correction circuit but with the circuit disabled will be poorer than with the circuit enabled. This does not represent a malfunction. If possible, uncorrected reference images should be obtained at the time of acceptance, after major repair and if the daily uniformity worsens (see Section 2.4.3).

Scintillation cameras fitted with lookup tables used for uniformity correction, which are acquired on-site, should determine that the corrections were loaded within one week of acceptance testing. For a period of several months after installation, the uniformity of new camera systems may change as the electronics and the phototubes readjust. This may cause the uniformity correction data to become invalid. Thus, for the first few months after installation, the uniformity with corrections applied should be checked each day of use. Appearance of non-uniformities would indicate that the corrections should be recollected.

Limits of acceptability

Method 1: Visual image

Although there are no absolute limits of acceptability, the occurrence of local hot or cold areas that are visible, or visualization of the PMT pattern, is not acceptable. At acceptance testing, if the image obtained appears to differ from that obtained at the factory or at installation, corrective action should be initiated. First, the uniformity corrections should be reacquired by the manufacturer's representative. If this fails to correct the problem, further action should be initiated by the manufacturer's representative. In any case, the image should appear uniform and PMTs should not be evident. The image should reflect the correct shape of the detector. The edges should not be jagged.

At routine testing, the image should be comparable to the reference image. The uniformity must be adequate for clinical imaging. Evident non-uniformities would call for follow-up action.

Methods 2 and 3: Digital analysis

A value of the NEMA integral or differential uniformity obtained at acceptance that is above the manufacturer's worst case value would call for corrective action initiated through the manufacturer's representative.

Action levels should be established at the time of acceptance testing. The clinical procedures to be performed with the system (e.g. planar only, whole body or quantitative SPECT) will determine the stringency of the action levels. If these actions levels are exceeded at routine testing, follow-up action should be initiated. The first step would be to reacquire correction field flood data, if appropriate.

Conclusion

Record all results in order to detect a worsening trend in the values. Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.4. Test of intrinsic flood field uniformity through narrowed and asymmetric (off-centred) PHA windows

Purpose of test

To test the intrinsic flood field response of a scintillation camera through a range of narrowed and asymmetric (off-centred) PHA windows.

Materials

- Point source (see Section 2.1.7.8) consisting of 10–20 MBq (0.3–0.5 mCi) of ^{99m}Tc in solution in a suitable container mounted in the source holder. The count rate should not be greater than 30 000 counts/s with the manufacturer's default PHA window.
- Source mounting for point source (see Section 2.1.7.8).
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the lead mask centrally on the crystal housing.
(3) Mount the source in the source mounting.

(4) Centre a 15% PHA window on the photopeak (see Section 2.4.2).
(5) Acquire an image on the display device using a preset count of 1.5×10^7. Make a hard copy of the image.
(6) Repeat steps (4) and (5) for a window width of 10%, checking that the window remains centred on the photopeak.
(7) Set a 10% PHA window asymmetrically over the lower half of the photopeak (126–140 keV). This is the asymmetric low energy window. This window should acquire counts only from the lower half of the photopeak. Repeat step (5).
(8) Set a 10% PHA window asymmetrically over the upper half of the photopeak (140–154 keV). This is the asymmetric high energy window. This window should acquire counts only from the upper half of the photopeak. Repeat step (5).
(9) Remove the source and lead mask. Replace the collimator.

Data analysis

Visually compare the images, noting particularly any increased non-uniformities at the narrower PHA window. Obtain the quantitative uniformity values as described in the test in Section 2.3.3: Test of intrinsic flood field uniformity, for each of the images collected with a centred PHA window. Record the values only for reference. In general, careful inspection of the images is sufficient, unless clinical studies are to be performed with narrow PHA windows.

Observations

This test is intended to be performed as a reference test at the time of acceptance and at half-yearly intervals, or if a uniformity problem is suspected.

Interpretation of results

A scintillation camera should maintain its intrinsic flood field uniformity at a 15% PHA window. The uniformity may degrade with a 10% PHA window in a properly functioning camera. However, the non-uniformities should be consistent over the field of view. It is not unusual to see the pattern of the PMTs. The images acquired through the asymmetric windows will also show PMTs. If one or more areas stand out, it is indicative of poorly tuned PMTs or lack of integrity of the optical coupling between the photomultiplier–light guide assembly and the NaI(Tl) crystal. In either case, the detector should be checked by the manufacturer's representative.

For a further discussion of this test and further examples, refer to Sections 2.2.1 and 2.2.2 of Ref. [3].

Conclusion

The results cannot be compared with the manufacturer's specifications unless the manufacturer quotes the values for the PHA window used. Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.5. Test of intrinsic flood field uniformity for radionuclides other than ^{99m}Tc

Purpose of test

To test the intrinsic flood field response of a scintillation camera for all other radionuclides used for clinical imaging.

Materials

- Point source (see Section 2.1.7.8) consisting of 10–20 MBq (0.3–0.5 mCi) of the radionuclides used for clinical imaging, in solution in a suitable container mounted in the source holder. The count rate should not be greater than 30 000 counts/s with the appropriate clinically used PHA window(s).
- Source mounting for point source (see Section 2.1.7.8).
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the lead mask centrally on the crystal housing.

For each radionuclide in turn:

(3) Mount the source in the source mounting.
(4) Centre the clinically used PHA window(s) on the respective photopeak(s) (see Section 2.4.2).

(5) Acquire an image with a preset count of approximately 3×10^7. Use a matrix size that produces pixel sizes with a linear dimension of 6.4 mm ± 30%. A 64 × 64 matrix size can be used for any camera with the largest dimension of the UFOV under 400 mm. If the UFOV is larger, a 128 × 128 matrix should be used. The chosen preset count should result in about 10 000 counts in the centre pixel.
(6) Obtain a hard copy of the image with the normal contrast settings, then adjust the image contrast so that low contrast defects are visible (a high contrast image). Record the background and contrast settings. Obtain a second hard copy image with the high contrast settings. Store the digital image data for further analysis.
(7) Remove the source.
(8) Repeat steps (3)–(7) for each radionuclide in turn. Then remove the lead mask and replace the collimator.

Data analysis

Visually compare the images, noting particularly any increased variations in brightness or density with different clinically used radionuclides.

Obtain the quantitative uniformity values as described in the test in Section 2.3.3: Test of intrinsic flood field uniformity. Record the values for reference.

Observations

This test is intended to be performed as a reference test at the time of acceptance and at half-yearly intervals. If a radionuclide other than ^{99m}Tc is used often, this test should be performed more frequently with that radionuclide.

If it is possible to load flood field correction data for each of the radionuclides, the test should, if possible, be performed with and without the appropriate corrections enabled. The correction data should be loaded just prior to acceptance testing.

The uniformity indices for the tested radionuclides should not be significantly higher than the manufacturer's specifications for ^{99m}Tc. The PMTs should not be visible in any of the flood field images.

This test is essential if the scintillation camera has a uniformity correction circuit that uses a single stored reference flood field image to derive the correction matrix for images at all photon energies (see Fig. 28).

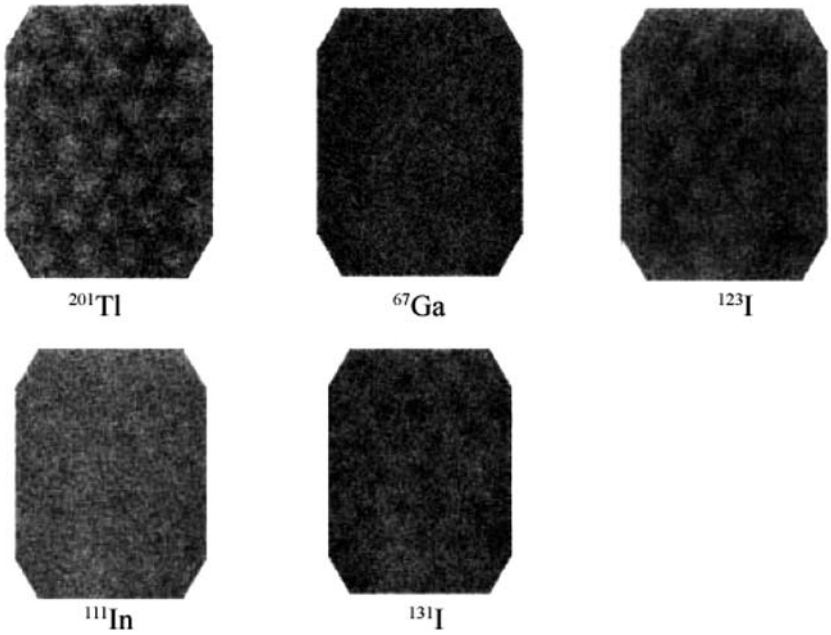

FIG. 28. Comparison of intrinsic uniformity for ^{201}Tl, ^{67}Ga, ^{123}I, ^{111}In and ^{131}I on the same scintillation camera, using a ^{99m}Tc uniformity correction map and using the preset energy windows for each radionuclide. The images show loss of uniformity, especially at low photon energies (^{201}Tl, ^{67}Ga). The uniformity obtained at different photon energies without a radionuclide specific uniformity correction is unacceptable. For this scintillation camera system, an energy specific uniformity correction is needed for each radionuclide used (see Ref. [3]).

For a further discussion of this test and additional examples, refer to Section 2.2.3 of Ref. [3].

Interpretation of results

A scintillation camera should maintain its intrinsic flood field uniformity for all radionuclides used for clinical studies. If not, corrective action should be initiated.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.6. Test of intrinsic spatial resolution

Purpose of test

To test the intrinsic spatial resolution of a scintillation camera in terms of the FWHM of its line spread function.

Method 1: Visual image method

To be used if special phantoms and software are not available.

Materials

- Point source (see Section 2.1.7.8) consisting of 20–40 MBq (0.5–1 mCi) of ^{99m}Tc in solution in a suitable container.
- Quadrant bar phantom with bar widths of approximately 2, 3, 3.5 and 4 mm (see Section 2.1.7.8).
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head.
(2) Position the lead mask centrally on the crystal housing.
(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(5) Position the quadrant bar phantom so that it is supported on the detector head housing and as close to the crystal housing as possible. The bars should be carefully aligned with the X and Y axes of the detector face.
(6) Acquire an image at a preset count of 60 000. Use the largest matrix size available. Make a hard copy for reference.
(7) Rotate the quadrant bar phantom through 90º and repeat step (6). Repeat this process two additional times. Invert the phantom and acquire a similar set of four images so that the smallest bars are imaged in each quadrant in each direction. A total of 8 images are to be acquired.
(8) Remove the source, quadrant bar phantom and lead mask. Replace the collimator.
(9) Accurately measure the widths, *B*, of the bars in the quadrant bar phantom.

Data analysis

(1) Determine the widths of the smallest bars that the scintillation camera can resolve in the X and Y directions. This can be done by visual inspection of the images. Note any areas of poor spatial resolution, which may correspond to the location of the PMTs or the edge of the field of view.
(2) Estimate the intrinsic spatial resolutions in the X and Y directions in terms of the FWHM of the line spread function, using the relationship:

$$\text{FWHM} = 1.75B \tag{4}$$

where B is the width of the smallest bars that the camera can resolve.
(3) Average the values in the X and Y directions.

Observations

This test is intended to be performed as an acceptance and reference test at half-yearly intervals. The quadrant bar phantom must be matched to the spatial resolution of the scintillation camera so that at least one set of bars is not fully resolved. The increments of bar width from one quadrant to the next should be small so that the spatial resolution can be estimated with reasonable accuracy. It is recommended that the bar pattern be included with the purchase of the scintillation camera.

Interpretation of results

Compare the estimated values for FWHM in the X and Y directions with the manufacturer's worst case values, at acceptance testing. Compare the estimated values with the reference values, at routine testing.

For further discussion of this test and image examples, refer to Section 2.2.3.1 of Ref. [3].

Limits of acceptability

An acceptance test producing a value of the FWHM that is 20% or more above the manufacturer's worst case value would call for corrective action to be initiated through the manufacturer's representative.

Follow-up action should be initiated following routine testing if the average value of the FWHM is 10% or more above the reference value, or if areas within the UFOV show significant worsening of the spatial resolution.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

Method 2: Quantitative method

This test can only be carried out if the manufacturer provides the appropriate test patterns (see Fig. 26) and the analysis software and the representative is present at the time of testing. This test is to be performed at acceptance.

Materials

- Intrinsic spatial resolution phantoms (see Fig. 26) supplied by the manufacturer.
- Point source (see Section 2.1.7.8) containing about 40 MBq (1 mCi) of ^{99m}Tc in solution in a suitable container.

Procedure

(1) Remove the collimator from the detector head.
(2) Carefully position the first spatial resolution phantom on the detector head. The slits will be parallel to the X or Y direction according to the instructions provided by the manufacturer.
(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(5) Acquire a digital image using the parameters for total count and matrix size as instructed by the manufacturer.
(6) Remove the first spatial resolution phantom and carefully position the second spatial resolution phantom with slits parallel to the other direction. Follow exactly the instructions provided by the manufacturer.
(7) Acquire a second digital image using the parameters for total count and matrix size as instructed by the manufacturer.
(8) Remove the source and the resolution phantom. Replace the collimator.

Data analysis

Using the special analysis software provided by the manufacturer, proceed with the analysis instructions to obtain the intrinsic resolution for the FWHM and the full width at tenth maximum (FWTM) in the UFOV and CFOV. This should be done in both the X and Y directions.

Observations

This test is intended to be performed only as an acceptance test.

Interpretation of results

The values of FWHM and FWTM, averaged over the X and Y directions, in both the UFOV and CFOV, should be compared to the manufacturer's specified worst case values.

Limits of acceptability

A value of FWHM or FWTM measured at acceptance testing, averaged over the X and Y directions in both the UFOV and CFOV, that is greater than the manufacturer's worst case value would call for corrective action to be initiated through the manufacturer's representative.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.7. Test of system flood field uniformity

Purpose of test

To test the system flood field response of a scintillation camera for all multihole collimators used.

Materials

- Flood phantom (see Section 2.1.7.8) containing 200–400 MBq (5–10 mCi) of ^{99m}Tc in solution; or
- ^{57}Co flood source of similar activity.

Procedure

(1) Mount the collimator to be tested on the detector head. Turn the head to face vertically upward.
(2) Place the flood phantom or flood source at a distance of about 10 cm above the collimator face.
(3) Centre the appropriate manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Acquire an image on the display device on hard copy at a preset level of 30 000 counts and a matrix size of 512 × 512. If a flood field uniformity correction circuit is available, it should be switched off, unless the correction data have been collected intrinsically, in which case it should be left enabled.
(5) Remove the flood phantom or flood source.
(6) Repeat steps (1)–(4) for all multihole collimators used.

Data analysis

Visually inspect the images, noting any increased variations in brightness or density not apparent in the corresponding intrinsic flood field image acquired in the test in Section 2.3.3: Test of intrinsic flood field uniformity. Collimator defects may appear as linear artefacts covering large or small areas of the image.

Observations

This test is intended to be performed as a reference test at the time of acceptance and at half-yearly intervals, or if damage to a collimator is suspected. It is important to perform this test to ensure that the flood field response remains uniform for all collimators used. Low energy collimators in particular may be damaged during shipment. This occurs because the thin lead septa can separate if subjected to an impact. Separation of the septa will appear on the images as parallel lines of increased count density (see Fig. 29). If an object has dented or scraped the collimator face, an area of reduced count density will be seen where the septa have been bent. In a high contrast image, sections may have linear artefacts. If the count densities in regions of interest of equal size, placed over areas of varying intensity, differ by more than 2%, the collimator should be rejected.

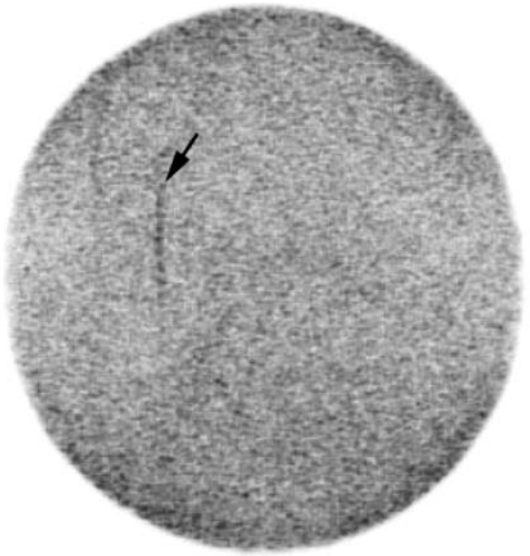

FIG. 29. Test of system flood field uniformity. ^{99m}Tc, 15% energy window, 30 million counts, low energy high resolution collimator. The image shows a discrete line (black arrow) of higher counts indicating that the collimator septa are separated. The collimator requires replacement. This type of problem is likely to occur with a foil collimator and can originate during shipment of a new camera (see Ref. [3]).

If a cone beam collimator is included, the image should be symmetrical in appearance.

If a flood phantom is used, check that the contents are thoroughly mixed to provide a uniform source. If poor mixing is suspected, the phantom should be rotated through 90º and a new image acquired. Poor mixing is confirmed if the non-uniform features move with the phantom.

It may be noted that some uniformity correction circuits require a reference flood image to be acquired for each collimator. For this test, this correction must be turned off so as not to mask the collimator defects.

Interpretation of results

A scintillation camera should maintain its system flood field uniformity for all multihole collimators used. If any significant variations in uniformity not apparent in the intrinsic flood field image are observed, a replacement collimator should be obtained from the manufacturer's representative as soon as possible.

For a further discussion of this test and additional image examples, refer to Sections 2.2.9.4 to 2.2.9.8 of Ref. [3].

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.8. Test of system spatial resolution and spatial linearity

Purpose of test

To test the system spatial resolution of a scintillation camera in terms of the FWHM of its line spread function. This test should be performed for each parallel hole, low energy collimator.

Method 1: Visual image method

The method given here is a supplement to the digital method. It is used to provide reference images for future quality control using a quadrant bar phantom. The method also checks the system spatial resolution over the entire field of view. Therefore, both methods should be performed.

Materials

- Flood phantom (see Section 2.1.7.8) containing about 200–400 MBq (5–10 mCi) of ^{99m}Tc; or
- ^{57}Co flood source of similar activity;
- Quadrant bar phantom (see Section 2.1.7.8) with bar widths and bar spacings of about 4, 6, 8 and 10 mm.

Procedure

(1) Mount the collimator to be tested on the detector head. Turn the head to face vertically upward.
(2) Position the quadrant bar phantom on the face of the collimator. Carefully align the bars with the X and Y axes of the detector face.
(3) Place the flood phantom or flood source on the quadrant bar phantom.
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).

(5) Acquire an image on the display device on hard copy at a preset level of 6×10^6 counts. Use the maximum matrix size available.
(6) Rotate the quadrant bar phantom through 90° and repeat step (5). Repeat this process two additional times, then invert the phantom and acquire a similar set of four images so that the smallest bars are imaged in each quadrant in each direction. This will require 8 images.
(7) Repeat steps (2)–(6), but with the quadrant bar phantom at a distance of 10 cm from the face of the collimator. The phantom can be supported on styrofoam cups or other non-scattering material.
(8) Repeat steps (2)–(6), but with the quadrant bar phantom at a distance of 10 cm from the face of the collimator and with a tissue equivalent scattering medium between the phantom and the collimator.
(9) Repeat steps (1)–(8) for all available parallel hole, low energy collimators.
(10) Remove the flood phantom or flood source and the quadrant bar phantom.
(11) Accurately measure the widths, B, of the bars in the quadrant bar phantom.

Data analysis

(1) Determine the widths of the smallest bars that the scintillation camera can resolve in the X and Y directions with the phantom at the face of the collimator, at a distance of 10 cm from the face in air and at a distance of 10 cm in a scattering medium. This can be done by visual inspection of the images.
(2) Estimate the system spatial resolution in the X and Y directions in terms of the FWHM of the line spread function using the relationship in Eq. (4):

$$\text{FWHM} = 1.75B \tag{4}$$

where B is the width of the smallest bars that the camera can resolve. Repeat this procedure for each of the imaging conditions.
(3) Average the values in the X and Y directions.
(4) Note also whether the lines are straight, indicating good spatial linearity.

Observations

This test is intended to be performed as an acceptance test for collimators. The quadrant bar phantom must be matched to the spatial resolution of the scintillation camera, so that at least one set of bars is not resolved. Such a phantom can be made locally from lead sheeting at least 3 mm thick.

A tissue equivalent scattering medium can be fashioned from layers of plastic (e.g. Perspex, Lucite) or chipboard. A plastic or wooden box filled with uncooked rice may also be used.

Interpretation of results

The estimated values of FWHM for each collimator, averaged over the X and Y directions, at a distance of 10 cm from the face of the collimator in air and at a distance of 10 cm in a scattering medium should be compared with the manufacturer's worst case values.

If the lines are straight at the surface of the collimator but are wavy at 10 cm, this indicates that the septa in the collimator are not parallel, which will cause a loss of resolution and contrast in clinical images (see Section 2.2.3.1 of Ref. [3]).

Limits of acceptability

If a value for the FWHM is obtained that is 20% or more above the manufacturer's worst case value for the collimator in question, the collimator should be replaced. Initiate action, as soon as possible, through the manufacturer's representative with a view to its replacement.

If the scintillation camera has a linearity correction and the lines appear wavy, the manufacturer's representative should be contacted in order to reacquire the linearity correction map. If there is no linearity correction available, the lines may be slightly wavy and no action need be taken.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

Method 2: Quantitative method

This method is the method of choice for accurate assessment of the FWHM and FWTM values for the system spatial resolution.

Materials

- System spatial resolution phantom (see Section 2.1.7.8) containing about 55–80 MBq (1.5–2 mCi) of ^{99m}Tc in solution in each of the four line sources.
- Linear graph paper.

Procedure

(1) Mount the collimator to be tested on the detector head. Turn the head to face vertically upward.
(2) Position the system spatial resolution phantom on the face of the collimator. With the line sources in the centre of the detector, align them carefully parallel to the X and Y axes.
(3) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Acquire a digital image at a preset level of 6×10^6 counts in a 1024×1024 matrix, a 512×512 matrix with a zoom of 1.5, or a 256×256 matrix with a zoom of 2 or higher. For this acquisition, use the highest matrix size allowed by the 'profile' software program in the computer.
(5) Repeat steps (2)–(4), but with the system spatial resolution phantom at a distance of 10 cm from the face of the collimator in air.
(6) Repeat steps (2)–(4), but with the system spatial resolution phantom at a distance of 10 cm from the face of the collimator and with a tissue equivalent scattering medium between the phantom and the collimator.
(7) Repeat steps (1)–(6) for all available low energy multihole collimators.
(8) Remove the system spatial resolution phantom.

Data analysis

(1) Obtain a printout of counts in successive pixels in a narrow section perpendicular to the pair of vertical lines in the digital image to obtain the FWHM and FWTM values in the X direction. The profile program in the image analysis software package can be used. The section may be up to 8 pixel elements broad.
(2) Plot the data as a profile of total count per pixel number against pixel number on linear graph paper. Draw a smooth curve through the data points, being certain both peaks are included.
(3) Determine the separation, S, of the peaks in pixels.

(4) For each peak, calculate the FWHM, *W*, in number of pixels by linear interpolation between adjacent pixels surrounding the half maximum value. Use the highest pixel count in the peak as the maximum.
(5) Calculate the FWHM of each peak in millimetres as

$$\text{FWHM} = \frac{WD}{S} \tag{5}$$

where *D* is the distance between the line sources in millimetres.
(6) Average the FWHM values for the two peaks.
(7) Repeat steps (1)–(6) for 3 or 4 additional sections chosen at different positions along the line. Average all the FWHM values.
(8) Similarly, determine the FWTM by repeating steps (1)–(7) but at one tenth the maximum count.
(9) Repeat steps (1)–(8) for profiles perpendicular to the pair of horizontal lines in the digital image to obtain the FWHM and FWTM values in the Y direction.
(10) Repeat steps (1)–(9) for the images with the phantom at a distance of 10 cm from the face of the collimator in air and at a distance of 10 cm in a scattering medium.
(11) Repeat steps (1)–(10) for all available low energy multihole collimators.

Observations

This test is intended to be performed as an acceptance test for collimators. To increase the counts in the profile, it is possible to take a section more than 8 pixels wide. If this is done, however, care must be taken to align the sources accurately so that the images of the lines lie exactly parallel to the X and Y axes of the image matrix. If not, broadening of the peaks will occur.

The test can be performed on medium and high energy collimators at a distance of 10 cm by taking a much wider section across the images of the lines to average out the effect of the collimator hole size (see Section 2.3.2.3 of Ref. [3]).

Ideally, the matrix size and zoom should be chosen so that there are 10 sampling points within the FWHM. The maximum pixel value in the profile should be 10 000 counts.

Interpretation of results

The calculated values of the FWHM and FWTM for each collimator, averaged over the X and Y directions, at a distance of 10 cm from the face of the collimator in air and at a distance of 10 cm in a scattering medium should be compared with the manufacturer's worst case values.

The values in the X and Y directions as determined for a given collimator should not differ by more than 5%.

Limits of acceptability

If a value of FWHM or FWTM is obtained that is 10% or more above the manufacturer's worst case value for the collimator in question, action should be initiated through the manufacturer's representative with a view to its replacement.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.9. Test of system planar sensitivity

Purpose of test

To test the count rate response of a scintillation camera to a radionuclide source of known radioactivity.

Materials

Planar sensitivity phantom (see Section 2.1.7.8) containing an accurately known amount of radioactivity, about 40 MBq (1 mCi) of ^{99m}Tc or other radionuclide such as ^{131}I, in solution. The radioactivity is determined by measuring the syringe containing the radionuclide solution to be transferred to the phantom in a radionuclide (dose) calibrator. After transferring the radioactivity to the phantom, again measure the empty syringe in the radionuclide (dose) calibrator. This will determine the residual activity in the syringe after the transfer. By subtracting the latter from the former, the radioactivity in the phantom can be determined. The exact time of day corresponding to the activity determination is also recorded. A separate phantom is required for each radionuclide used.

Procedure

(1) Mount a low energy, parallel hole collimator on the detector head. Turn the head to face vertically upward.
(2) Cover the face of the collimator with a plastic sheet. Place the phantom containing the ^{99m}Tc 10 cm from the surface of the covered collimator face.
(3) Centre the manufacturer's default PHA window on the photopeak, or the window width used by the manufacturer in determining the specified performance values (see Section 2.4.2).
(4) Collect an image over a total time of 100 s. Record the total counts in the image frame and the exact time of day.
(5) Remove the phantom and count the background for the same time period. Record the total counts in the image frame.
(6) Repeat steps (1)–(5) for all other low energy multihole collimators.
(7) Repeat steps (1)–(5) with the phantom containing ^{67}Ga or ^{111}In for medium energy multihole collimators and ^{131}I for high energy multihole collimators.

Data analysis

(1) Express all data as net count rates, i.e. corrected for background.
(2) Correct the data for each low energy collimator for the decay of ^{99m}Tc from the time of the measurement of the syringe to the time of the phantom count. (A decay correction is not necessary for ^{67}Ga, ^{111}In or ^{131}I.)
(3) Calculate the planar sensitivity of each collimator in counts per second per becquerel, or equivalent units to match the manufacturer's specifications.

Observations

This test is intended to be performed as an acceptance and reference test for collimators and at half-yearly intervals. The accuracy of the results is clearly limited by the accuracy with which the activity of the radionuclide can be determined. This in turn depends on the accuracy of the radionuclide calibrator used. Accuracy within 5% is sufficient to indicate whether the sensitivities of different collimators are comparable to the manufacturer's specifications.

Even if the activity cannot be determined accurately, sensitivities may still be evaluated relative to that of a selected collimator. Manufacturer's specifications are commonly stated in terms of relative sensitivities, with the

exception of one collimator for which the absolute sensitivity is given. The test is instructive in illustrating the wide variation in imaging times that will be required to attain a given count using different collimators.

This test is also useful as a reference test to determine the consistency over time of the count rate response of the camera and therefore should be performed on a half-yearly basis. This is especially important for multihead SPECT systems.

Interpretation of results

The sensitivity value for each collimator should be compared with the manufacturer's worst case value, allowance being made for the accuracy with which the activity can be determined. For acceptance testing, the radionuclide and energy window used by the manufacturer to determine the specified values must be the same as those used in this test.

Limits of acceptability

If a sensitivity value is obtained that is 10% or more below the manufacturer's worst case value for the collimator in question, the collimator should be checked for damage and action initiated through the manufacturer's representative with a view to its replacement.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.10. Test of collimator hole angulation

Purpose of test

To test the septal alignment and angulation for all parallel hole collimators used.

Materials

Point source (see Section 2.1.7.8) consisting of 200–320 MBq (5–8 mCi) of ^{99m}Tc in solution in a suitable container.

Procedure

(1) Mount the collimator to be tested on the detector head. Turn the head to face the most distant wall.
(2) Place the point source in a source holder on the wall or other location so that it is at least 2.5 m from the face of the collimator. Properly align the source with the centre of the collimator.
(3) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Acquire an image on the display device on hard copy, at a preset level of 5×10^6 counts and with the largest matrix size available.
(5) Remove the point source.
(6) Repeat steps (2)–(5) with the source imaged at four additional locations on the collimator. Each location should be approximately halfway to the edge of the field of view.
(7) Repeat steps (1)–(6) for each of the parallel hole collimators.

Data analysis

Visually inspect each image separately displayed with the maximum matrix size, noting the shape of the point source as imaged. The image in the centre of the collimator should be basically round in appearance. The images should be symmetrical. Hole angulation or collimator defects may appear either as streaks or as a severely distorted shape.

Observations

This test is to be performed as a reference test at the time of acceptance, or if damage to a collimator is suspected. It is possible that only one of the five images from a given collimator will appear abnormal. This occurs as some foil collimators are constructed in layers. It is possible for the holes in one layer to be non-parallel with the holes in the rest of the collimator (see Fig. 30).

Interpretation of results

There are no manufacturer's specifications for this test. Because the shape of the images may vary from collimator to collimator, some judgement on the part of the observer is necessary.

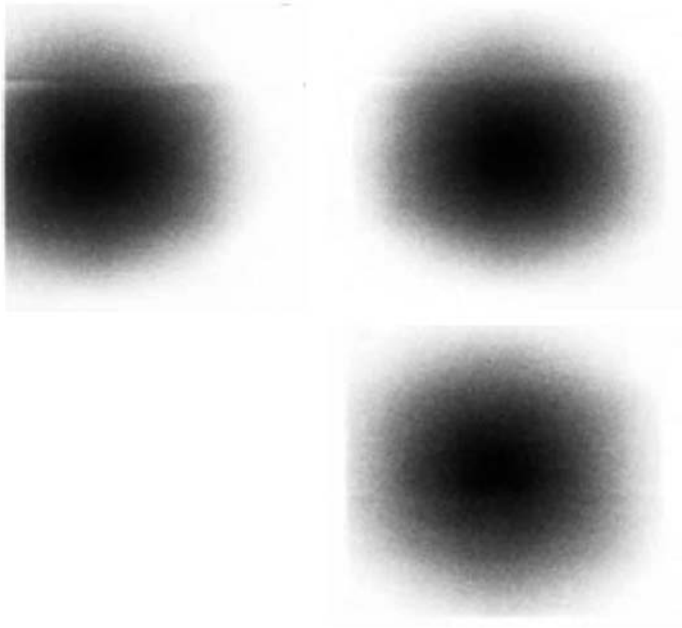

FIG. 30. Collimator septa and hole alignment assessed by a distant point source. The lines of discontinuity seen in the top two images indicate misalignment of the collimator holes. This collimator is unacceptable and must be replaced (see Ref. [3]).

For a further discussion of this test see Section 2.2.9, and for additional image examples see Sections 2.2.9.2 and 2.2.9.3 of Ref. [3].

Limits of acceptability

If the collimator holes do not appear to be parallel or are otherwise not aligned, action should be initiated through the manufacturer's representative to replace the collimator.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.11. Test of intrinsic count rate performance

The purpose of this test is to test the intrinsic count rate performance of a scintillation camera in terms of its response to an increasing flux of incident gamma radiation. This section describes four methods for testing the count rate performance. Newer digital scintillation cameras should be tested using alternative 1. Older instruments can be tested using alternative 2 or 3. If unsure of the method to use, refer to the manufacturer's performance specifications for the instrument to be tested. All cameras should be tested to determine the maximum count rate (alternative 4).

2.3.11.1. Intrinsic count rate performance — the decaying source method (alternative 1)

This is the preferred method of testing intrinsic count rate performance. However, it takes about two days to complete and the scintillation camera cannot be used for anything else. Only one head of a multiple detector camera can be tested at a time.

Materials

- Place a radiation source consisting of ^{99m}Tc in solution contained in a small vial in an open lead pot at least 4 mm thick and with a well depth of 50 mm. Cover the top with 6 mm of copper. The initial activity should be approximately 80 MBq (2 mCi). Adjust the activity so that the input count rate is larger than the count rate required to cause fold over in the observed count rate.
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Turn the head to face vertically downward.
(2) Position the lead mask centrally on the crystal housing.
(3) Position the source on the central axis of the detector at a distance of about 1.5 m from its face (see Fig. 31). The collimated cone of radiation must extend fully across the smaller dimension of the UFOV and care must be taken to minimize scatter.

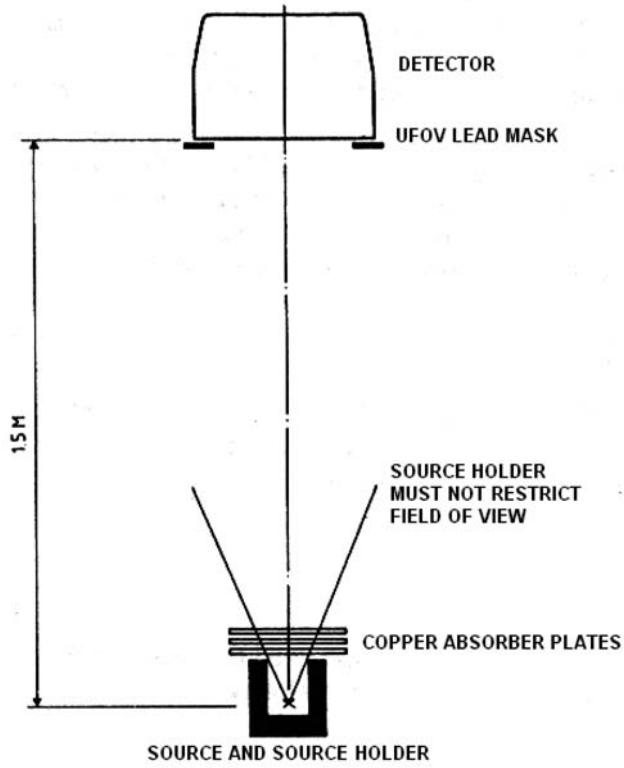

FIG. 31. Test of intrinsic count rate performance (alternatives 1 and 2). Positioning of radiation source in relation to detector.

(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2) using a low count rate. Do not manually readjust the PHA window during the test. If possible, place the camera in a 'normal' counting mode (not a high count rate mode).

(5) Remove the source. Count the background for a preset time of 10 min. Record the background count rate, Bg. Replace the source.

(6) Set up the initial data acquisition time for 10 s. For each acquisition, record the start time (t_i) and elapse time (t_{ei}), where (i) is the number of the data point. Measure the time relative to the start of each acquisition to the time of the measurement of the first data point. Collect at least 100 000 counts for each data point (C_i). The data should be acquired for 10 s or 100 000 counts, whichever requires the longest time.

(7) Continue acquiring data points. Each measurement should be performed when the observed count rate (OCR) drops by 10 000 counts/s below the previous measurement. The last (n-th) data point should be acquired when the OCR drops below 4000 counts/s.

Data analysis

Determine the OCR_i for each data point by the following formula:

$$OCR_i = \frac{(C_i - Bg) \times t_i \times 0.693}{21672 \times \left(1 - \exp\left(\left(\frac{-t_i}{21672}\right) \times 0.693\right)\right)} \quad (6)$$

where all measurements are in seconds, or fractions thereof, and 21 672 is the half-life of ^{99m}Tc in seconds.

(1) Determine the input count rate (ICR_i) for each data point by the following formula:

$$ICR_i = OCR_i \times \exp\left(\left(\frac{(t_n - t_i}{21672}\right) \times 0.693\right) \quad (7)$$

(2) Determine the OCR at 20% count loss by linear interpolation between the two closest points of Eq. (8):

$$OCR_i = 0.8 \times ICR_i \quad (8)$$

The purpose of Eq. (6) is to scale the measurements (calculate the OCR corrected for background) to the time when the measurement of each data point, t_i, began.

When the measurement is performed at high count rates, the duration of the measurement has little or no effect. That is to say, that ^{99m}Tc will decay very little during the few seconds of measurement at the high count rates of, for example, 150 000 counts/s. However, when the count rate approaches 4000 counts/s, the measurement will take more than 25 s to collect 100 000 counts, during which time ^{99m}Tc will decay by roughly 0.01%.

Equation (6) takes care of this problem by subtracting the appropriate number of background counts from the number of counts collected and then scaling this number according to the exponential decay law. The resulting OCR

points are not scaled to the time at which the measurement of each data point began, i.e. OCR_i is the number that would be obtained if it were possible to measure instantaneously only those counts coming from the source (no background).

Equation (7) simply extrapolates the measurement taken at the last point (at a very low count rate) to the points at high count rates. This is reasonable because the dead time at the count rates below 4000 counts/s is only a fraction of a per cent. This relative error is propagated through extrapolation, but it is not amplified (i.e. it will always have the same percentage value).

The largest effect, by far, is caused by the background measurement. Effort should be maintained to minimize variations in background throughout the test.

The test takes approximately two days. Only attempt this test if environmental conditions (e.g. power, temperature and humidity) and background conditions are stable. An alternative method, described next, can be used.

2.3.11.2. Intrinsic count rate performance — copper absorber method (alternative 2)

This method is described in NEMA Standards Publication NU 1 1994 [6] but not in the later publication NU 1 2001: Performance Measurements of Scintillation Cameras [8]. The procedure below derives from the NU 1 1994 publication.

Materials

- A radiation source consisting of ^{99m}Tc in solution contained in a small vial. Place the vial in an open lead pot at least 4 mm thick and with a well depth of 50 mm. The initial activity should be about 20 MBq (0.5 mCi).
- Fifteen absorbers fabricated from sheet copper about 0.25 cm thick, each about 6 cm × 6 cm square, numbered consecutively 1 to 15.
- Lead mask (see Section 2.1.7.8).
- Linear graph paper.

Part 1: Calibration of absorbers

The absorbers must first be accurately calibrated with respect to their attenuation of ^{99m}Tc gamma radiation. This may be done as follows.

Procedure

(1) Remove the collimator from the detector head. Turn the head to face vertically downward.
(2) Position the lead mask centrally on the crystal housing.
(3) Position the source on the central axis of the detector at a distance of about 1.5 m from its face (see Fig. 31).
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(5) Remove the source. Count the background for a preset time of 100 s. Record the background count rate. Replace the source.
(6) Place absorbers 13–15 over the source in numerical order, with absorber 13 uppermost. These absorbers remain in place for the rest of the procedure, providing scatter free transmitted radiation (see Fig. 20). Adjust the source activity so that the OCR is in the range 1000–3000 counts/s.
(7) With absorbers 13–15 in place, count for a preset time of 200 s. Record on an appropriate form (Table 3) the exact time of day corresponding to the midpoint of the measurement and the net count rate, A_0, corrected for background.
(8) Place absorber 12 on top of absorber 13. Count for a preset time of 100 s. Record on the form the exact time of day corresponding to the midpoint of the measurement and the net count rate, A_1, corrected for background.
(9) Place absorber 11 on top of absorber 12. Count for a preset time of 100 s. Record on the form the exact time of day corresponding to the midpoint of the measurement and the net count rate, A_2, corrected for background.
(10) Continue for each absorber 10, 9, etc., finishing with absorber 1.
(11) Remove all absorbers.

Data analysis

Follow the instructions below to fill in the form shown in Table 3.

(1) Correct the value of A measured in step (7) for radioactive decay to the times of day corresponding to the midpoints of each of the measurements of steps (8), (9) and (10). Enter on the form the corrected count rates A_0', A_0'', A_0''', etc.

TABLE 3. FORM USED WHEN CALIBRATING THE COPPER ABSORBERS

Identity of added absorber	Time of day	Count rate without added absorber[a] (counts/s)	Count rate with added absorber[a] (counts/s)	Attenuation factor
		A_0		
1		A_0'	A_1	$f_1 = A_1/A_0'$
2		A_0''	A_2	$f_2 = A_2/A_0''$
3		A_0'''	A_3	$f_3 = A_3/A_0'''$
4		etc.	etc.	etc.
5				
6				
7				
8				
9				
10				
11				
12				

[a] Corrected for radioactive decay to the time of the relevant measurement.

(2) For each of the absorbers 1–12, calculate the attenuation factor, *f*, given by the ratio of the count rate A_1, A_2, A_3, etc., to the corresponding corrected value of A_0', A_0'', A_0''', etc. This factor is the ratio of transmitted to incident gamma radiation flux for the absorber in question. Enter on the form the values of *f*. (With sheet copper absorbers 0.25 cm thick, the values should be about 0.6.)

(3) Calculate the mean, *f*, of the individual values of f_i. Examine the dispersion of the latter about the former. If the copper sheets have a uniform thickness such that no individual value differs from *f* by more than 1%, the single value *f* may be used in their place. Otherwise, the individual measurements are to be used.

Part 2: Determination of count rate

Once the absorbers are calibrated, continue with part two of the test. Once the absorbers are calibrated, they should rarely require recalibration. If precalibrated absorbers are available, only steps (1)–(5) of the procedure of part 1 are necessary.

Procedure

(1) Increase the source activity so that the observed count rate is in the range 1000–3000 counts/s with absorbers 1–15 in place over the source in numerical order, with absorber 1 uppermost.
(2) With absorbers 1–15 in place as described, register the count for a preset time of 100 s. Record on an appropriate form (Table 4) the exact time of day corresponding to the midpoint of the measurement and the net OCR, C_0, corrected for background. (At this relatively low count rate, count loss should be negligible and, hence, the ICR, R_0, and the OCR, e.g. C_0, should be equal.)
(3) Remove the uppermost absorber, absorber 1, thereby increasing the incident gamma radiation flux and the ICR in inverse proportion to the attenuation factor of the absorber removed. Record the counts for a preset time of 20 s. Record on the form the exact time of day corresponding to the midpoint of the measurement and the net count rate, C_1, corrected for background.
(4) Remove absorber 2. Again, register the counts for a preset time of 20 s. Record on the form the exact time of day corresponding to the midpoint of the measurement and the net count rate, $C_{1–2}$, corrected for background.
(5) Continue until only absorbers 13–15 remain over the source.
(6) Remove the source and lead mask. Replace the collimator.

Data analysis

(1) Correct the value of C_0 measured in step (2) for radioactive decay to the times of day corresponding to the midpoints of each of the measurements of steps (3)–(5). Enter on the form the corrected count rates, C_0', C_0'', C_0''', etc. If the time between the midpoints of the measurement of step (2) and the final measurement of step (5) is less than 10 min, this correction may be omitted and the uncorrected value of C_0 used in calculation. (It should be noted that all points on the curve are calculated based on C_0; therefore, this measurement must be highly accurate. Note also that if corrections

TABLE 4. FORM TO BE USED IN DETERMINATION OF COUNT RATE PERFORMANCE

Absorbers removed	Time of day	OCR with all absorbers in place[a] (counts/s)	OCR with one or more absorbers removed (counts/s)	Cumulative attenuation factor	Input count rate (counts/s)
		C_0			$R_0 = C_0$
1		C_0'	C_1	f_1	$R_1 = C_0'/f_1$
1–2		C_0''	C_{1-2}	$f_1 \times f_2$	$R_{1-2} = C_0''/f_1 \times f_2$
1–3		C_0'''	C_{1-3}	$f_1 \times f_2 \times f_3$	$R_{1-3} = C_0'''/f_1 \times f_2 \times f_3$
1–4		etc.	etc.	etc.	etc.
1–5					
1–6					
1–7					
1–8					
1–9					
1–10					
1–11					
1–12					

[a] Corrected for radioactive decay to the time of the relevant measurement.

for radioactive decay are applied, this must be done as indicated. In particular, it is not permissible to refer back the OCR recorded in steps (3)–(5) to the time of the measurement of step (2).)

(2) Assuming that count loss is negligible under the conditions of the measurement of step (2), so that the corrected values of C_0 also represent the ICR with all absorbers in place, calculate the ICRs, R_1, R_{1-2}, R_{1-3}, etc., for the conditions of each of the measurements of steps (3)–(5), by dividing the corrected values of C_0 by the corresponding cumulative attenuation factors for the absorbers removed. (Thus, the ICR after removal of absorber 1 is given by C_0'/f_1; the rate after removal of absorbers 1 and 2 is given by $C_0''/f_1 \times f_2$; the rate after removal of absorbers 1, 2 and 3 is given by $C_0'''/f_1 \times f_2 \times f_3$ and so on). If, as previously indicated, the dispersion of the individual values of f is sufficiently small, the single value

may be used in their place and the cumulative attenuation factors then become f, f_2, f_3, etc., otherwise, the individual values are to be used. Enter on the form the ICRs.

(3) Record the results on a graph showing OCR, C, against ICR, R, on linear graph paper (see Fig. 32). A spreadsheet could also be used.

(4) Determine from the graph the values of C and R for which:

$$C = 0.8R \tag{9}$$

These values correspond to a 20% count loss and are thus those for $C_{-20\%}$ and $R_{-20\%}$.

(5) Determine from the graph the maximum OCR.

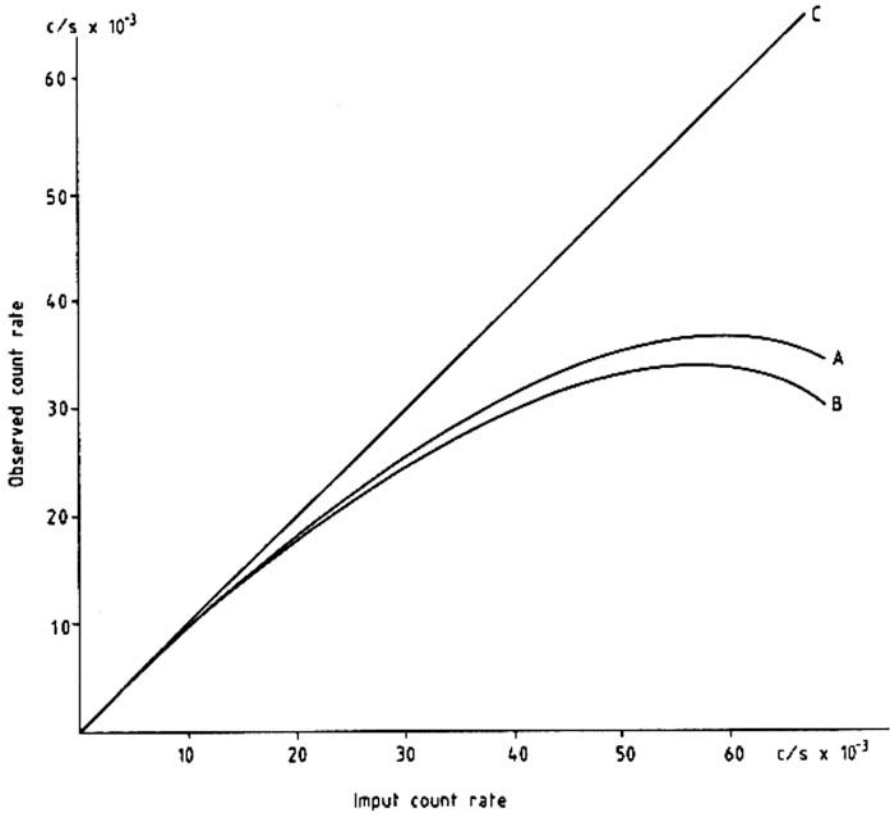

FIG. 32. Test of intrinsic count rate performance (alternatives 1 and 2). Graph of OCR against ICR. A: OCR data obtained with scintillation camera alone; B: OCR data obtained on digital image processor; C: line of identity for no count loss.

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals. For the images stored in the computer, the counts in the entire image frame should be used. In this test, many newer cameras will automatically switch to a high count rate mode without operator interaction.

Interpretation of results

At acceptance testing, the graph of OCR against ICR should be compared with the manufacturer's worst case specifications. The values of $R_{-20\%}$ in the low and high count rate modes should similarly be compared with the manufacturer's worst case values. The value of maximum count rate should be similarly treated. The effect of multiple detector heads may reduce the count rate response and should be particularly investigated.

At routine testing, the values of $R_{-20\%}$ and the maximum count rate should be compared with the reference values.

Limits of acceptability

At acceptance testing, a value of $R_{-20\%}$ that is 10% or more below the manufacturer's worst case value would call for corrective action initiated through the manufacturer's representative.

At routine testing, a change in the value of $R_{-20\%}$ by more than ±20% from the reference value would call for follow-up action.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.11.3. Test of intrinsic count rate performance — two source method (alternative 3)

Purpose of test

To test the intrinsic count rate performance of a scintillation camera in terms of the count rate corresponding to a 20% count loss (two source method). This method is appropriate for most cameras manufactured before 1996.

Materials

- Two point sources each consisting of about 2 MBq (50 μCi) of ^{99m}Tc in solution in suitable containers. The count rate from both sources together under the conditions of the test should be similar to the manufacturer's specified or worst case value for the OCR corresponding to a 20% count loss. The activities of the sources should be within 10% of each other.
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Turn the head to face horizontally.
(2) Position the lead mask centrally on the crystal housing.
(3) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Suspend one source in air near the central axis of the detector and away from other objects so as to minimize radiation scatter at a distance of 1 m or more from the detector face.
(5) Register the count for a sufficient time to accumulate a count of 10^6. Record the count rate.
(6) Suspend the second source beside the first, but so that neither interferes with the detector's 'view' of the other. Register the count for the two sources for the same time period. Record the count rate.
(7) Remove the first source. Register the count for the second source alone for the same time period. Record the count rate.
(8) Remove the second source. Register the background count for the same time period. Record the background count rate.
(9) Repeat steps (5), (6) and (7), reversing the order of the sources.
(10) Remove the remaining source and lead mask. Replace the collimator.

Data analysis

(1) Express all data as net count rates (counts/s) corrected for background.
(2) Calculate for each set of data the pulse pair resolving time, τ, in seconds from:

$$\tau = \frac{2 \times R_{12}}{\left(\mathrm{i}_1 + R_2\right)} \times \ln \sqrt{\frac{\left(R_1 + R_2\right)}{R_1 \times R_2}} \tag{10}$$

where R_1 and R_2 are the net count rates of the first and second sources and R_{12} is the net count rate of the two sources together, all in counts per second. Average the two values to obtain τ.

(3) Calculate the ICR for a 20% loss, $R_{-20\%}$, from:

$$R_{-20\%} = \frac{1}{\tau} \times \ln\left(\frac{10}{8}\right) = \frac{0.2331}{\tau} \tag{11}$$

(4) Calculate the OCR for a 20% count loss, $C_{-20\%}$, from:

$$C_{-20\%} = 0.8 \times R_{-20\%} \tag{12}$$

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals. If the manufacturer's specifications are not available, the count rate of the two sources together should be about 120 000 counts/s for cameras manufactured in the 1990s and later, 60 000 counts/s for scintillation cameras manufactured between 1978 and 1989 and 20 000 counts/s for cameras manufactured earlier.

In order to eliminate the effect of radioactive decay, the same time interval should be maintained between the three measurements in each data set.

An alternative to suspending the sources in air is to place them 1 m from the detector face with a 6 mm or more thickness of sheet copper interposed between source and detector. This method will also minimize radiation scatter.

If the scintillation camera is fitted with a high count rate mode circuit, the test should be repeated with this circuit enabled. For the images stored in the computer, the counts in the entire image frame should be used.

This method is not appropriate for fully digital scintillation camera systems (most cameras manufactured from the mid-1990s onwards).

Interpretation of results

At acceptance testing, the values of $R_{-20\%}$ in the low and high count rate modes should be compared with the manufacturer's worst case values.

At routine testing, the values of $R_{-20\%}$ should be compared with the reference values.

Limits of acceptability

At acceptance testing, a value of $R_{-20\%}$ that is 10% or more below the manufacturer's worst case value would call for corrective action initiated through the manufacturer's representative.

At routine testing, a change in the value of $R_{-20\%}$ by more than ±20% from the reference value would call for follow-up action.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.11.4. Test of maximum count rate (alternative 4)

Purpose of test

To test the maximum count rate of a scintillation camera.

Materials

- Point source (see Section 2.1.7.8) consisting of about 4 MBq (100–500 μCi) of ^{99m}Tc in solution in a suitable container.
- Lead mask (see Section 2.1.7.8).
- Movable stand with mounting for point source.

Procedure

(1) Remove the collimator from the detector head. Turn the head to face horizontally.
(2) Position the lead mask centrally on the crystal housing.
(3) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Mount the source on the movable stand. Position the latter so that the source is on the central axis of the detector (see Fig. 33). To minimize radiation scatter, the source should not be close to other objects.

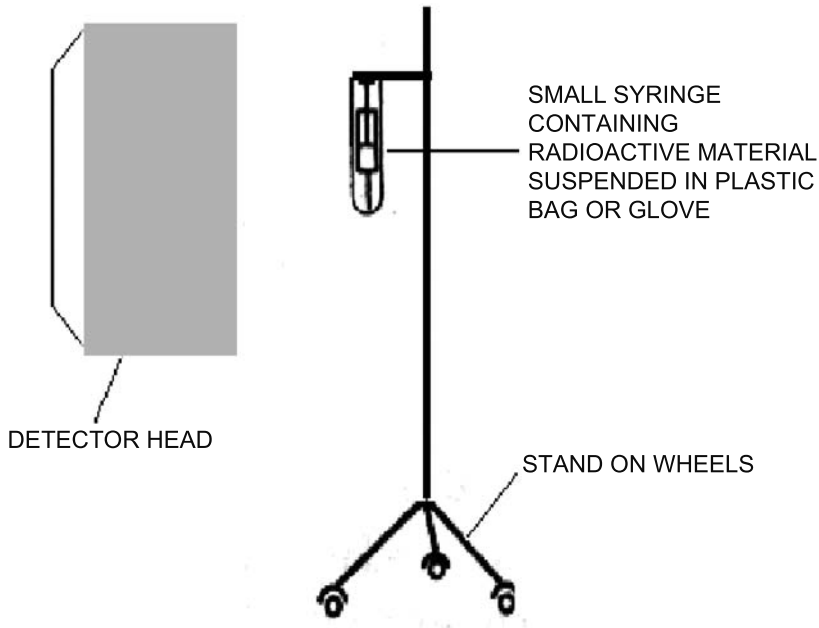

FIG. 33. Test of maximum count rate. Positioning of point source in relation to detector.

(5) Register the count rate as the source is moved progressively closer to the detector face. The count rate will increase to a maximum and then decrease. Record the maximum count rate.
(6) Remove the source, stand and lead mask. Replace the collimator.

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals. This test is very susceptible to scatter.

Interpretations of results

At acceptance testing, the value of the maximum count rate should be compared with the manufacturer's worst case value with like circuits enabled. This parameter is useful only as a camera characteristic that can be measured easily and routinely. In clinical imaging, the camera cannot be operated at the maximum count rate.

Limits of acceptability

At acceptance testing, a value of maximum count rate that is 10% or more below the manufacturer's worst case value would call for corrective action initiated through the manufacturer's representative. At routine testing, a change in the value of maximum count rate by more than ±20% from the reference value would call for follow-up action.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.12. Test of basic computer timing

Purpose of test

To test that the basic timing functions of the computer in the camera–computer system are correctly performed.

Materials

- Radiation source consisting of ^{99m}Tc in solution contained in a small vial placed in an open lead pot with walls and floor about 6 mm thick. The initial activity should be about 10 MBq (300 μCi).
- Stopwatch.

Procedure

(1) Set the real time clock of the computer to the correct time of day, if possible.
(2) Remove the collimator from the detector head.
(3) Place the point source within the field of view of the camera in such a way as to produce a count rate of approximately 5000 counts/s.
(4) Set up a static data acquisition with a requested collection time of 100 s.
(5) Start data acquisition and the stopwatch simultaneously.
(6) Record the stopwatch time at the end of the acquisition.
(7) Record the collection time as indicated by the computer.
(8) If there is access to the real time clock, record the clock time at the start and end of the data collection or look at the elapsed time recorded by the computer.

(9) Repeat steps (3)–(8) with source placed to produce a count rate of approximately 40 000 counts/s.
(10) Remove the source. Replace the collimator.
(11) If appropriate, record the time of day given by the computer some hours later.

Data analysis

Compare the requested collection time with the times indicated by the stopwatch and the computer. Note any differences. Also, note any differences between the correct time of day and that given by the computer.

Observations

This test is intended to be performed as a reference test at the time of acceptance and at half-yearly intervals. The test is appropriate only for camera–computer systems in which data acquisition is controlled by the computer independently of the camera start/stop operation. The clock used for timing an acquisition may be different from the ordinary day clock and both need to be tested, the former by timing the acquisition and the latter by recording the time of day.

Interpretation of results

Any time differences greater than the accuracy of timing by the stopwatch (e.g. 0.1 s) are significant. If any such differences are noted, the procedure should be repeated. A large systematic error may be due to the difference between 50 Hz and 60 Hz.

Limits of acceptability

A time difference of 1% or greater at either count rate may indicate a failure and should be investigated further.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.13. Test of computer timing in dynamic acquisition

Purpose of test

To test that the timing functions of the computer in a camera–computer system are correctly performed in dynamic acquisition.

Materials

- Radiation source consisting of ^{99m}Tc in solution contained in a small vial placed in an open lead pot at least 4 mm thick. The initial activity should be about 10 MBq (300 μCi).
- Stopwatch.

Procedure

(1) Remove the collimator from the detector head.
(2) Place the point source within the field of view of the camera in such a way as to produce a count rate of approximately 20 000 counts/s.
(3) Acquire a single static image without zoom in the computer with a requested collection time of 20 s.
(4) Set up a dynamic acquisition protocol for a large number of frames with the shortest frame time allowed by the system (e.g. 100 frames of 0.2 s each with a total requested collection time of 20 s).
(5) Start the acquisition and the stopwatch simultaneously.
(6) Record the elapsed time as indicated by the stopwatch at the end of the data acquisition.
(7) In acceptance testing, repeat steps (5)–(7) for each available data collection format (e.g. 64 × 64 byte, 64 × 64 word, 128 × 128 byte), including list mode where appropriate.
(8) Remove the source. Replace the collimator.

Data analysis

(1) Determine the total count, C_s, in the static image.
(2) Determine the count, C_f, in each frame of the dynamic study.
(3) Perform a χ^2 test on the C_f values to establish whether the variation in counts can be plausibly attributed to chance alone, referring to tables of χ^2 to obtain its 95% confidence limits for the corresponding sample size and number of degrees of freedom.

(4) For each frame, calculate the apparent frame time, T_f, by the formula:

$$T_f = \frac{C_f \times T_s}{C_s} \quad (13)$$

where T_s is the collection time for the static image (20 s).

(5) Calculate the mean of the set of apparent frame times.

(6) Calculate the apparent collection time for the dynamic study as the sum of the apparent frame times. Calculate the requested collection time as the sum of the requested frame times.

Observations

This test is intended to be performed as a reference test at the time of acceptance and at half-yearly intervals.

In an ideal system, the requested collection time, the indicated elapsed time and the apparent collection time should be identical. There are two types of timing error. Time may be lost between frames (see Fig. 34(a)), in which case the elapsed time exceeds the requested collection time. Alternatively, the individual frame times may differ systematically from the requested frame time, as will be evident from their calculated values. In this case, the apparent collection time differs from the requested collection time (see Fig. 34(b)).

Random fluctuations in frame times will result in an unacceptably large χ^2 value. The activity in the source, matrix size and collection time should be such that saturation in the digital image does not occur.

Interpretation of results

If the variation in apparent frame times is excessive, this may indicate malfunction in either the camera–computer interface or the camera (e.g. a drifting PHA window). In such a case, the camera should be checked carefully to ensure that it is performing satisfactorily before further investigations are carried out. If anomalous results are obtained, it is desirable to repeat the tests at higher and lower count rates (e.g. 40 000 and 5000 counts/s).

Limits of acceptability

Time lost between frames should be not more than 5% of the shortest frame time and time lost per frame should likewise be not more than 5%.

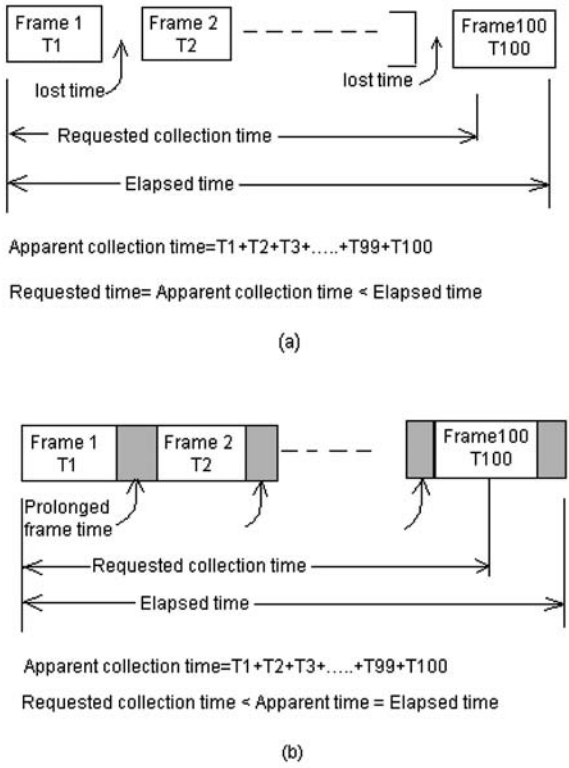

FIG. 34. Computer timing errors: (a) loss of time between frames, (b) prolonged individual frame times.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.14. Test of ECG gated acquisition

Purpose of test

To test that a camera–computer system used in ECG gated acquisition is able to respond properly to the ECG signal.

Materials

- Point source (see Section 2.1.7.8) consisting of about 100 MBq (3 mCi) of ^{99m}Tc in solution in a suitable container. A source and container as used in the test in Section 2.3.11 may also be used in this test.
- ECG.

Procedure

(1) Place the point source within the field of view of the camera.
(2) Connect a normal volunteer to the ECG leads. The volunteer should relax during the test. No radioactivity is injected.
(3) Start an ECG gated acquisition using a normal clinical protocol as regards collection time, counts or number of heartbeats.
(4) In acceptance testing, repeat the study with the volunteer occasionally moving an arm to produce spikes on the ECG and to check that the beat–rejection system is functioning.
(5) If gated list mode collection is available, repeat steps (1)–(4) in this mode.
(6) Remove the point source. Disconnect the ECG leads from the volunteer.

Data analysis

(1) Define an ROI that includes the image of the point source and generate a time–activity curve.
(2) Calculate the mean and the maximum deviations from the mean of the data points in the first three quarters of the time–activity curve.

Observations

This test is intended to be performed as a reference test at the time of acceptance, at half-yearly intervals and if problems are suspected.

It is important that the interface be triggered correctly from the ECG and that the gated time–activity curves (volume curves) which result are undistorted. There may be significant delay between the occurrence of the R wave and the triggering of the computer due to improper signal adjustment or the presence of electronic circuits between the ECG and the computer. The portion of the curve representing end-systole may be shifted by improper triggering. The R wave detection is often based on the positive slope of the R wave generated by the ECG monitor. A negative R wave will, in such a situation, create a delay in end-diastole. Many sophisticated phantoms have been designed for checking the shape of the time–activity curve and for

ensuring that the timing of the R wave detection is correct. Clinical left ventricular volume curves generated in a gated cardiac study should be examined carefully to confirm that the start of the volume curve is at end-diastole and that no delay has been introduced.

An ECG simulator should be available for checking the integrity of the data collection system and replaces the patient in the test above. The test proposed here is simpler in that it checks whether the system can actually respond to a patient's ECG. It is not a complete test, but adequate if the rest of the software and hardware perform according to specifications.

Some systems may exhibit counts falling off at the end of the time–activity curve due to variations in the length of the cardiac cycle of the subject (when a fixed frame time is applied).

Interpretation of results

The time–activity curve should be examined carefully as regards constancy of counts over the first three quarters of the curve. Any deviation of the time–activity curve from a horizontal line is significant, especially over the first three quarters of the curve.

Limits of acceptability

Limits of acceptable variation may be set at three times the standard deviation of the random counting error (the square root of the mean count) over the first three quarters of the time–activity curve.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.15. Test of multiple window spatial registration

Purpose of test

To test that the images acquired at different photon energies superimpose when more than one PHA is used simultaneously in an additive or subtractive mode.

Materials

- Point source consisting of about 40 MBq (1 mCi) of ^{67}Ga in solution in a small vial, in a lead shield 6 mm thick and having a circular aperture 3 mm in diameter.
- Quadrant bar or OHTP phantom.

2.3.15.1. Method 1: Visual image method

Procedure

(1) Remove the collimator from the detector head. Turn the head to face vertically. Place the spatial resolution phantom as close to the crystal face as possible. Place the unshielded source in a source holder on the ceiling, aligned with the centre of the crystal.
(2) If the scintillation camera has two PHAs, centre one default PHA window on each of the 93 keV and 296 keV photopeaks. If three PHAs are available, centre two windows as above and centre a third window on the 184 keV photopeak.
(3) Adjust the source activity so that the count rate does not exceed 10 000 counts/s in any PHA channel when the source is placed close to the exposed face of the crystal housing.
(4) Acquire separate images through each of the PHA channels at a preset count of at least 5×10^6 using the largest matrix size available.
(5) Acquire an image with all three PHA channels contributing to the image.
(6) Remove the source and replace the collimator.

2.3.15.2. Method 2: Digital analysis method

(1) Remove the collimator from the detector head. Turn the head to face horizontally. Put a table directly adjacent to the scintillation camera as a source support. Place the source on the table.
(2) If the scintillation camera has three PHAs, centre the default PHA windows on the 93, 184 and 296 keV photopeaks.
(3) Adjust the source activity so that the count rate does not exceed 10 000 counts/s in any PHA channel when the source is placed close to the exposed face of the crystal housing.
(4) Position the source on the X+ axis of the detector face at about 75% of the distance from the centre to the edge, noting the exact source position.

(5) Acquire separate digital images in a 256 × 256 matrix through each of the PHA channels, acquiring about 10 000 counts in the pixel with the highest count.
(6) Repeat steps (4) and (5) for a source position on the X– axis at about 75% of the distance from the centre to the edge, and for similar positions on the Y+ and Y– axes.
(7) Remove the source. Measure accurately the distance between the two X and two Y source positions.
(8) Replace the collimator.
(9) Determine the coordinates of the X and Y source positions on each image, using count profiles or ROIs.

Data analysis

(1) In the visual method, carefully compare the images acquired with each single PHA window with the image collected with the three PHA windows combined. If the image of the combined PHA windows shows any degradation of resolution or any evidence of a double image not seen in the single PHA window images, misregistration may be the cause and the digital method should be performed (see Fig. 6).
(2) In the digital method, if the locations at which the sources appear or the addresses of the pixels with the highest counts do not coincide, determine the displacements, in millimetres, in the X and Y directions for each image, using the measured distances between the source positions to derive a scale or a conversion factor in millimetres per pixel relating image distance to object distance.

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals. It should also be performed if degradation in the quality of images acquired with the simultaneous use of more than one PHA is noted.

For a further discussion of this test and further examples refer to Section 2.4 of Ref. [2].

Interpretation of results

At acceptance testing, preferably carried out by method 2, the values of the X and Y displacements should be compared with the manufacturer's worst case values. The visual method is not accurate enough to determine small

displacements, but will alert the user to a large displacement that would affect the use of the multiple PHA capability.

At routine testing, the values of the X and Y displacements should be compared with the reference values. If the multiple PHA capability is used clinically, the test should be performed on a routine basis.

Limits of acceptability

At acceptance testing by method 2, a value of the X or Y displacement that is 10% or more above the manufacturer's worst case value would call for corrective action initiated through the manufacturer's representative.

At routine testing by method 2, a change in displacement by more than 20% from the reference value would call for similar corrective action.

At either acceptance or routine testing by method 1, significant observed displacement would call for follow-up action. An absolute position difference between two PHA windows should never exceed 1–2 mm. Pending corrective action, clinical studies with a single PHA channel could continue.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.16. Test of detector head shielding leakage

Purpose of test

To test that the detector head of a scintillation camera responds only to radiation incident upon the crystal after transmission through the collimator.

Materials

Point sources (see Section 2.1.7.8) consisting of about 4 MBq (100 μCi) of ^{99m}Tc and a radionuclide with a photon energy corresponding to the specified design energy of the camera in a suitable container.

Procedure

(1) Mount a collimator appropriate to the gamma radiation energy of the source on the detector head.

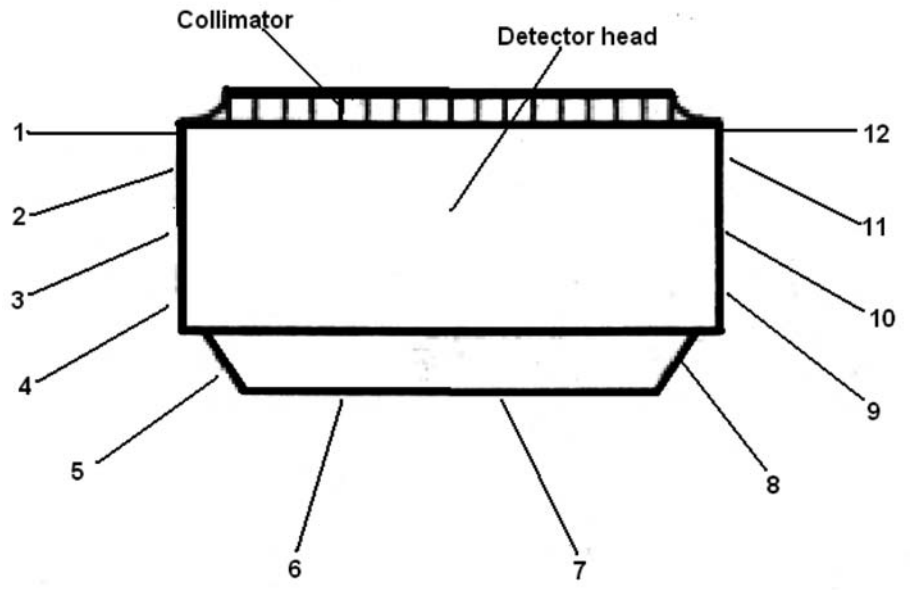

FIG. 35. Test of detector head shielding leakage. Twelve sites at which to position the source in order to test for shielding leakage.

(2) Centre the manufacturer's default PHA window for the radionuclide concerned on the photopeak (see Section 2.4.2).
(3) Position the source consecutively at twelve sites around the detector head shielding and record the number of counts at each site for a preset time of 100 s (see Fig. 35). In addition, investigate sites of joints in the shielding, exit points of cables and other reduced shielding areas.
(4) Position the source in the centre of the field of view at a distance of 10 cm from the face of the collimator. Record the number of counts for a preset time of 100 s.
(5) Remove the source and measure the background count, *B*, for the same time period.

Data analysis

Determine the maximum count from the measurements made around the detector shielding. Calculate the shielding leakage by dividing this maximum count by the count obtained through the collimator and expressing this as a percentage.

Observations

This test is intended to be performed as an acceptance test. This test should be performed for ^{99m}Tc and a radionuclide having an energy that corresponds to the maximum design energy of the camera.

Interpretation of results and limits of acceptability

If any abnormal results are recorded, the test should be repeated after checks to make sure that there are no nearby radiation sources, including patients to whom radioactive materials have been administered, and that there is no radioactive contamination of the instrument or its surroundings. If the abnormality persists, corrective action should be initiated through the manufacturer's representative. For ^{99m}Tc, leakage should be negligible. For higher energy radionuclides, leakage should meet the manufacturer's specification.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.17. Test of routine spatial resolution and spatial linearity

Purpose of test

To test the spatial resolution and spatial linearity of a scintillation camera on a weekly basis.

2.3.17.1. Method 1: Flood source method

To be used if a flood source is available.

Materials

- ^{99m}Tc flood phantom (see Section 2.1.7.8) containing about 200–400 MBq (5–10 mCi); or
- ^{57}Co flood source of similar activity.

- OHTP phantom (see Section 2.1.7.8) matched to camera resolution. Optimal hole diameter and minimum interhole spacing, *s*, is given by:

$$s = \text{FWHM}/1.75 \quad (14)$$

Procedure

(1) Mount a low energy, high resolution parallel hole collimator on the detector head. The same collimator must be used consistently in the test. Turn the head to face vertically upward.
(2) Position the OHTP phantom on the face of the collimator with the pattern carefully aligned with the X and Y axes of the detector face.
(3) Place the flood phantom or flood source on the OHTP phantom.
(4) Centre the manufacturer's default PHA window on the photopeak of the radionuclide concerned (see Section 2.4.2).
(5) Acquire an image at a preset count of at least 10^6. Use the maximum matrix size available.
(6) Remove the flood phantom or flood source and the OHTP phantom.

Data analysis

Visually inspect the image, noting particularly whether the images of the holes are distinct and separate from each other and whether there are significant deviations from linearity in the X or Y direction.

Observations

This test is intended to be performed as a reference test at the time of acceptance and routinely at weekly intervals. The test may be performed with a quadrant bar phantom or a PLES phantom in place of the OHTP phantom. The quadrant bar phantom has the advantage that each quadrant has a set of lines of different widths and the same phantom can be used for different cameras. However, it is advisable to image the smallest bars in each direction in the field of view. This requires eight acquisitions. With care, over time, this can be accomplished by selective rotation and inversion of the phantom weekly.

The OHTP phantom has the advantage that it allows the entire field of view to be examined simultaneously in both the X and Y directions with one acquisition. However, its hole diameter and interhole spacing must be matched to the spatial resolution of the camera/collimator for it to be a critical test.

Selection of the appropriate phantom thus requires prior knowledge of the resolution (unless a set of phantoms of differing hole sizes is available).

If a PLES phantom is used, the bar width and inter-bar spacing must be matched to the resolution of the camera/collimator for a critical test. As with the OHTP phantom, therefore, selection of the appropriate phantom requires prior knowledge of the resolution (unless a set of phantoms of differing bar widths is available). Moreover, such a phantom must be imaged in two positions at 90° to each other for an examination of the entire field of view in the X and Y directions.

Interpretation of results

The image should be compared with the reference image and with recently acquired images to identify any changes and trends in either spatial resolution or spatial linearity. If deterioration in resolution is noted, corrective action must be initiated.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.3.17.2. Method 2: Point source method (alternative 1)

To be used if a flood source is not available.

Materials

- Point source (see Section 2.1.7.8) consisting of 40–100 MBq (1–3 mCi) of ^{99m}Tc in solution in a suitable container.
- Source mounting for point source (see Section 2.1.7.8).
- OHTP phantom (see Section 2.1.7.8) matched to camera resolution. Optimal hole diameter and interhole spacing, *s*, is given by Eq. (14).

Procedure

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the OHTP phantom so that it is supported on the detector head housing and located as close to the crystal housing as possible, with the rows of holes carefully aligned with the X and Y axes of the detector face.

(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak.
(5) Acquire an image at a preset count of at least 10^6. Use the maximum matrix size available.
(6) Remove the source and the OHTP phantom. Replace the collimator.

Data analysis

As for Method 1: Flood source method.

Observations

If ^{99m}Tc is used, the point source method has the advantage of requiring a lower activity than the flood source method. Further, it does not require the filling of a phantom and thus exposes personnel to a lower radiation dose. Its disadvantage is that it requires the collimator to be removed from the detector head, with increased chance of crystal damage. The method that is chosen should be performed consistently.

Interpretation of results

As for Method 1: Flood source method.

Conclusion

As for Method 1: Flood source method.

2.3.17.3. Method 2: Digital analysis method (alternative 2)

This test is to be used to test the spatial linearity and spatial resolution of a camera–computer system on a weekly basis.

Materials

- Point source (see Section 2.1.7.8) consisting of 40–100 MBq (1–3 mCi) of ^{99m}Tc in solution in a suitable container.
- Source mounting for point source (see Section 2.1.7.8).
- The coarsest available OHTP phantom (e.g. hole diameter and interhole spacing 4.8 mm).

Procedure

(1) Remove the collimator from the detector head. Align the head and the source mounting.
(2) Position the OHTP phantom so that it is supported on the detector head housing and as close to the crystal housing as possible, with the rows of holes carefully aligned with the X and Y axes of the detector face.
(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak.
(5) Acquire an image at a preset count of at least 10^6. Use the maximum matrix size available.
(6) Remove the source and the OHTP phantom. Replace the collimator.

Data analysis

Method A (requiring no special software):

(1) Visually inspect the image, noting particularly whether the individual holes are distinct and separated from each other over the entire field of view, and whether there are significant deviations from linearity in the X or Y direction over the field.
(2) As a visual aid, place a horizontal marker line, as used for profile generation, on the digital image. Determine whether the separation of the rows is constant by noting the displacements needed to align the marker line with the consecutive lines of holes. Repeat with a vertical marker line for the columns.

Method B (requiring special software):

(1) Determine the locations of the centres of the images of the individual holes and whether they lie in a regularly spaced manner.
(2) Estimate the spatial resolution over the field of view.
(3) Estimate the variation in point source sensitivity over the field of view.

Observations

This test is intended to be performed as a reference test at the time of acceptance and routinely at weekly intervals. The hole diameter and interhole spacing should be chosen so that the images of the individual holes are clearly separated.

Interpretation of results

The results should be compared with the reference results and with those of recently performed tests to identify any changes and trends in either spatial resolution or spatial linearity. Both images should be linear over the entire field of view without local distortion in the hole pattern. The images of all the holes should be identical, without local variation. Numerical values for deviations are difficult to obtain without special software.

Variations in the positions of the images of the holes can be caused by spatial distortion in the scintillation camera alone or in the entire system. Rotating the image by use of the image orientation switches may isolate the cause. If the distortion rotates, then the problem lies in the camera. If the distortion remains unchanged in position, then the problem is in the display system.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.4. OPERATIONAL CHECKS

2.4.1. Check of collimator and detector head mountings and collimator damage

Purpose of test

To check the collimator and detector head mountings in a scintillation camera. To check for any damage to the collimator.

Procedure

(1) Inspect all collimators and detector head mountings for mechanical or other defects, with particular regard to the safety of patients and staff. Check the detector head drive mechanism for correct function.
(2) Visually inspect the collimators for damage.

Interpretation of results

(1) Any abnormal finding should dictate immediate withdrawal of the instrument from operational use pending corrective action.
(2) If there is any suspicion of damage to the collimator, a system uniformity test using a flood source must be conducted immediately before further clinical studies are performed with the collimator.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate follow-up action taken.

2.4.2. Check of energy calibration of PHA

Purpose of test

To centre the manufacturer's default PHA window of a scintillation camera on the photopeak.

Materials

- Point source (see Section 2.1.7.8) consisting of about 30 MBq (700 μCi) of ^{99m}Tc or other radionuclide to be used clinically, in solution in a suitable container, giving a count rate not greater than 30 000 counts/s after completion of the calibration procedure.
- Mounting for point source (see Section 2.1.7.8).

2.4.2.1. Method 1: Recommended method for scintillation cameras fitted with a multichannel analyser

Procedure

(1) Without removing the collimator from the detector head, align the head and the source mounting.
(2) Mount the source in the source mounting.
(3) Set the PHA to the gamma radiation energy of the radionuclide in use.
(4) Centre the manufacturer's default PHA window on the photopeak, using the multichannel analyser display for this purpose.
(5) Record all relevant control settings.

2.4.2.2. Method 2: Alternative method for scintillation cameras not fitted with a multichannel analyser (if relevant instructions are not available)

Procedure

(1) Without removing the collimator from the detector head, align the head and source mounting.
(2) Mount the source in the source mounting.
(3) Set the PHA to the gamma radiation energy of the radionuclide in use.
(4) Proceed according to the instructions in the operation manual.
(5) Record all relevant control settings.

Data analysis

Record the results on a control chart designed to cover an interval of about 3 months, which shows the PHA setting plotted against the date on linear graph paper. If a change from previous values is observed, the procedure should be repeated several times in succession and frequently thereafter to monitor for short term fluctuations.

Observations

The test can be performed with the collimator removed, if a point source consisting of about 4 MBq (100 μCi) of the radionuclide is used.

Interpretation of results

The high voltage or PHA setting should be compared with the reference value and with recent values to identify any changes or trends. Short term fluctuations in the PHA setting of a scintillation camera may arise from unstable power supplies, temperature changes or electronic circuit faults. Long term trends in the setting may indicate failure in one or more photomultipliers, deterioration of the crystal or physical separation of the photomultiplier–light guide assembly from the crystal. If short term fluctuations occur, it will not be possible to use the camera clinically until corrective action has been taken. If, however, the settings change only slowly, it may be possible to continue to use the camera, provided the energy calibration of the PHA is checked before each patient study.

Limits of acceptability

Any change in PHA setting from the reference value would call for further investigation.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.4.3. Check of flood field uniformity and sensitivity

Purpose of the test

The purpose of this test is to check the flood field uniformity and the sensitivity of a scintillation camera. The most convenient method should be followed.

2.4.3.1. Method 1: Flood source method (system uniformity) (to be used if a flood source is available)

Materials

- ^{57}Co flood source of 200 MBq (5 mCi) for a small field of view camera or 400 MBq (10 mCi) for a large field of view camera.
- Alternatively, a flood phantom (see Section 2.1.7.8) containing an accurately known activity of ^{99m}Tc, about 200 MBq (5 mCi) for a small field of view camera or 400 MBq (10 mCi) for a large field of view camera. The activity is determined by using a radionuclide (dose) calibrator to measure the syringe containing the radionuclide solution to be transferred to the phantom and the residual activity in the syringe after the transfer. The latter is then subtracted from the former. The exact time of day corresponding to the activity determination is also recorded.

Procedure

(1) Mount a low energy, parallel hole collimator on the detector head. The same collimator must be used consistently in the test. Turn the head to face vertically upward.

(2) Place the flood phantom or flood source at least 10 cm from the face of the collimator. (Maintaining a distance between the flood source and collimator is especially important when using a ^{57}Co flood source.)
(3) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(4) Acquire an image on the display device with a hard copy at a preset count of 2×10^6 for a small field of view camera or 5×10^6 for a large field of view camera.
(5) Record all imaging parameters, including the preset count, the count time and the time of day corresponding to the midpoint of the count.
(6) Remove the flood phantom or flood source.

Data analysis

(1) Visually inspect the image for non-uniformities.
(2) For a ^{57}Co flood source, calculate the activity of the source by correcting for decay on a weekly basis. For the ^{99m}Tc flood phantom, calculate the activity of the contents at the time of day corresponding to the midpoint of the count by correcting for radioactive decay from the time of the activity determination.
(3) Calculate the sensitivity in counts per second per becquerel.

Observations

The test should be performed on a daily basis in order to check the condition of the camera for clinical studies. Thus, if the camera has a uniformity correction circuit, the test should be performed on a daily basis with the circuit enabled. However, at the start of each week the test should, if possible, also be performed with the circuit disabled, to monitor for defects that may be hidden in the corrected images, e.g. from the early failure of a photomultiplier. It should be appreciated that the width of the PHA window considerably influences the measured sensitivity. The test should, therefore, always be performed with the same window width.

The contents of the flood phantom should be mixed thoroughly to provide a uniform source. If poor mixing is suspected, the phantom should be rotated through 90° and a new image acquired. Poor mixing is confirmed if the non-uniform features in the image move with the phantom (see Fig. 36).

For a further discussion of phantoms and more examples, refer to Section 2.2.10 of Ref. [3].

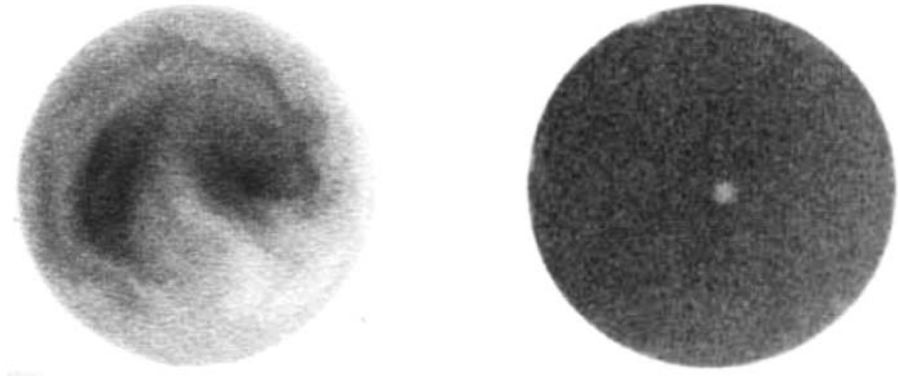

FIG. 36. Fillable flood sources: Non-uniform mixing and air bubbles. Left: An example of a routine extrinsic uniformity image using ^{99m}Tc in a fillable flood source. The ^{99m}Tc was injected into the water in the flood source and was left to disperse over a period of about one hour. The non-uniformity shows that dispersion was insufficient to produce a homogeneous distribution of ^{99m}Tc and that actual mixing must take place. Right: The image shows a small air bubble in the centre of the field of view. The cold indentation at the top left of the field of view was also due to an air bubble (see Ref. [3]).

Interpretation of results

The image should be compared with the reference image and with recent images to identify any changes or trends. The sensitivity value should likewise be compared with the reference value and with recent values.

No significant change in uniformity should be detectable. If non-uniform features are present, it may be possible to take account of them and proceed with clinical studies, depending on the extent of the defects and the clinical studies to be performed. In any case, the person who will interpret the clinical results must inspect the flood field image and take responsibility for proceeding. Any corrective action needed should be initiated as soon as possible.

Change in sensitivity may indicate incorrect energy calibration of the PHA or could result from impaired energy resolution or non-uniformity in flood field response.

Limit of acceptability

Any detectable change in uniformity would call for further investigation. A change in sensitivity by more than ±10% from the reference value would likewise call for further investigation.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.4.3.2. Method 2: Point source method (intrinsic uniformity) (to be used if a flood source is not available)

Materials

- Point source (see Section 2.1.7.8) consisting of a known activity (10–40 MBq (0.2–1 mCi) depending on the camera and the distance) of ^{99m}Tc in solution in a suitable container. The activity is determined by measurement in a radionuclide (dose) calibrator and the exact time of day that corresponds to the activity determination should also be recorded.
- Source mounting for point source (see Section 2.1.7.8).
- Lead mask (see Section 2.1.7.8).

Procedure

(1) Remove the collimator from the detector head. Align the detector head and source mounting.
(2) Position the lead mask centrally on the crystal housing.
(3) Mount the source in the source mounting.
(4) Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
(5) Acquire an image on the display device with hard copy, at a preset count of 2×10^6 for a small field of view camera or 5×10^6 for a large field of view rectangular detector.
(6) Record all imaging parameters, including the preset count and the count, time and the time of day corresponding to the midpoint of the count.
(7) Remove the source and source mounting. Replace the collimator.

Data analysis

(1) Visually inspect the image for non-uniformity.
(2) Calculate the activity of the source at the time of day corresponding to the midpoint of the count by correcting for radioactive decay from the time of the activity determination.
(3) Calculate the sensitivity in counts per second per becquerel.

Observations

The test should be performed on a daily basis in a manner to check the condition of the camera for clinical studies. Thus, if the camera has a uniformity correction circuit, the test should be performed on a daily basis with the circuit enabled. However, at the start of each week the test should, if possible, also be performed with the circuit disabled in order to monitor for defects that may be hidden in the corrected images, e.g. from an early stage of failure of a photomultiplier.

It should be appreciated that the width of the PHA window considerably influences the measured sensitivity. The test must, therefore, always be performed at the same window width. Equally, the distance between the source and the detector face must be kept constant.

This method has the advantage of requiring a lower activity than that described in Section 2.4.3.1 (Method 1: Flood source method). Further, it does not require the filling of a phantom and thus exposes personnel to a lower radiation dose. Its disadvantage is that it requires the collimator to be removed from the detector head, which increases the risk of crystal damage. The method chosen should be used consistently.

For a further discussion and examples, refer to Section 2.2.10 of Ref. [3].

Interpretation of results

As for Method 1: Flood source method.

Limits of acceptability

As for Method 1: Flood source method.

Conclusion

As for Method 1: Flood source method.

2.4.4. Check of background count rate

Purpose of test

To check the background count rate of a scintillation camera under the conditions of routine clinical imaging with a particular radionuclide.

Procedure

(1) Mount the collimator to be used on the detector head. Turn the head to face vertically downward.
(2) Adjust the position of the detector head so that it is over the centre of the patient bed.
(3) Set all controls to the default settings for the radionuclide concerned (see Section 2.4.2).
(4) Perform a count over an interval of 100 s with no radiation sources in the vicinity. Record the background count rate.

Interpretation of results

The value of the background count rate should be compared with the reference value and with recent values to identify any changes or trends. A significant increase in background count rate may indicate radioactive contamination of the instrument or its surroundings, or increased environmental radiation from local sources. Alternatively, it may indicate electrical noise. Radioactive contamination may be on the instrument itself, particularly on the collimator face, on the patient bed, on the floor, in the waste bin or even on the person carrying out the test. Local radiation sources may include patients to whom radioactive materials have been administered.

If an abnormal result is recorded, the test should be repeated after checks to make sure there are no nearby radiation sources and that there is no radioactive contamination of the instrument or its surroundings. If contamination is detected, the area involved should be cleaned. Studies may then usually proceed if the detector and collimator are not directly contaminated.

An unaccountably high background count rate should be monitored over a period of days to see whether it decreases with radioactive decay or whether it persists. An image should be acquired to help determine its origin. The cause may be an electrical fault.

For examples, see Section 6.2 of Ref. [3].

Limits of acceptability

A change in background count rate by more than 20% from the reference value would call for further investigation.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

2.4.5. Check of film handling and processing

Purpose of test

To check the adequacy of film handling and processing for a scintillation camera.

Procedure

(1) Visually inspect the flood field image obtained in the test in Section 2.4.3: Check of flood field uniformity and sensitivity, for lack of clarity, irregular background, streaks, smudges, signs of static discharge or any other defects such as may be due to inadequate film handling or processing techniques.
(2) Check the temperature of the film developer.

Observations

The darkroom should be free from light leaks and fitted with proper safety lights. The humidity must be sufficiently high to prevent static discharges that may occur when separating boxed film or loading or unloading film cassettes.

The chemicals used in processing must be replenished regularly and kept at a controlled temperature to assure consistent film density. Inadequate mixing of the developer will result in streaking or smudging.

Interpretation of results

Any inadequacies in film handling or processing techniques revealed by defects in the image should be rectified forthwith.

For image examples, see Section 7.2 of Ref. [3].

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3. WHOLE BODY SCANNING SYSTEMS

3.1. INTRODUCTION

Scintillation cameras with a whole body scanning option can scan the length of a patient in one or two passes. Two passes are necessary if the lateral field of view of the detector cannot span the width of the patient body completely, i.e. if the lateral field of view is less than approximately 50 cm.

The scan is performed by either moving the scintillation camera or by moving the patient bed such that the total length of the patient passes the field of view of the detector(s). If two passes are required, both halves of the patient are scanned separately, switching automatically from one half scan to the next.

A scintillation event recorded by the scintilation camera generates an X and a Y coordinate signal. A position sensor is used to measure the current position of the scanning device (the position of the moving scintillation camera or the patient bed). The Y coordinate is defined as the longitudinal direction, i.e. the direction of the scanning motion. An electronic circuit adds the current position of the scanning device to the Y coordinate of the event within the actual scanning device, thus producing a coordinate value that corresponds to the position of the event along the long axis of the patient. Both the X and the modified Y coordinates are used to build up the whole body scan, an image of the distribution of the radiotracer within the whole body, analogous to the formation of the image in the conventional scintillation camera.

When a scan starts or stops, first an electronic window opens with a speed equal to the speed of the subsequent mechanical motion. Then, the gantry starts moving. When the mechanical motion comes to an end, the electronic window gradually closes again. This ensures that the count densities at the longitudinal borders of the scanned area are the same as in the central parts of the scan.

3.2. TEST SCHEDULE AND DESCRIPTION OF TESTS

3.2.1. Test of system spatial resolution without scatter

3.2.1.1. Method 1 (to be used as an acceptance test at acceptance and after major repairs or adjustments)

Purpose of test

System spatial resolution without scatter is measured parallel and perpendicular to the direction of motion and expressed as FWHM and FWTM of the line spread function.

Materials

- Line sources containing about 200 MBq (5 mCi) of ^{99m}Tc in each line source. The length of each source is equal to the width of the scanned field of view perpendicular to the direction of motion.
- The activity of the sources is adjusted to yield a count rate between 10 000 and 20 000 counts/s.

Procedure

(1) Mount the collimator to be tested on the detector head.
(2) Position one line source at the centre of the scanned field of view, perpendicular to the direction of motion to within 1 mm. The second source shall be placed parallel to the first one, at a distance of 100 mm.
(3) Centre the manufacturer's default PHA window on the photopeak.
(4) Set the scan speed to a typical clinically used value.
(5) Choose the acquisition matrix size such that the digital resolution perpendicular to the line source is not less than 0.25 of the FWHM of the system resolution of the collimator being used. The digital resolution parallel to the line sources should not be less than 25 mm and no more than 30 mm.
(6) Perform scans both above and below the table using digital acquisition. The camera is positioned at a distance of 100 mm from the sources to the face of the collimator.
(7) Reposition the line sources parallel to the direction of motion, with one line source at the centre of the scanned field of view to within 1 mm. The second source is placed parallel to the first one, at a distance of 100 mm.
(8) Repeat steps (3)–(6).

Data analysis

(1) Generate a 25–30 mm thick profile perpendicular to the line sources.
(2) Determine the separation of the peaks in pixels.
(3) Calculate an average value in millimetres per pixel from the separation of the maxima and the known line spacing. This calculation is done separately, parallel and perpendicular to the direction of the motion.
(4) For the central capillary tube, calculate the FWHM and the FWTM, using linear interpolation, for each segment. A segment is a 25–30 mm thick profile. Average the values of the FWHM and the FWTM separately for the tubes parallel and perpendicular to the direction of motion.

Observations

This test is intended to be performed as an acceptance test and after major repairs or adjustments. It follows the NEMA definition and method set forth in NU 1 1994 [6] and NU 1 2001 [8].

Most problems associated with whole body scanning systems will affect the spatial resolution. Resolutions parallel and perpendicular to the direction of motion are reported separately, as they are controlled by different mechanisms. The resolution parallel to the direction of motion, measured with line sources perpendicular to that direction, is affected by the motion control, camera scale calibration and collimator quality. The perpendicular resolution, measured with line sources parallel to the direction of motion, is affected primarily by the mechanical alignment of the camera and the table and by collimator resolution.

Interpretation of results

The calculated average values of the FWHM and the FWTM for each collimator should be compared with the manufacturer's specifications, reported by the manufacturer as a class standard.

Limits of acceptability

If a value of the FWHM or the FWTM is obtained that is 10% or more above the manufacturer's worst case value for the collimator in question, corrective action should be initiated through the manufacturer's representative.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.1.2. Method 2 (to be used for routine testing, for dual path systems and to check for longitudinal misalignment)

Purpose of test

To compare visually the scanning mode spatial resolution with the resolution of static imaging.

Materials

- Uniform flood source.
- A spatial resolution test pattern such as an OHTP or a quadrant bar pattern.

Procedure

(1) Place resolution test pattern and flood source on the middle of the bed. If a quadrant bar phantom is used, orient at an angle of 45° to the scan direction.
(2) Perform scan with distance from resolution test pattern as close as possible and use speed setting as typical for whole body bone scan.
(3) Acquire static image of resolution test pattern using same distance and total counts.
(4) Perform two more scans and adjust scan limits such that start and stop regions of the largest possible field of view are included completely in one of the images.
(5) For a dual head system, perform one more scan for the second head with phantom in central position.

Data analysis

Compare images visually.

Observations

This test should be performed as a reference test at half-yearly intervals.

If two passes are necessary for the whole body scan, the scanned image of the phantom can also be used as a sensitive indicator to check the longitudinal alignment of the two halves of the scan.

The original position signals undergo a transformation that allocates the actual positions to the positions within the scan matrix. The transformation of the longitudinal coordinate in particular depends on the position sensor of the scanning device. It may either monitor the position indirectly by counting the impulses of the stepping motor moving the bed or the gantry, or it may monitor the position of the scanning device directly via independent sensors. Vibrations, mechanical resistance along the path of the scanning device, etc., produce deterioration of the longitudinal resolution and may affect, to a lesser extent, the lateral resolution.

Interpretation of results

Each image obtained by scanning should show the same resolution as the static image. The overall shape of the image should represent the shape of the object (see Fig. 37).

For further image examples, see Sections 4.2 and 4.3 of Ref. [3].

Limits of acceptability

If the image of the scan is visibly worse than the static image, or the image is distorted, corrective action should be initiated through the manufacturer's representative.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.2. Test of scan speed

3.2.2.1. Method 1 (to be used if the position encoder gives the actual position of the scanning device)

Purpose of test

This test checks the correct speed of the electronic window and non-uniformities of scan speed.

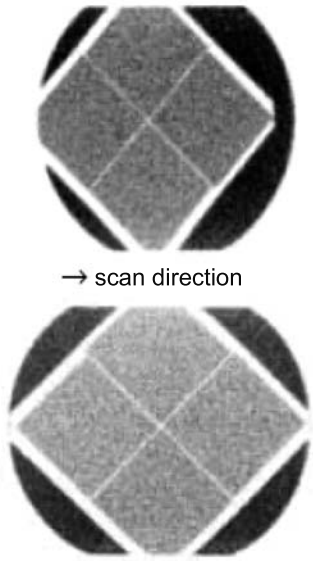

FIG. 37. Whole body scan — misregistration of position signals. Scintillation camera with whole body scanning option, LEHR collimator, ^{57}Co, 20% energy window. The first whole body scan (top) shows a diamond shaped bar pattern instead of the expected square pattern, owing to a misregistration of the position signals during scanning. This problem was solved after service. The repeat scan (bottom) was acceptable (see Ref. [3]).

Materials

- Flood source with ^{99m}Tc or ^{57}Co covering the scan window completely.
- Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
- Count rate should be below 20 000 counts/s.

If no flood source is available, a line source filled uniformly with ^{99m}Tc can be used. The length of the source should be such that the source, when mounted perpendicular to the direction of the scan on the collimator, covers the lateral field of view completely.

Procedure

(1) Fix the flood source or line source to the collimator of the scintillation camera.
(2) Perform the whole body scan.
(3) Repeat for different scan speeds.

Data analysis

(1) Visually inspect scans for intensity variations.
(2) Draw longitudinal profile through middle of digital image.
(3) For a two pass scan, draw two longitudinal profiles, one through the middle of each pass.

Observations

This test is intended to be performed as a reference test at half-yearly intervals.

During a scan, the scanning device is used to transform the location of an event from the coordinate system of the scintillation camera into the coordinate system of the whole body. Misallocations are caused by differences between the actual and the measured position of the scanning device. Furthermore, non-uniformities of speed cause artefacts in count densities in different parts of the image. Also, the electronic window used to adjust the intensities at the start and stop regions of a scan should open and close with the same speed as that of the mechanical motion.

Scan speed is tested only if the position encoder gives the actual position of the scanning device. If the position encoder determines position indirectly, for example, from the number of steps of a stepping motor, use an alternative test of scan speed (method 2).

The test may also be used for simultaneous acquisition with dual head systems. However, the image of the lower detector will show shadow effects due to absorption of radiation by the patient bed.

Interpretation of results

The intensity profile should be flat. Fluctuations in profile intensity of more than 5% should be followed up by inspection and clearing of any obstruction from the scan path. The border regions of the profile containing the transition areas between mechanical scan and electronic sliding window should have the same intensity as the rest of the profile.

For image examples, see Section 4.1 of Ref. [3].

Limits of acceptability

If fluctuations in profile intensity continue to exceed 5%, corrective action should be initiated through the manufacturer's representative. If the border regions increase or decrease linearly, the electronic opening or closing speed of the window needs adjustment.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.2.2. Method 2 (to be used if the position of the scanning device is obtained indirectly, e.g. by counting the steps of the stepping motor driving the scan)

Materials

Ten point sources containing approximately 20 MBq (0.5 mCi) of ^{99m}Tc each.

Procedure

(1) Measure activity of each point source accurately with a radionuclide (dose) calibrator.
(2) Position point sources at known distances equally spaced along the longitudinal axis of bed.
(3) Adjust scan limits such that the electronic window will include the sources at the start and the end of the scan.
(4) Perform scans at different speeds.

Data analysis

(1) Determine intensity and position of each point source in the digital image.
(2) Calculate relative intensity of each point source, using the central source as reference and normalizing to equal activities.
(3) If total duration of the scan exceeds 30 min, perform additional decay corrections.

Observations

This test is intended to be performed as a reference test at half-yearly intervals.

For single pass scanners, first position point sources parallel to the longitudinal axis centred in one half of scanned field of view, perform scans, then position point sources in other half of field of view. Repeat scans. For dual head cameras, the scans should match in scale.

Interpretation of results

Variations of relative intensities, after correction for activity and decay, should not exceed 5%. Otherwise, follow up as indicated above.

Limits of acceptability

If fluctuations in profile intensity continue to exceed 5%, corrective action should be initiated through the manufacturer's representative. If the border regions increase or decrease linearly, the electronic opening or closing speed of the window needs adjustment.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.3. Test of exposure time corrections

Purpose of test

To test that there is no reduction in the count density at the lateral edges of the whole body image.

Materials

- Flood source with ^{99m}Tc or ^{57}Co covering scan window completely.
- Centre the manufacturer's default PHA window on the photopeak (see Section 2.4.2).
- Count rate should be below 20 000 counts/s.

Procedure

(1) Fix flood source to collimator of scintillation camera.
(2) Perform whole body scan.
(3) Repeat for different scan speeds.

Data analysis

(1) Visually inspect scans for intensity variations.
(2) Draw lateral profile through middle of digital image.

Observations

This test is intended to be performed as a reference test at half-yearly intervals.

It is normal that the noise of the uniformity image apparently increases towards the lateral borders. This is due to amplification by multiplication. For dual head systems, each head has to be tested separately in order to avoid the shadow effects of the patient bed.

For scintillation cameras with a circular field of view, the electronic scanning window may not be rectangular. This may also apply to rectangular cameras with corners cut off. Since the path lengths for different lateral positions then vary, a correction to adjust the intensity in the resulting image is applied, essentially consisting of multiplying the intensity in the image at a given lateral coordinate by the inverse of the chord length. For analogue recording, this is done by an intensity modulation, and for a digital image, the correction is performed by digital multiplication of each pixel with the appropriate correction factor.

Interpretation of results

The images should show uniform intensity. The profiles should be flat and horizontal.

Limits of acceptability

Service is required if the images show visible non-uniformities or if the difference between the maximum and the minimum count of the profile exceeds 20%.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.4. Test of scan path separation of dual path (two pass) scanners

Purpose of test

To visualize and assess the significance of an image overlap from a dual pass scanner.

Materials

Flood source as used in test in Section 3.2.3: Test of exposure time corrections.

Procedure

Perform scan as in test in Section 3.2.3: Test of exposure time corrections.

Data analysis

Visually inspect the whole body image for a gap or overlap between the two scan paths.

Observations

This test is intended to be performed as a reference test at half-yearly intervals.

The field of view of many scintillation cameras does not span the width of the body of a patient. The patient is therefore scanned two times in succession, with each scan covering one half of the body (two pass scanner). The images are electronically combined such that they ideally give the impression of one scan. Electronic imperfections cause a lateral dissociation of both scans, producing either a narrow gap between the two images or a strip of increased count density caused by an overlap of the two scans.

Interpretation of results

Either an excessive gap or overlap between the two scan paths that form the whole body image indicates a misalignment of the scan paths.

For image examples, see Section 4.3 of Ref. [3].

Limits of acceptability

An electronic adjustment is required by the manufacturer's representative if the images overlap or are separated by a gap of greater than 5% of the width of the lateral dimension.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

3.2.5. Test of longitudinal alignment of dual path scanners (two pass scanners)

Purpose of test

To test that the two scan paths are aligned in the longitudinal scan direction.

Materials

- Flood source.
- A spatial resolution test pattern, such as the quadrant bar pattern (preferred) or the OHTP.

Procedure

(1) Place the resolution pattern and flood source on the middle of the bed. If a quadrant bar phantom is used, orient it at an angle of 45° to scan the direction.
(2) Perform a scan with the collimator as close as possible to the resolution pattern and use a scan speed setting typical for a whole body bone scan.

Data analysis

Compare images visually and check for discontinuity in the image of the resolution pattern.

Observations

This test is intended to be performed as a reference test at half-yearly intervals and should be performed as an adjunct to the test of system spatial resolution (see Section 2.3.8 (Method 2)).

If the field of view of a scintillation camera does not span the width of the body of a patient, the patient is scanned two times in succession, each scan covering one half of the body. The images are electronically combined such that they ideally give the impression of one scan. Electronic imperfections may cause a longitudinal dissociation of both scans, producing a misalignment along the long axis of the scans between the two scan images.

Interpretation of results

The scan composed of the half images should show a smooth transition of the resolution pattern between the two scan path images.

Limits of acceptability

If the scan exhibits visible discontinuity, corrective action should be initiated through the manufacturer's representative.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4. SPECT SYSTEMS

4.1. INTRODUCTION

The most basic type of SPECT system comprises a conventional scintillation camera mounted on a special gantry and connected to an appropriate computer system. This type of system enables a series of images acquired around a patient to be reconstructed to give a set of transaxial images, similar to those obtained by X ray CT, which constitute a 3-D image of that part of the patient being scanned.

While SPECT imaging is extensively used in nuclear medicine imaging, special attention is needed with respect to quality control. SPECT systems will not produce adequate results unless great care is taken with the performance and set-up of both the scintillation camera and all the other component parts of the system. There are additional requirements for double (or multiple) head systems, which are further considered in Section 5.

4.1.1. Basic principles

The basic principle of a SPECT system dependent on the rotating camera concept is that a series of planar images are collected while the camera is rotated through either 180° or 360° around the patient. These planar images are called projection images and are used to create transaxial slice images by filtered backprojection of the data into the transaxial plane. Figure 38 is a diagram of such a system with various axes and, in particular, with the axis of rotation indicated and identified (see Section 4.1.2). Each row of pixels across the projection image gives a projection line, a profile of counts for a common Y value in that image. The counts in these projection lines may be backprojected at the appropriate angle across the transaxial plane, which would result in a first order approximation of the data that gave rise to the set of projection images.

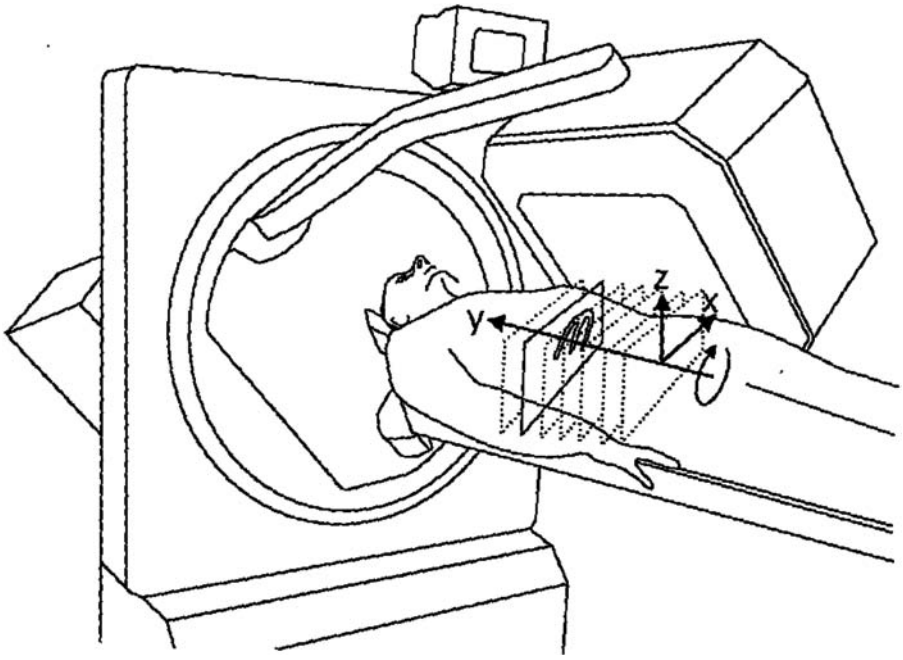

FIG. 38. Diagram of a SPECT system showing the axis of rotation (see Ref. [17]).

4.1.1.1. Some important definitions

Projection ray: A line perpendicular to a projection, going through the volume which is being examined or reconstructed, such that the sum of data along the ray (or weighted sum) is equal to the value of that point on the projection from which the ray was cast.

Projection line: The set of values along one line, at some given angle, the value at each point corresponding to the sum of values along the projection ray cast from that point

Projection image: A 2-D projection (e.g. a function of X and Y), at some given angle, comprising a set of parallel projection lines.

Sinogram: An image formed from a set of projection lines, for some fixed offset (e.g. a constant value of X), for angles around the object to be reconstructed (see Fig. 39).

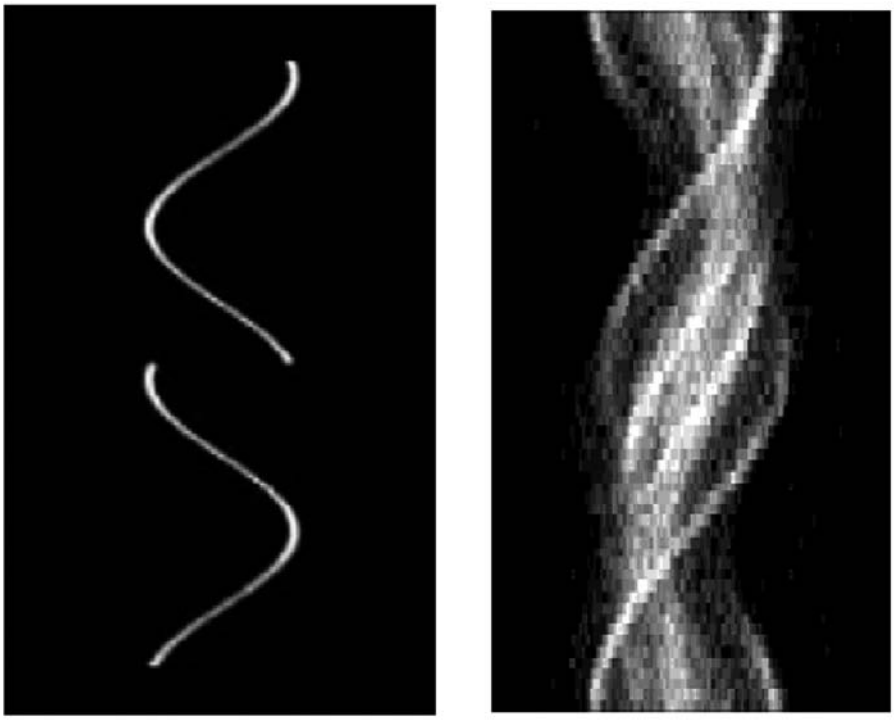

FIG. 39. Example sinograms. The horizontal direction is the X axis and the vertical direction is the projection angle. Left: Sinogram of a single point source, placed off axis and imaged by a dual head camera, with each head rotating through 360°. Right: Sinogram of a bone scan imaged by a single head camera rotating through 360°.

The filtering, backprojection, attenuation and other corrections of SPECT data are performed using a digital computer. The original data are stored as a series of projection images and, depending upon the operator's commands, can be reconstructed to give one or a number of transaxial slices after appropriate filtering and processing. Once the transaxial slices have been created, it is possible to use the same data to create sagittal, coronal, or oblique slices through the object, essentially by reordering the data as shown in Fig. 40.

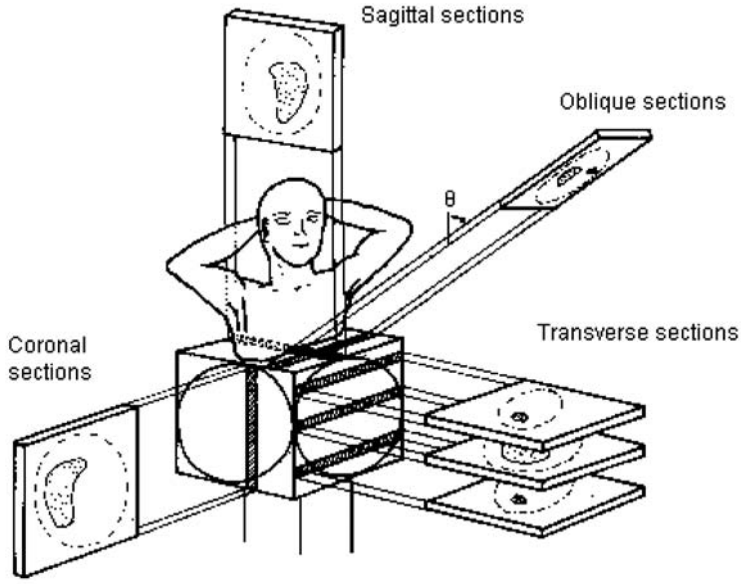

FIG. 40. Transaxial, sagittal and coronal planes.

It is possible to reconstruct transaxial images that are far from optimum, because of the interdependence of performance of a scintillation camera, its motion and the reconstruction algorithm. While the degradation of planar images is usually easy to recognize in clinical images obtained from a poorly functioning or poorly adjusted scintillation camera, this is not usually the case in SPECT, where it is quite possible to produce poor images, or images containing artefacts, without the situation being properly recognized.

This section gives details of some basic procedures for testing such systems, but is not fully comprehensive. In addition to understanding how to use a scintillation camera, it is assumed that the user of a SPECT system is competent in using a computer. Some of the tests require rather more expertise than those described in previous sections. As an aid, a section giving definitions of some special terms is included.

4.1.2. Special terms

To aid in the description of the tests in this section, a number of special terms will be defined.

(1) The detector plane is the front surface of the scintillation camera, and is usually considered to be the front surface of the crystal, not the front face of the collimator. The X axis is defined as illustrated in Fig. 38, perpendicular to the axis of rotation, with the Y axis being parallel to the axis of rotation.

(2) The axis of rotation is that axis (as illustrated in Fig. 38) about which the camera rotates. The axis of rotation is usually, to a good approximation (but typically not exactly), horizontal. In addition, the line that may be drawn along the central axis of the bed is usually not exactly parallel with the axis of rotation. The axis of rotation defines the centre of rotation (a point) as the intercept between the axis of rotation (a line) and a perpendicular drawn from the centre of the detector plane when the detector is parallel to the axis. A centre of rotation offset exists when the perpendicular drawn from the detector plane does not intercept the axis of rotation. A number of systems now determine the centre of rotation offset for every angle. In addition, the centre of rotation offset may be a function of the type of orbit used, in particular for non-circular orbits. The septa of parallel hole collimators may not be exactly aligned perpendicular to the detector face and thus may also introduce a centre of rotation offset.

(3) The home position is the position to which the system returns, often automatically, and is where the angle of rotation is considered to be zero.

(4) The angle of rotation is the angle between the perpendicular line dropped from the centre detector plane when the camera is at some given position and the same line when the camera is at its zero or home position, typically when the camera is horizontal.

(5) The angle of tilt or head tilt is the angle between the detector plane and the axis of rotation, measured along the axis of rotation. It should normally be 0° when the system is correctly set up and remain at 0° for all angles of rotation.

(6) A projection image is a conventional planar image obtained for some angle of rotation. It is a matrix of size $n \times n$, typically 64×64 or 128×128, being the raw acquisition matrix size. A projection line is defined as a line in the projection image, being the set of pixels having a common Y value.

(7) Tomographic acquisition is the process of collecting a set of all projection images for each of the angular sample positions within the total angle of rotation, typically either 180° or 360°.

(8) The transaxial plane is the plane perpendicular to the axis of rotation intersecting with the detector plane along a line, corresponding to a particular projection line.

(9) The sagittal plane is the plane parallel to the axis of rotation and passing through the patient anterior–posterior.
(10) The coronal plane is the plane parallel to the axis of rotation passing through the patient from left to right. Transaxial, sagittal and coronal planes are all orthogonal, as illustrated in Fig. 40.
(11) Oblique planes are planes constructed from the transaxial, sagittal and/or coronal planes and lying at some angle with respect to any one or a combination of those planes.
(12) An image, termed a sinogram, can be generated by choosing some minimum and maximum values for Y for the set of projection images. The projection lines so defined in each image may then be summed to generate one single line of data for each projection image. These lines are then placed in successive order in a new matrix. Each line is placed in a position corresponding to the angle at which it was obtained. This image is generally termed a sinogram and is illustrated in Fig. 39. The term sinogram derives from the fact that a point source placed off axis will project into a sine wave image in the set of projection profiles used to create this new matrix.
(13) The reconstruction thickness is the number of pixels along the Y axis of the raw projection images, which were summed before reconstruction of one single transaxial slice. While the reconstruction thickness is a number of pixels, the slice thickness is the effective resolution in millimetres of a transaxial slice along the Z axis (see (14)). One suitable definition of the slice thickness is the FWHM of the profile through a point source along the Z axis.
(14) After reconstruction of a set of sinograms (each corresponding to one transaxial slice), an axis is defined parallel with the axis of rotation (perpendicular with the slice plane) which is normally termed the Z axis. This is actually parallel with the Y axis of the raw projection images. Thus, it is important to distinguish the X and Y axes in the raw projection images and the X and Y axes in a reconstructed transaxial slice.
(15) During step and shoot acquisition, the camera moves to some angle of rotation, stops, collects a projection image and then rotates to the next angular position. During continuous rotation, the system does not stop and projection images are normally formed over small angular increments while the camera is rotating. Uniform angular increments are required for step and shoot acquisition while continuous rotation requires that the system either maintains a uniform angular velocity, or corrects for variations in speed.

(16) The angular increment for a step and shoot system is the difference in the angle of rotation for successive projection image positions. The total number of angles or views is the total number of angular positions for which projection images are obtained. The total angle of rotation is the angle through which the system rotates when collecting data, normally 360°, although often, in cardiac tomography, 180° is employed. The camera can rotate either clockwise or anticlockwise, when the system is observed along the axis of rotation towards the gantry.

(17) The radius of rotation is the perpendicular distance between the detector plane (the front surface of the crystal or collimator) and the axis of rotation and should be constant for a so-called circular orbit. For practical purposes, it is often necessary to measure this distance from the front face of the collimator and then correct for the thickness of the collimator.

(18) A non-circular orbit is obtained by moving either the detector head, the gantry, the bed, or a combination thereof, during rotation such that the detector plane is not always at a constant distance from the axis of rotation. Some other, well-defined orbit, such as an ellipse (see Fig. 41) or patient contouring 'peanut', is described. The goal of such orbits is to reduce the distance between the detector and patient and improve the spatial resolution in the tomographic plane.

(19) The most commonly employed type of reconstruction is termed filtered backprojection. The term reconstruction filter is employed for the filter used before backprojection of the data in this type of reconstruction. The ramp filter is the sharpest filter normally employed and implies that no extra smoothing takes place during reconstruction. Many different types

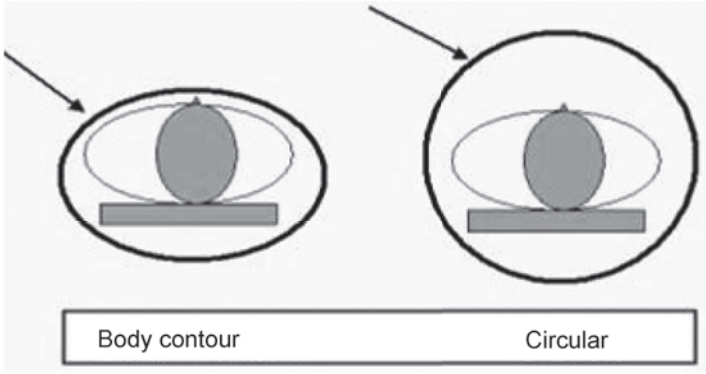

FIG. 41. Circular and non-circular orbits illustrated by the solid contours (indicated by arrows). Note that in body contour (or elliptical orbit), the detector is closer (on average) to the patient, compared with circular orbits.

of filter (window functions) may be employed, for example, the Shepp-Logan, the Hamming, the Hann and the Butterworth. In general, these are smoothing filters applied additionally to the ramp filter. However, more recently, iterative reconstruction methods are being employed. The use of more complex iterative reconstruction techniques does not affect many of the parameters that need to be checked as part of the quality assurance programme for SPECT. For example, the need for an accurate centre of rotation correction will still exist. The iterative reconstruction methods will, however, influence the signal-to-noise ratio found in the final images and may slightly affect the resolution and contrast of detected objects. These methods may have a considerable influence on the uniformity of reconstruction of a uniform object, in particular with respect to the attenuation correction.

(20) Attenuation correction is that part of the reconstruction process whereby those counts (events) assumed lost due to attenuation within the object are restored. This correction may be performed prior to, during or after the main reconstruction operation and usually requires knowledge of the distribution of attenuating tissue, for example, of the outside surface of the patient (the body contour) and not just of the distribution of the activity within the body. Depending on the system, various methods have been employed for determining the body contour of the patient: using external markers; performing a radionuclide transmission study; conducting an X ray CT scan on a hybrid SPECT/CT system; or using an additional lower energy acquisition window to detect scattered events. A transmission study or an X ray CT scan can also determine the attenuation coefficient of different tissues within the body and an attenuation correction considering these differences can be performed.

(21) Some systems also perform a scatter correction to eliminate the effects resulting from scattered photons registered within the photopeak window. Such a scatter correction may sometimes be performed by software alone, by collecting data from a different (lower) energy window or by use of asymmetric or multiple energy windows.

(22) The attenuation correction coefficient is the value used in the attenuation correction process. It is system dependent and depends in particular on whether scatter correction has been performed. When only a body contour has been determined, typically only a single value is used. Where a full transmission image has been acquired, various tissue dependent values may be employed, normally as part of an iterative attenuation correction procedure.

(23) Fan beam reconstruction and cone beam reconstruction are the modified reconstruction procedures employed when a converging collimator is used instead of a parallel hole collimator. There are various types of converging collimator that have been tried, in particular fan beam and cone beam collimators. The use of such techniques requires considerable extra care in setting up the SPECT system and its associated software for obtaining a clinical SPECT image.

(24) The partial volume effect is the loss of signal (normally observed as a loss of contrast) that occurs when an object partially occupies the sensitive volume of an imaging instrument (usually in space, but also in time), i.e. slice thickness effects and the point spread function effects. Movement effects are also included. Partial volume effect causes an apparent loss of contrast of small objects. This effect will depend not only on the size of the objects but to some extent on the shape of the object.

4.1.3. Components of the system

The two principal components of the system are a conventional scintillation camera system and a computer to which it is interfaced. However, in addition, there will be:

(1) *Patient bed (pallet or couch)*. Tomographic beds are normally specially designed and differ considerably from conventional scintillation camera beds. They are much narrower, so that the camera can rotate with a small radius of rotation. They are made of special material in order to minimize attenuation. They are often designed so that the long axis of the bed can be aligned with the axis of rotation. Finally, the bed height and sometimes the horizontal bed position (shifting the bed laterally) can be controlled either manually or under motorized control. In particular for brain SPECT, there should be a support provided for the patient's head that permits the radius of rotation to be reduced to a minimum and permits the patient's head to be tilted at the desired angle.

(2) *Gantry*. Tomographic gantries are designed to rotate the camera head(s) about the patient. Often, they are mechanically rather massive and in many cases move under the control of a microprocessor interfaced to the main computer. This controller may comprise just a rotation controller or a much more complex system running more or less autonomously.

(3) *Rotation controller*. This device controls the rotation of the camera around the axis of rotation. It is normally interfaced to the main computer. For a step and shoot system this interface controls the angular increment between successive projection images. For a continuous

rotation system, this interface controls the speed of rotation. In addition, it often permits the camera to be returned to its home position. Such a controller also controls, where appropriate, the lateral position of the gantry and any other mechanical motions under the control of the system. In many cases, on modern systems, the rotation controller not only determines the radius of rotation of a system for all angles, but also the exact position of the head for each individual angle so that various known orbits can be executed.

(4) *Emergency stop and other patient safety devices.* All tomographic systems have (or should have) an emergency stop button to prevent or abort motion that might injure a patient. Some systems have, in addition, a patient safety device, such as a pressure sensitive pad on the collimator face, that stops motion or automatically moves the detector away from the patient when the system touches the patient couch or the patient.

(5) *Position readout devices.* These are devices whereby information such as angular position and radius of rotation are displayed. They vary considerably from system to system. In particular, most systems have some method for checking the tilt of the head, e.g. a spirit level attached to the camera head. Such position readout devices should not be relied upon for accuracy, in particular with respect to the centring of the system.

Performance characteristics

Many of the performance characteristics of a SPECT system are similar or identical to those described in previous sections and will not be repeated. A series of parameters describing the conventional performance of the camera are defined elsewhere, for example:

(1) Energy resolution (Section 2.1.6.2);
(2) Flood field uniformity (Section 2.1.6.3);
(3) Spatial distortion (spatial non-linearity) (Section 2.1.6.4);
(4) Differential ADC linearity (Section 6.1.2.3);
(5) Integral ADC linearity (Section 6.1.2.3);
(6) Spatial resolution (Section 2.1.6.1);
(7) Count rate response (Section 2.1.6.6).

In most cases, a tomographic system is much more sensitive to poor performance with respect to these parameters than for conventional imaging. This applies in particular to poor uniformity, as discussed in Section 4.1.3.4: Tomographic uniformity. Energy resolution is also of special importance. Poor energy resolution can cause, for example, loss of contrast (see Section 4.1.3.3).

Furthermore, in tomography, many performance characteristics do not have a single (unique) value. For example, tomographic spatial resolution (see Section 4.1.3.5) will vary as a function of position within a transaxial slice. It will also vary considerably as a function of the radius of rotation (as defined in Section 4.1.2) and with the reconstruction technique employed.

In addition to the parameters describing conventional (scintillation camera) performance, it is desirable to define a number of other parameters in order to describe tomographic performance. These include:

(1) Slice thickness;
(2) Tomographic signal-to-noise ratio;
(3) Tomographic contrast;
(4) Tomographic uniformity;
(5) Tomographic resolution;
(6) Linearity of tomographic response;
(7) Quantitative accuracy in tomography;
(8) Precision of estimation of the centre of rotation;
(9) Tomographic sensitivity — slice and volume.

4.1.3.1. Slice thickness

The slice thickness of a transaxial slice is defined for the purposes of this publication as the FWHM, measured from the response of the system to a point source placed at some known radial distance from the axis of rotation, along a line parallel to the axis of rotation. It is not constant with respect to position within the transaxial slice. Important values are the slice thickness at the centre (i.e. along the axis of rotation) and at some known radial distance, e.g. 10 cm from the axis of rotation.

The importance of the slice thickness is that it describes the spatial resolution of the system along the Z axis for that reconstruction. A reconstruction thickness of about 3 pixels results in a resolution similar to the conventional spatial resolution of the scintillation camera at the corresponding depth in tissue.

The reconstruction thickness (see Section 4.1.2) is quite different; it is the number of projection lines used to reconstruct one transaxial slice. The Z resolution of the system does *not* correspond to the size of a pixel multiplied by the reconstruction thickness used in the reconstruction. For example, if the pixel size is 3 mm and the reconstruction thickness is 1 pixel, the slice thickness is nevertheless likely to be of the order of 10–20 mm. However, when the reconstruction thickness becomes much larger, for example >5 pixels for 3 mm pixels, the reconstruction thickness starts to dominate and largely determines

the slice thickness. Thus, the slice thickness is dependent on the reconstruction thickness. Conventionally, slice thickness is measured for a reconstruction thickness of 1 pixel such that it is nearly independent of reconstruction thickness.

Variations in slice thickness, for example, across the transaxial field of view, cause variations in the observed response of the system (pixel contents in the reconstructed image) and can cause considerable difficulties with respect to quantification.

4.1.3.2. Tomographic signal-to-noise ratio

In planar imaging, the signal-to-noise ratio at a point in the image is well understood. The number of counts in a pixel (or for the whole field of view) behaves according to Poisson statistics and therefore has a variance equal to the number of counts in that pixel (or the sum of counts in the whole image). As a result of the tomographic reconstruction process, this is not true for the signal-to-noise ratio in tomography. The noise is no longer Poissonian, it does not have a uniform (white) power spectrum and it depends on a number of parameters: counts acquired, the distribution of the counts, the reconstruction process and other components of the reconstruction algorithm such as scatter and attenuation correction. In particular, if the standard deviation of a number of pixels is measured, for a uniform area, the standard deviation will depend on the size and shape of the area in which it is measured. In general, the greater the smoothing imposed by the windowing filter in the tomographic reconstruction (for filtered backprojection), the smaller the value of the standard deviation will appear to be and the better the apparent signal-to-noise ratio. In addition, the signal level (raw values of the pixels) may depend on, and vary considerably with, the characteristics of the reconstruction algorithm itself, and in particular, the gain of the filters and their values for zero frequency. Thus, in practice, the raw signal-to-noise ratio is not well behaved and most workers in the field have preferred using a measure of tomographic contrast, or tomographic contrast-to-noise ratio, as described in the following section.

4.1.3.3. Tomographic contrast

Tomographic contrast is an important indicator of how well a system is performing with respect to detection of small lesions. It is defined using the following procedure.

Place a sphere of some known size, but with a size much greater than the spatial resolution of the system in order to minimize partial volume effects, within a volume containing a uniform concentration of activity. After

reconstruction, estimate the value (V_{bgd}) of pixels in the reconstructed image in the neighbourhood of the sphere, but outside the region corresponding to the sphere. Estimate also the value of pixels within the region corresponding to the sphere (V_{sph}). Contrast for this size lesion may then be calculated as:

$$\text{Contrast} = \frac{V_{sph} - V_{bgd}}{V_{sph} + V_{bgd}} \tag{15}$$

Other possible definitions exist and have been employed. However, the fundamental concept is to estimate the capability of the system to detect a known change in activity concentration for a given size of (spherical) object. In particular, contrast is very dependent on the size of the lesion used to estimate it.

Tomographic contrast is important in that it determines the detectability of small lesions. It is affected by many different properties of the system, in particular energy resolution, the contribution of scatter and the reconstruction filter. Tomographic contrast decreases as the size of the object becomes comparable to, or smaller than, the spatial resolution of the system, or when the object only partially fills the reconstruction slice, effects which are termed the point spread function effect and partial volume effect (see Section 4.1.2).

The contrast-to-noise ratio of such a lesion is the ratio of contrast to the noise in the tomographic image, normally expressed solely as a number when noise is expressed in the same units as contrast. The noise, as indicated above, is the standard deviation in a region where conventionally a region of similar size to the lesion being detected is employed, close to the position of the lesion.

4.1.3.4. Tomographic uniformity

Tomographic uniformity is the uniformity of the reconstruction of a slice through a uniform distribution of activity. At present, there is no consensus as to how a number, or parameter, corresponding to the NEMA uniformity index for planar images [6, 8], may be determined from a tomographic image, although there have been suggestions of extending the NEMA definitions for integral and differential uniformity and applying them to the full set of projection images. There are two main components to tomographic non-uniformity.

In the first, lack of tomographic uniformity may be observed (for a circular orbit) as circular artefacts or rings centred about the point corresponding to the centre of rotation of the system (Fig. 42). In clinical

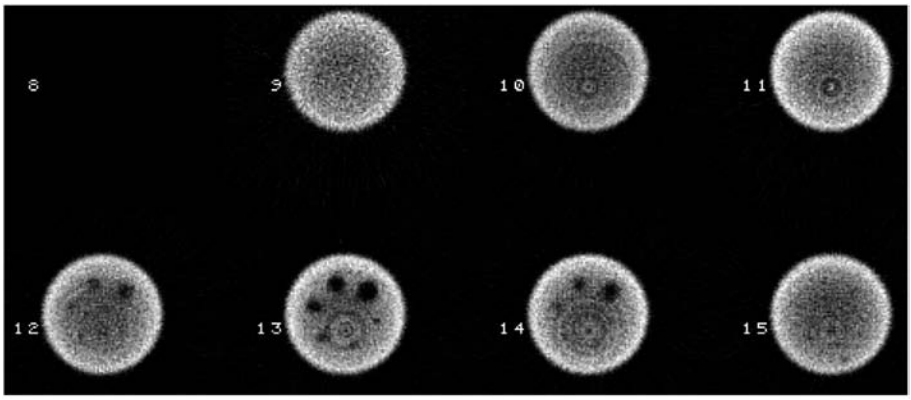

FIG. 42. Seven transverse slices of the sphere section of the Data Spectrum ECT phantom (Jaszcak phantom), imaged with a 360° total angle of rotation and uncorrected for attenuation. Distinct ring artefacts are seen in different slices. Note that the rings are centred around the centre of rotation of the detector and that the phantom is positioned slightly off centre (see Ref. [3]).

studies, incomplete rings (wedges, etc.) may be observed which can often give the impression of corresponding to abnormal activity distributions in the transverse images.

The second major cause of tomographic non-uniformity is attenuation and this results in changes in overall signal response as a function of depth inside the object. This is often seen as a bowl shaped decrease inside the reconstruction.

One method of estimating tomographic non-uniformity is to estimate the 'contrast' of a circular (ring) artefact with respect to the uniform background against which it is observed, for example, by plotting a profile and estimating the depth or height of the 'notch' created in the profile by the artefact.

Planar non-uniformity is considerably amplified by the tomographic reconstruction process. This amplification is an inverse function of the distance from the axis of rotation. Figure 43 shows tomographic non-uniformity and its relationship to planar uniformity and distance from the axis of rotation. It also shows that close to the axis of rotation, tomographic non-uniformity may be many times greater than planar non-uniformity.

4.1.3.5. Tomographic resolution

Tomographic resolution is defined here in terms of the FWHM of the filtered backprojection reconstruction of a point source in a transaxial slice. It

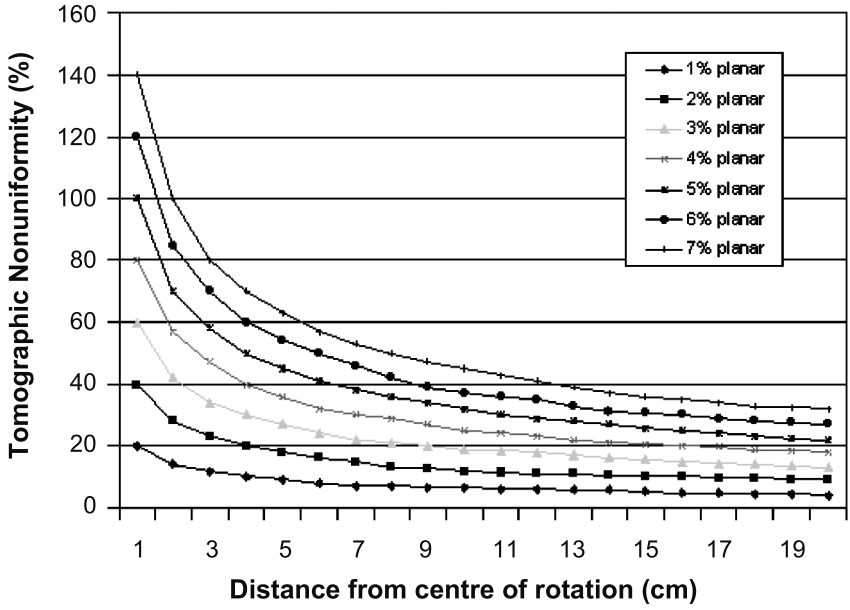

FIG. 43. Amplification of non-uniformity with distance from the centre of rotation for planar uniformity of 1–7%. Note the amplification of the non-uniformity near the centre of rotation.

is measured on the reconstructed image. Tomographic resolution determines the sharpness of the image, as in all types of imaging. However, non-circular point source response functions are often observed in tomography. Resolution is also affected by position, for example, distance from the axis of rotation within the slice plane.

It must be noted that tomographic resolution is likely to be non-isotropic and will vary considerably as a function of position within the tomographic slice. It will also vary considerably as a function of the collimator used, for example, depending on whether a high resolution or a general purpose collimator is employed, the type of acquisition performed, the radius of rotation and whether a circular or non-circular orbit is used. The observed tomographic resolution will also depend on the reconstruction filter being used. It will be worse, the less sharp the filter is. The tests recommended here suggest using a ramp, or, if this is not available, the sharpest filter possible.

4.1.3.6. Linearity of activity estimation

When a series of different objects, each with a known concentration of activity, is simultaneously placed within a tomographic system, the observed reconstructed values for each object may be plotted against the known activity concentrations. Ideally, such a plot should result in a straight line. The linearity of the system is estimated from such a plot. Note that such a plot does *not* estimate the quantitative accuracy of the system. Note also that this is distinct and not the same as spatial distortion, which is also termed 'linearity' by certain authors.

Linearity is one component in the calibration of the system when used for quantitation. Scatter and attenuation may cause considerable deviations away from a linear response. The linearity of the system is important for permitting the comparison of values observed in different regions after tomographic reconstruction, for example, comparing different ROIs within the brain when estimating cerebral blood flow.

Care should be taken when estimating the linearity of the system, since estimates of 'response' may be dependent on position. For example, the use of a circularly symmetrical phantom can give very misleading results, suggesting that the system is much more linear than when a non-symmetrical phantom is employed. A phantom for assessing the linearity of the activity response of a SPECT system can be constructed with a collection of tubes, each with a specific, known activity concentration, in an array of positions inside a circular cylinder of 20 cm diameter or a torso shaped cylinder filled with a uniform solution of water.

4.1.3.7. Quantitative accuracy of the system in tomography

The quantitative accuracy of the system is described by the deviation from the true value involved in estimating the activity concentration at some position in the reconstructed image in absolute terms, i.e. becquerels per millilitre. The value estimated by the system is compared with the actual value at some point within the object. This error should be distinguished from the precision of the system, which is the reproducibility with which a value may be estimated. The linearity of the system is a measure of the relative accuracy of such estimates, that is, the ratio of the estimates of different activity concentrations. If the system is linear, the estimated ratio is the same as the true ratio and a plot of true activity against estimated activity is (over a certain range) a straight line. However, the fact that a system is linear guarantees that it is neither accurate nor precise.

Quantitative accuracy is affected by scatter, attenuation, choice of reconstruction filter, variations in slice thickness, partial volume, point spread function effects (see Section 4.1.2), etc. Thus, estimates of activity made to test quantitative accuracy need to be performed in a variety of different configurations, for example, for different sizes of object, different positions within the object and different amounts of scattering material. In principle, SPECT systems are capable of quantitative accuracy; in practice they are seldom used to determine, quantitatively, activity distributions in absolute units of becquerels per millilitre.

4.1.3.8. Precision of the estimation of the axis of rotation

The perpendicular line passing through the axis of rotation is supposed to pass though the centre (the central axis) of the projection image. For any given projection line in a projection image, the distance between its central point and the point corresponding to the intersection with the perpendicular line dropped from the true centre of rotation may be measured. This distance is termed the centre of rotation offset. It can be estimated as a function of the angle of rotation and is critical in setting up a tomographic system. Errors in the estimation of the centre of rotation cause loss of resolution and, in extreme cases, can cause point sources to be reconstructed as rings (see Fig. 44).

While it is not important that the centre of rotation be exactly in the centre of the image, this offset (the error in the axis of rotation) must be taken into account by the reconstruction software. In fact, the error of estimation of the centre of rotation may well vary to some extent when the system rotates; it is not a single value, but a function of the angle of rotation. Very few systems include such second order effects. For non-circular orbits, similar reasoning applies. In order to perform a reconstruction, the system needs to 'know' where the detector is at any given angle, although it may certainly vary in a more complex manner than for a circular orbit. For extremely large centre of rotation offsets, the tomographic field of view will be reduced.

4.1.3.9. Tomographic sensitivity — slice and volume

The planar sensitivity is normally estimated in terms of number of counts detected for some known activity in a standard phantom. The tomographic sensitivity can be estimated in a similar manner. A known activity concentration is placed into a specified phantom, normally a uniform cylinder. The number of counts detected is then determined for a slice through the object and for the complete object. The result is normally expressed as counts

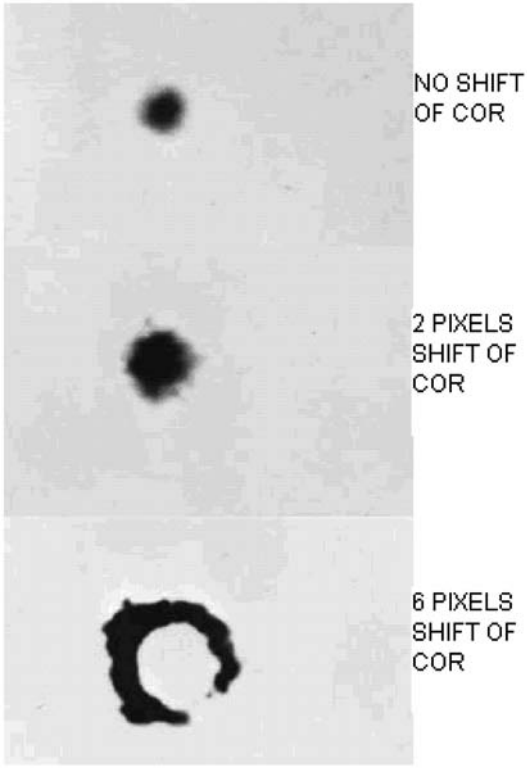

FIG. 44. The effect of reconstruction of a point source with no centre of rotation offset and with 2 pixel and 6 pixel centre of rotation offsets. With a large centre of rotation offset, the transaxial tomographic image of the point source becomes a ring (bottom image).

detected per second per unit of activity concentration. This resulting sensitivity is then typically used to compare different systems having different types of detector or geometry, often with confusing results.

It is the planar sensitivity that can be used to compare single or multiple head SPECT systems. Changes in tomographic sensitivity can be estimated from changes in planar sensitivity. It is complicated to compare one system of a given geometry against another, perhaps very different system by consideration of the number of counts detected for some specified phantom, even though this is included in the NEMA and IEC tests for SPECT systems. Since such a measure is considered to be inappropriate, no test is recommended here.

4.1.4. Operational considerations

In general, the operating conditions are similar to those described in previous sections, in particular with respect to temperature, humidity, etc. However, considerable care needs to be exercised with respect to the mechanical performance of the system to ensure that it is safe to use with patients. Mechanical parts, for example, worn worm drives, collimator attachment devices, patient couch and counterweights, must be inspected regularly. Where special safety devices exist to protect the patient, these should be tested on a regular basis to ensure that they actually function as specified.

As for any complex system, documentation is essential and the guidelines previously suggested should be followed. Preventive mechanical, computer hardware and software maintenance are all-important and the manufacturer's guidelines should be observed.

Error logging is at least as important as for planar scintillation camera–computer systems and all unexpected events should be recorded in the corresponding logbook.

4.1.4.1. Test conditions

The operational conditions to be used in the subsequent tests are based on the following recommendations on how the system is to be used.

(1) The recommendations pertaining to scintillation cameras (Section 2.4) must be followed.
(2) The recommendations pertaining to computer systems (Section 6) must be followed.
(3) The detector plane must be parallel with the axis of rotation, i.e. the tilt angle must be zero, or as close to zero as possible.
(4) The radius of rotation (for a circular orbit) must be known with reasonable accuracy for each tomographic test described here.
(5) The set-up of the energy window must be performed accurately and standard conditions employed for each test as recommended by the manufacturer.
(6) The set-up of the system with respect to uniformity is especially critical. Thus, these tests are to be performed with a uniformity correction applied as recommended by the system manufacturer, as specified for the use of the system in tomography. This normally means that a hardware and (where recommended) software uniformity correction must be used. In some cases, a specific uniformity correction for each radionuclide used in clinical imaging must be applied. In other cases, the use of one uniformity

correction is acceptable for all radionuclides. It must be ensured that in the collection of the uniformity correction data, the radiation flux across the entire detector surface is uniform for both intrinsic and extrinsic correction data. Usually, it is recommended to collect many more counts than normal for conventional imaging, typically 30 million to 100 million counts, using the same collimator (if extrinsic) with the same radionuclide and energy window as are to be used for the tomographic study.

(7) The corrections for the centre of rotation and alignments of the detector head with respect to the patient pallet must be applied. These corrections must be measured according to the manufacturer's recommendations and be determined on a frequent basis.

Some special requirements exist for tomographic acquisitions. Where it is specified in these tests that the finest matrix size available should be used for a tomographic acquisition, the largest matrix size available on the system should be employed, providing that the data acquired can then be reconstructed.

Before the tomographic tests are performed, the system must be set up in the manner proposed by the manufacturer. This is of particular importance with respect to the adjustment of the ADCs, for example, the control of gain and centring. This is especially important for multiple head SPECT systems (see Section 5).

4.1.4.2. Tests to be performed

On installation of the system, great care should be taken to ensure that the system is installed in a proper and stable environment and is functioning optimally. The following tests are the minimum set of tests that should be performed for SPECT imaging, in addition to those specified in Section 2, which tests basic planar performance:

(1) Physical and mechanical inspection of the SPECT system (see Section 4.3.1);
(2) Test to determine the absolute size of a pixel (see Section 4.3.2);
(3) Test of tomographic uniformity (see Section 4.3.3);
(4) Test of tomographic resolution in air (see Section 4.3.4);
(5) Test of tomographic resolution with scatter (see Section 4.3.5);
(6) Test of centre of rotation offset and alignment of axes (see Section 4.3.6);
(7) Test of slice thickness at centre of slice (see Section 4.3.7);
(8) Test of variations of uniformity and sensitivity with rotation of the system (see Section 4.3.8);

(9) Total performance test (see Section 4.3.9);
(10) Operational check of system function and centre of rotation offset (see Section 4.4.1).

Tests for absolute tomographic sensitivity and for the linearity of measurement of activity of the system have not been included in this publication.

4.1.4.3. Radiation sources and materials required

In addition to the radiation sources and material described in Section 2.1.7.8, some additional phantoms are required. These are:

(1) *A tomographic uniformity phantom*. This is a cylinder that may be filled with a uniform concentration of well-mixed activity. A typical size for such a phantom would be for the cross-section to be a circle of radius 10 cm and for the length to be preferably at least 10 cm. (An alternative is to use the appropriate section of a commercially available tomographic phantom such as the Data Spectrum ECT phantom (usually referred to as the Jaszczak phantom) (Data Spectrum Corporation, USA). All such phantoms normally include a section that can be used for this purpose.) A Data Spectrum ECT phantom (Jaszczak phantom) with inserts is illustrated in Fig. 45.
(2) *A tomographic point source*. This is a conventional point source, preferably small enough so that it can be contained within a 2 mm sphere, which can be placed at various points within the field of view of the system. The simplest way to obtain a suitable point source is to use a fine bore 1 mL syringe where the volume of activity is contained in less than 0.5 mL. It is important that the 'length' of liquid within the syringe (and not the needle) be as small as possible. The total activity of the source is not critical, but should be of the order of 40 MBq (1 mCi). This requires that the ^{99m}Tc be of high specific activity to achieve such a concentration. Alternatively, a small drop contained in the tip of a capillary tube can be used.
(3) *A SPECT resolution phantom*. This is a phantom with uniform attenuation, for example, the uniformity phantom, or alternatively a disk of Lucite, within which the tomographic point source described above can be placed. For example, a phantom may be used that comprises a disk of scattering material of 10 cm radius and about 5 cm thick with a series of holes of about 1 mm diameter bored at various radial distances from the centre. In particular, there should be one hole in the centre of the phantom. It should also be possible to support the phantom in air, at the centre of rotation.

(4) *A total performance phantom*. This is a phantom containing patterns of known form and spheres of known size, which may either be filled with activity of known concentration or 'cold' water (i.e. water with no radioactivity added). Several commercial phantoms exist. Examples are the Data Spectrum ECT phantom (Jaszczak phantom) (see Fig. 45) and the Carlson phantom (see Fig. 46). They contain inserts with different test objects which are to be detected or resolved and could be manufactured locally.

(5) *A liquid filled (conventional) flood source*. This is described in Section 2.1.7.8 and is used for testing planar uniformity and for setting up the uniformity correction matrices.

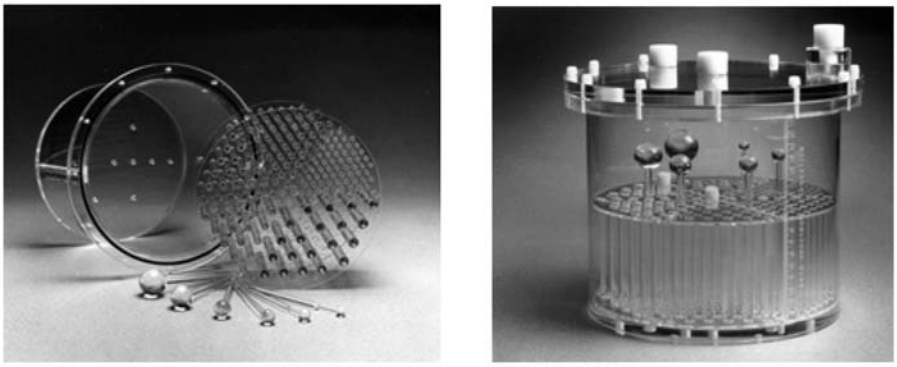

FIG. 45. Data Spectrum ECT phantom (Jaszczak phantom), showing the individual inserts for resolution and contrast assessments (left) and the assembled phantom (right).

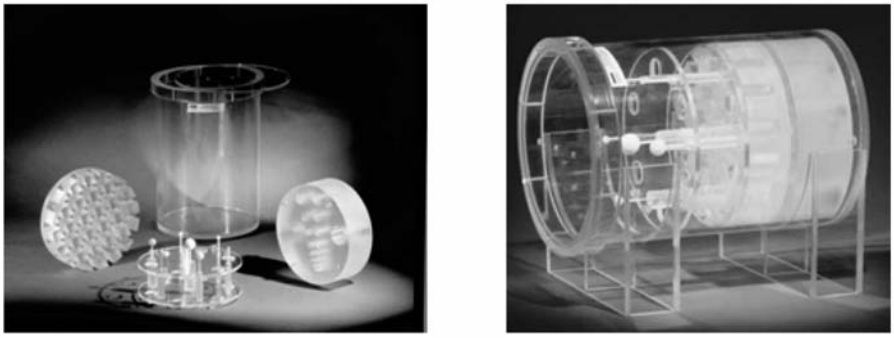

FIG. 46. Carlson phantom showing the individual inserts for resolution and contrast assessments (left) and the assembled phantom (right).

In general, phantoms should be filled such that their contents will mix. This is particularly important for the phantoms containing uniform sections. This may be performed by leaving space in the phantom after filling it with water such that, when the activity is added, the phantom can be mechanically agitated to ensure good mixing before the final portion of water is added. Alternatively, the water and activity can be mixed prior to filling the phantom. It is sometimes helpful to dilute the concentrated activity with a small amount of ink before mixing in order to provide a visible indication of the distribution of the activity. In addition, care must be taken when filling the phantom so as not to spill the activity. In particular, the phantoms themselves should be watertight. Care should be taken to ensure that this is the case before placing the phantom in the tomographic system where the detector could become contaminated or damaged. Some commercial systems exist which facilitate filling and mixing. Such phantoms should be shielded whenever appropriate to reduce radiation dose to staff, particularly when not in use.

4.2. TEST SCHEDULE

The frequency with which the specified tests should be performed will depend critically on how rapidly the system changes (Table 5). Many systems are quite stable and these tests need only be performed infrequently. Some systems do not perform in this way and considerable variations are observed from day to day, or even (for very unstable systems) hourly. Such systems are unlikely to be usable for tomography. The stability of the system must be determined in order to establish the frequency with which tests must be performed. Constant environmental conditions are essential.

4.3. ACCEPTANCE AND REFERENCE TESTS

4.3.1. Physical and mechanical inspection of the SPECT system

Purpose of test

To check the mechanical performance of the system and its capability to rotate the scintillation camera.

TABLE 5. LIST OF RECOMMENDED QUALITY CONTROL TESTS FOR SPECT

		Frequency of routine testing				
Section (test) no.	Test	Acceptance and reference	Daily	Weekly	Monthly	Half-yearly
	Acceptance and reference tests					
4.3.1	Physical and mechanical inspection of the SPECT system	X (reference)	X			
4.3.2	To determine the absolute size of a pixel	X				X
4.3.3	Tomographic uniformity of the camera	X (reference)				X
4.3.4	Tomographic resolution in air	X				X
4.3.5	Tomographic resolution with scatter	X				X
4.3.6	Centre of rotation offset and alignment	X		X[a]		
4.3.7	Slice thickness at centre of slice	X (reference)				X
4.3.8	Variations of sensitivity and uniformity with rotation of the system	X (reference)				
4.3.9	Total performance	X (reference)				X
	Operational checks					
4.4.1	Check of routine function and centre of rotation offset		X[a]	X[a]		

[a] In a well-functioning stable system, the check or test of centre of rotation offset should not be required at the frequency indicated and could be performed at the frequency indicated in the next column (i.e. weekly instead of daily and monthly instead of weekly). If the stability of a SPECT system is such that the centre of rotation offset changes daily, then this system should not be used for tomographic studies until the problem of instability has been resolved.

Materials

A spirit level, a set of accurate rulers of various lengths and a stopwatch. Some manufacturers now provide suitable test equipment, for example, a laser, by means of which the mechanical installation can be checked accurately, but sometimes available only on installation.

Procedure

(1) Check the system for damage.
(2) Rotate the scintillation camera. Check for constancy of speed, vibration, the presence of mechanical noises and whether it stops correctly at the end of rotation. Check both clockwise and anticlockwise rotation. In particular, use the stopwatch to check if rotational speed changes as a function of angular position. The motors may have to work much harder to lift the head, in contrast to when the head is descending. Ensure, where possible, that the head is adequately counterbalanced for the different collimators available.
(3) Where appropriate, check that the system returns to its 'home' position accurately and reliably.
(4) For a step and shoot system, check that the correct number of angles is used, for each possible angular increment that can be selected.
(5) Check that the head is mechanically centred with respect to the axis of rotation to within the manufacturer's specifications. Typically, this should be accurate to within 1–2 mm. If the errors are taken into account by the reconstruction software and if such errors are consistent, then errors up to 1 cm may be accepted if the test for tomographic resolution gives acceptable results (see Section 4.3.4). With errors larger than 1 cm, electronic correction of the centre of rotation offset becomes difficult.
(6) Check that the Y axis of the head is parallel to the axis of rotation. This may also be difficult to measure mechanically without special purpose equipment and the manufacturer must check this at the time of installation. The test for centring given in Section 4.3.6 can provide information about the alignment of the Y axis to the axis of rotation.
(7) Check that the long axis of the bed is reasonably parallel to the axis of rotation, where appropriate, and centred. This may be checked by measuring the distance from the bed to the camera with the head at 90° and then at 270°. These two distances should be within 1 cm of each other. This should also be checked for two extreme positions of the bed (where appropriate), which will confirm that the bed is reasonably parallel to the axis of rotation.

(8) Check that any readings of head position, for example, angle, or distance from the axis of rotation, are accurate by measuring the corresponding distances, angles, etc. The radius of rotation is most easily performed by taking a measurement from the head to some fixed point, for example, on the bed, then rotating the head by 180°, repeating the measurement and dividing the sum of the distances by two.
(9) Check that the gantry is vertical. This can be ensured by placing a spirit level on the collimator at 0° and at 180° along the Y axis. The reading should be the same in both cases. If not, the manufacturer should rectify the situation. This is important when adjusting the angle of tilt, if a spirit level is used. Errors in adjusting the verticality of the gantry will show up in the centring test (see Section 4.3.6).
(10) For a system using a spirit level on the back surface of the detector head, check if the collimator face is parallel to the back surface of the detector. Place a spirit level on the collimator with the detector head facing down and compare with the spirit level on the back surface of the detector. The two surfaces should be parallel.
(11) In many systems, as the radius of rotation is changed, the head tilt angle is adjusted to ensure that the head remains parallel to the axis of rotation. Check that the head remains level as the radius of rotation is increased.
(12) Visually check the collimators for damage.
(13) Check that the emergency stop button and any patient safety devices function.
(14) For non-circular orbits, check that the system performs the correct mechanical motions for several different selected orbits. Tests that are more detailed will normally be required and should be indicated in the manufacturer's documentation.
(15) Check that any cables do not become twisted or damaged when the system rotates. Check all such electrical connections, for example, if a slip ring is used, visually inspect it.

Data analysis

None.

Observations

This test is intended to be performed as a reference test and daily.

The capability of the system to perform the mechanical movements required without vibration is important. Many problems can be indicated by noises generated by the system, or from observing the time taken to rotate from position to position.

The radius of rotation as indicated on some readout display may indicate distance from the axis of rotation to the front of the collimator, or alternatively, to the front of the crystal. Note which of these is, in fact, the case.

Interpretation of results

If a SPECT system is not set up very carefully mechanically, it will not be capable of providing good SPECT images. The tolerances permitted are far less than for a conventional scintillation camera. Most of the errors resulting from such mechanical problems will become more obvious by the centring test (see Section 4.3.6).

Limits of acceptability

In order to perform good SPECT studies, the mechanical positioning of the head needs to be such that the centre of rotation offset is less than 1 mm. However, when adequate software or electronic centring methods exist, the centre of rotation offset needs to be less than 1 cm and reproducible to within 1 mm.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken. If the mechanical accuracy is not within tolerance and cannot be corrected by software, then the system should not be used for SPECT.

4.3.2. Test to determine the absolute size of a pixel

Purpose of test

To determine the absolute pixel size in the matrix used for tomographic reconstruction.

Materials

One or two point sources and an accurate ruler. This test should be performed for all the matrix sizes and tomographic zoom conditions used in clinical practice.

4.3.2.1. Method 1: Using one point source

Procedure

(1) Place the point source on the camera face along the X axis, about 5 cm from the edge of the field of view.
(2) Set up the system to perform a conventional static acquisition of about 50 000 counts using the finest possible matrix size, for example, 256 × 256 or 512 × 512. Ensure that no zoom is used.
(3) Acquire one planar image.
(4) Now move the point source horizontally to a position about 5 cm away from the other edge of the field of view, by a distance known to within 1 mm.
(5) Repeat the acquisition.

4.3.2.2. Method 2: Using two point sources

(1) Place two point sources as indicated in steps (2) and (4) above.
(2) Acquire one planar image using the finest matrix size available.

For both methods

(1) Repeat the whole procedure by placing the point sources along the Y axis.
(2) Repeat for all tomographic zoom conditions used in clinical practice.

Data analysis

The analysis requires the calculation of the centre of gravity of a point source along either the X or Y axis directions for each raw projection image. The centre of gravity in X (COGX) is estimated from:

$$\mathrm{COGX} = \frac{\sum_{i=i_1}^{i_2}\sum_{j=j_1}^{j_2} i \times \mathrm{MATRIX}(i,j)}{\sum_{i=i_1}^{i_2}\sum_{j=j_1}^{j_2} \mathrm{MATRIX}(i,j)} \tag{16}$$

along a profile of thickness j_1 to j_2 and width i_1 to i_2 bounding the point source, where i is the index of the matrix along the X axis and j corresponds to Y.

The centre of gravity along the Y axis (COGY) is similarly obtained from:

$$\mathrm{COGY} = \frac{\sum_{j=j_1}^{j_2}\sum_{i=i_1}^{i_2} j \times \mathrm{MATRIX}(i,j)}{\sum_{j=j_1}^{j_2}\sum_{i=i_1}^{i_2} \mathrm{MATRIX}(i,j)} \tag{17}$$

where j is the index in the Y direction.

The values COGX and COGY should be estimated to a fraction of a pixel.

(1) Calculate the centre of gravity of the point source(s) for each image. Four centre of gravity values must be obtained. These are:
(X1, Y1) and (X2, Y2) for the two positions along the X axis; and
(X3, Y3) and (X4, Y4) for the two positions along the Y axis.

(2) From each pair of observations, calculate the distances between the position of each of the pairs of point sources using:

$$\mathrm{DISTANCE\ X} = \sqrt{(\mathrm{X1}-\mathrm{X2})^2 + (\mathrm{Y1}-\mathrm{Y2})^2} \tag{18}$$

$$\mathrm{DISTANCE\ Y} = \sqrt{(\mathrm{X3}-\mathrm{X4})^2 + (\mathrm{Y3}-\mathrm{Y4})^2} \tag{19}$$

For the horizontal displacement, Y1–Y2 should be small, as should X3–X4 for the vertical displacement.

(3) Calculate the pixel size by dividing the distance between the point sources in millimetres by the corresponding distance in pixels. This gives the size in millimetres of the pixel used in order to perform this measurement.

(4) Multiply the value found in step (3) by an appropriate factor so that it corresponds to the matrix size and the pixel size as used in tomography, for example, by a factor of four if the data were collected in a 256 × 256 matrix (a matrix of 64 × 64 is used in tomography).

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals.

The pixel size is that required for tomographic reconstruction, for example, for the attenuation correction algorithm. The positioning of the point sources must be performed carefully and the distance between the first and second positions of the point source (or the positions of the two point sources) must be accurate to within 1 mm. The pixel size in the X direction and the Y direction may not be the same.

Interpretation of results

In order to measure resolution, in order to apply attenuation correction and in order to perform quantitative estimates of the size of organs, it is necessary to know the absolute size of a pixel in millimetres. Compare the pixel size in X and Y.

Limits of acceptability

The difference between the values in X and Y should be less than 5%.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.3. Test of tomographic uniformity of the system

Purpose of test

To test the tomographic uniformity of a rotating scintillation camera SPECT system.

Materials

The tomographic uniformity phantom should be filled with about 200–400 MBq (5–10 mCi) of ^{99m}Tc, making sure that the activity is well mixed.

Procedure

This test should be performed after the test for planar uniformity has been performed.

(1) Ensure that all the camera uniformity correction calibration procedures have been correctly performed.
(2) Place the phantom with its centre at least within 2 cm of the axis of rotation, as close as possible to the centre of rotation.
(3) Ensure that the central axis of the phantom is parallel to the axis of rotation.
(4) Set up a tomographic acquisition using a normal matrix size (e.g. 64 × 64 or 128 × 128) and the number of angles used clinically, using a circular orbit.
(5) Perform a standard tomographic acquisition, collecting a total of about one million counts per slice. This typically corresponds to 15 million total counts for a phantom 10 cm in length, or about 240 000 counts per angular position for a 64 angle acquisition.
(6) Perform uniformity correction as recommended by the manufacturer.
(7) Reconstruct the data with a ramp (or sharp) filter.
(8) Where possible, perform attenuation and scatter correction using the method prescribed by the manufacturer. The attenuation correction is essential unless special purpose software is used.

Data analysis

(1) Inspect images of the phantom at various transaxial positions.
(2) Place a profile about 5 pixels thick through the centre of the image (normally through the point corresponding to the centre of rotation). Estimate the depth or height of any artefacts corresponding to circular (ring) artefacts by measuring their contrast with respect to the surrounding activity, as defined in Section 4.3.3.
(3) Identify the minimum or maximum value corresponding to the location of a ring artefact as seen in the reconstructed image. Record this value, terming it $C_{\text{min/max}}$.

(4) Record the two values along the profile of the uniform source just beyond the edges of the artefact identified in step (3), terming them $C1$ and $C2$.
(5) Calculate $C_{ave} = (C1 + C2)/2$.
(6) Estimate the contrast as $(C_{min/max} - C_{ave})/(C_{min/max} + C_{ave})$.
(7) Repeat for all the other transaxial sections within the phantom and determine the maximum absolute value of contrast.
(8) For a central slice, determine the central value by averaging over 5 pixels (about 3 cm for a 64 × 64 matrix) on the profile corresponding to the centre of the phantom, or use a 5 × 5 pixel region of interest to give this value.
(9) Determine the edge value by averaging over 3 cm on the profile centred 2 cm from the observed edge (50% value).
(10) Where possible, measure the size of the body contour in the horizontal and vertical directions in pixels and convert into distances in millimetres.

Observations

This test is intended to be performed as a reference test, at half-yearly intervals and whenever a uniformity problem is suspected.

The uniformity of a rotating scintillation camera SPECT system must be as good as possible, because any non-uniformity is amplified by the tomographic reconstruction process. The planar uniformity of a scintillation camera when used in SPECT should be better than 4%. This is very difficult to achieve. However, if the NEMA integral uniformity index is worse than 6% after uniformity correction, it is clear that the camera needs attention and should be tuned. Compare the measured values with those obtained for conventional planar uniformity at the time of acceptance. In particular, those variations in planar uniformity lying along or close to the vertical (Y) axis are very important and the limits of acceptability should be much stricter.

All the reconstructed transaxial slices passing through the phantom should be inspected for circular (ring) artefacts, except those within 2 pixels of the edge of the phantom. It is helpful to mark the central point of the image, for example, by marking a horizontal and vertical profile through the centre of the tomographic slice. Artefacts are always circles centred about this point (for a circular orbit). All visible artefacts are significant. The measured values of contrast as calculated above are rather variable and it is advised that more than one estimate of the amplitude of a ring artefact be made. Do not sum together a number of transaxial slices or smooth the data since this may cause ring artefacts to disappear.

If the system has not been accurately centred (see Section 4.3.6), circular artefacts may not be visible because their effects have been 'smeared out'. When performing the attenuation correction, a typical value for the attenuation correction factor for ^{99m}Tc is 0.12 cm^{-1} when no scatter correction has been performed.

Interpretation of results

Thick rings, widely spaced, are usually indicative of variations in the uniformity of the camera itself. Narrow (thin) rings, often close together, are usually an indication of errors in the camera–computer interface, or of the digital part of any uniformity correction hardware. A sharp cold or hot spot may be seen exactly at the point corresponding to the centre of rotation and represents a problem in uniformity along the projection of the axis of rotation. A 1% non-uniformity at the centre of rotation will be amplified into about 20% non-uniformity in the reconstruction.

Limits of acceptability

The contrast measured between any ring artefacts and the uniform background, as measured using a profile, should not exceed 10%. The difference between the central value and the edge value should not exceed 10%. If this is not the case, the body contour should be checked first, followed by the attenuation correction coefficient (see Fig. 47) and pixel size used (see Section 4.3.2).

The horizontal and vertical directions measured on the body contour should be within 1 cm of the corresponding real dimensions of the phantom, if attenuation correction using this body contour is to be used.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

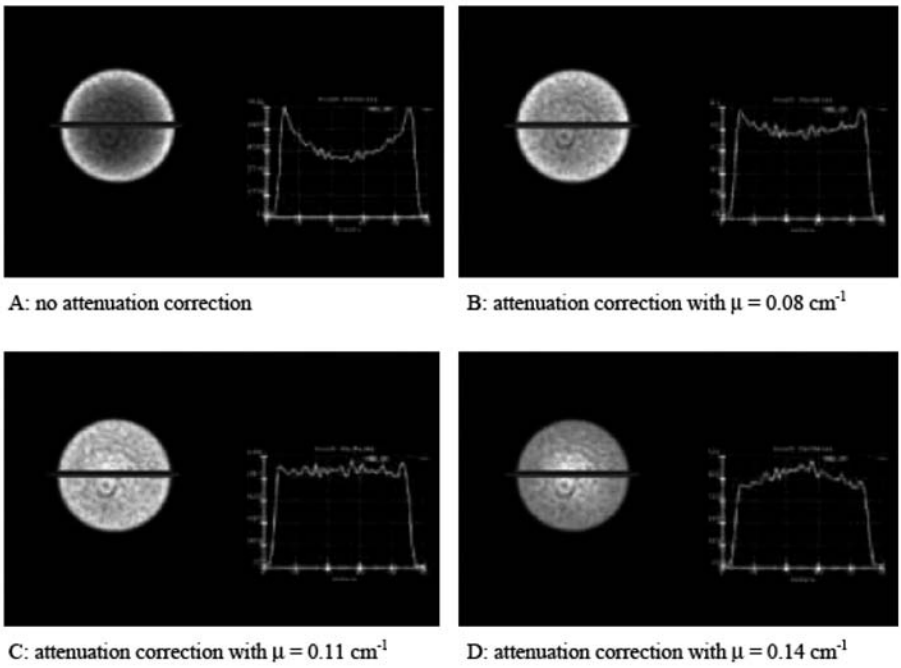

FIG. 47. Profiles to check the accuracy of attenuation correction. A cylindrical phantom, 20 cm in diameter, containing a homogeneous solution of ^{99m}Tc was used to acquire a high count data set. Acquisition: 128 × 128 matrix, 360° total angle of rotation, 128 projections, 800 000 counts in the first projection, radius of rotation 19 cm, circular orbit, pixel size 3.2 mm. Reconstruction: Filtered backprojection with a Butterworth filter, transverse slices. Attenuation correction was applied using different linear attenuation coefficients (Chang method). A profile was drawn across one of the transverse slices. A: No attenuation correction. B: Attenuation correction with μ = 0.08 cm^{-1}. C: Attenuation correction with μ = 0.11 cm^{-1}. D: Attenuation correction with μ = 0.14 cm^{-1}. Only for the attenuation correction using μ = 0.11 cm^{-1} is the profile through the slice essentially flat, apart from statistical fluctuations in the profile, indicating that the attenuation correction software is correct. For the image with a profile that is lower in the centre (B), the attenuation coefficient was too small. For the image with a profile that is higher in the centre (D), the attenuation coefficient was too large (see Ref. [3]).

4.3.4. Test of tomographic resolution in air

Purpose of test

To measure the tomographic resolution of the system in air and to ensure that the reconstruction process is not degraded by either the tomographic

acquisition or the reconstruction. Note that this is now considered to be the best test of centre of rotation accuracy.

Materials

A small point source of ^{99m}Tc, as used in the test for centre of rotation and alignment (see Section 4.3.6).

Procedure

(1) Place the point source in air within 1 cm of the centre of rotation, near the centre of the field of view.
(2) Set the radius of rotation to be approximately 15 cm, or if this cannot be achieved, to be as small as possible. Use a circular orbit of rotation.
(3) Perform a tomographic acquisition using the matrix size and number of angles used clinically, collecting about 10 000 counts per view.
(4) Reconstruct the data with filtered backprojection, using either a ramp filter or the sharpest filter that the system will permit.
(5) Perform a normal planar (static) acquisition at the home position, using the same acquisition matrix size, etc., as for the tomographic acquisition.
(6) Repeat steps (1)–(5) with the point source placed about 8 cm off axis.
(7) Repeat steps (1)–(5) with the point source placed on the axis of rotation, but close to the edge of the field of view (close to $+Y_{MAX}$ and $-Y_{MAX}$), as indicated for the centre of rotation test (see Section 4.3.6, step (5)).

Data analysis

(1) Draw a profile through the image of the point source in the reconstructed image and calculate the FWHM in both the horizontal and the vertical directions, estimating the FWHM as described in Section 2.3.8.
(2) Measure the FWHM in the horizontal direction on the planar image acquired at the home position.

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals.

There should be no significant difference between the FWHM calculated from the horizontal and vertical profiles in the reconstruction, when a circular orbit is used, in air. There may be differences for non-circular orbits, in scatter, and when the source is offset from the axis of rotation.

The interpretation of the results does not change when a larger radius of rotation than recommended has been used, but the absolute values of the FWHM will increase for both tomographic and planar images.

Interpretation of results

This is a useful system test to ensure that the centre of rotation of the system has been accurately calibrated, that the acquisition and reconstruction software is functioning correctly and that adequate performance can be obtained. Any error in the centre of rotation, any errors due to vibration, etc., will result in a loss of tomographic resolution with respect to planar resolution. If a filter other than a ramp filter has been used in reconstruction, there will be some degradation of tomographic resolution by comparison to planar resolution.

Set the display so that there is a very low maximum cut-off level, for example, by observing the background. The reconstructed image of the point source should appear to be round and, in particular, not distorted into shapes such as a comma, ellipse or ring. There are likely to be streaks radiating symmetrically from the point source but there should not be so-called 'preferred directions' where the streaks appear to be considerably more significant than others. There may be an 'edge' surrounding the image but this should not be much more significant than the amplitude of the streaks.

There should be no difference (except for the field of view) between the results obtained, as described above, with the point source in the central slice and the results obtained with the point source close to the edge of the field of view.

Limits of acceptability

Depending on the collimator, reconstruction filter and the radius of rotation, various values for the FWHM may be obtained. If the FWHM measured on the planar view is 12 mm (a typical value for a radius of rotation of 15 cm and a high resolution parallel hole collimator), then the FWHM on the tomogram should not be worse than 13.2 mm (i.e. 12 mm +10%) when a ramp filter has been used. The difference between planar and tomographic resolution should be no more than 2 mm or 10% (of the planar resolution), whichever is less. If it is worse, then the system has probably not been well centred. If this is the case, a new centre of rotation calibration should be performed. If, after recalibration, the FWHM is still poor, then this may be the result of vibration, or other such error, and the system needs adjusting. If a

filter other than a sharp (ramp-like) filter has been used, the tomographic resolution will be worse than predicted here.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.5. Test of tomographic resolution with scatter

Purpose of test

To check the tomographic resolution of the system in clinical conditions, that is, with a radius of rotation that is realistic and with scatter present. To give an indication of the resolution which is likely to be achieved clinically.

Materials

Use is made of the resolution (or an equivalent) phantom as described in Section 4.3.4. The point source is placed at the centre of this phantom. For example, with the special phantom described, the central hole is filled with high specific activity ^{99m}Tc such that the hole is completely full and contains about 20 MBq (0.5 mCi) of activity.

Procedure

(1) Place the centre of the phantom within 2 cm of the centre of rotation and close to the centre of the field of view.
(2) Adjust the radius of rotation so that it is about 15 cm, if possible, as in the test in Section 4.3.4: Test of tomographic resolution in air.
(3) Collect a tomogram under normal clinical conditions, using the usual matrix size and number of angles, collecting about 10 000 counts per angle.
(4) Reconstruct the data with filtered backprojection, using either a ramp filter or the sharpest filter that the system will permit.
(5) Where possible, perform an attenuation correction and any other appropriate corrections such as for scatter.
(6) Move the phantom so that its centre is about 5 cm away from the axis of rotation and repeat steps (3)–(5) using a larger radius of rotation.

Data analysis

(1) Draw profiles through the reconstructed image of the point source, horizontally and vertically.
(2) Measure the FWHM, on the horizontal and vertical profiles.
(3) Measure the maximum height of the background at the edge of the picture (within the area reconstructed) as a percentage of the central value.

Observations

This test is intended to be performed as an acceptance and reference test and at half-yearly intervals.

The resolution obtained is highly dependent on the radius of rotation used. There can be a considerable difference between the resolution measured in air and in scattering conditions. This can sometimes result from problems associated with the energy window.

Interpretation of results

Inspect the image with a low maximum cut-off value, as described in Section 4.3.4. There should be no difference between the horizontal and the vertical FWHM, for the central point source, and there should be little difference for point sources at the centre of rotation and when offset. With a low cut-off, the point source should still appear reasonably round and not elliptical.

Limits of acceptability

The resolution with scatter will be worse than in air. However, it should not change with time under the same conditions of measurement.

If the test of resolution in air gives good values while this test gives poor results, then the energy window should be checked as this could be the cause of such degradation. The background at the edge of the reconstructed area should be less than 5% of the peak value corresponding to the reconstructed point source.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.6. Test of the centre of rotation offset and alignment of axes

Purpose of test

To test the centre of rotation offset, alignment of the camera Y axis and head tilt with respect to the axis of rotation. This is considered a test to be performed if an error is observed with the test for resolution in air (see Section 4.3.4). This is an extended version of a test that should be described in the manufacturer's SPECT system manual.

Materials

A small ^{99m}Tc point source is used, together with some method of suspending it in air within the field of view, for example, by attaching the source to a long ruler, or a purpose-made supporting device. The computer system should be set up as specified by the supplier as indicated in step (3) below, noting the comments given previously in Section 4.1.3.

Procedure

(1) Using a spirit level, ensure that the camera is accurately aligned so that the head is parallel with the axis of rotation, i.e. that the head is not tilted (but see observations below).
(2) Suspend the point source in air within about 2 cm of the axis of rotation and within about 2 cm of the centre of the field of view.
(3) Perform a normal tomographic acquisition using the finest digital matrix size available, collecting about 10 000 counts at every angular position. An acquisition consisting of 32 angles over 360° is adequate for this test.
(4) Repeat steps (2) and (3) with the point source placed about 10 cm radial distance away from the centre of rotation.
(5) Repeat steps (1)–(4), placing the point source along the axis of rotation, but as far as possible away from the central slice, for example, within 5 cm of the edge of the field of view in the positive Y direction. It is important to ensure that the point source is always within the field of view of the camera throughout the tomographic acquisition.
(6) Repeat step (5) with the point source close to the edge of the field of view in the opposite direction. Note that, alternatively, if suitable software is available, three point sources may be used and a single set of measurements performed for the central point (steps (2), (5) and (6)).
(7) Perform steps (1)–(6) for rotation in the opposite direction (if the system can acquire data in both the clockwise and anticlockwise directions).

Data analysis

Most manufacturers' systems provide software to calculate and incorporate the correction required into the normal tomographic acquisition and reconstruction process. The test and the methods used vary considerably from system to system. The centre of rotation correction accuracy can be checked by using the resolution test in air (Section 4.3.4), or the raw values checked using the method given below. The method suggested does not depend on the special software provided by the manufacturer for this purpose.

The aim of this test is to estimate the centre of gravity of the image of the point source, angle by angle, and hence to estimate the position of the centre of rotation. Most software packages treat the centre of rotation, as used in the reconstruction, as being at $N/2 + 0.5$ (where the pixel on one edge is designated 1 and the pixel on the other edge, N). If this is not true, the calculations given below should be converted into the frame of reference used by the software provided. (For example, an alternative convention is to refer to the pixel on one edge as 0 and on the other edge as $N - 1$. In this case, the centre of rotation would simply be $(N - 1)/2 + 0.5$.)

Calculate the centre of gravity of the point source for each image, using the method given in the data analysis part of Section 4.3.2. The values COGX and COGY should be estimated to a fraction of a pixel. Two methods of analysing the results exist; both methods may require special purpose software to implement. Most tomographic systems provide a program to perform one or the other of the calculations given below.

Method A

(1) The offset from the centre of rotation should be calculated as follows. If X_0 is the value of COGX at 0° and X_{180} is the value at 180° degrees, then, if N is the number of pixels across the image (e.g. 256 if the data were collected in 256 × 256), the offset from the centre of rotation, R, is given by:

$$R_0 = (N + 1 - X_0 - X_{180})/2 \qquad (20)$$

(2) This value should be calculated for each pair of angles, θ, separated by 180° to generate a set of values $R(\theta)$.

Method B

(1) The COGX may be plotted as a function of angle over the total angle of rotation, here 360°.
(2) A sine function A + B sin($\theta + \phi$) can be fitted to this curve, where θ is the angle of rotation and A, B and ϕ are fitting constants.
(3) The value of A should be compared with the expected centre of the matrix (normally $(N + 1)/2$, as stated above). The difference between the constant A and the centre of rotation is the mean offset. Mathematically, this value should be identical to that calculated by method A.
(g) The fitted sine function should be subtracted from the observed curve to show the residuals. This indicates the variation in the centre of rotation as a function of angle of rotation; $R(\theta)$ is given by these residuals plotted against angle θ.

For both methods

(1) The value of $R(\theta)$ should be plotted as a function of angle.
(2) The mean value of $R(\theta)$, its standard deviation and maximum deviation from the mean value should be calculated.
(3) The centre of gravity of the point source along the Y axis should be calculated using the same method (for each detector head) and should be recorded for each angular position. A manipulation which may be used with some software to obtain the plot of the variations in the Y axis is to rotate the raw data by 90° and use the same software as that used for estimating the X axis variation.
(4) Convert the values thus determined into millimetres by using the known pixel size for the camera, as determined in Section 4.3.2, for the corresponding matrix size.

In particular, step (3) must be performed for each head separately for multiple head systems, since it is very important to check that the Y axis gains and offsets of each of the heads match. All of these calculations are identical for both the normal acquisition performed with the well centred sources, for the clockwise and anticlockwise acquisition, and for the various other positions of the point source away from the central slice.

Observations

This test is intended to be performed as a reference test and at weekly or quarterly intervals, depending on the stability of the system.

A SPECT system must be accurately centred if resolution is not to be degraded and this test is designed to ensure that the reconstructed image does not suffer from degradation resulting from this cause. Every millimetre loss of accuracy in centring, whether mechanical, electronic, within the camera head or in the interface, will degrade the resolution by a greater amount in the reconstructed image. While the two methods of analysis give the same value for the mean offset averaged over opposite views, the second method gives a better indication of variations in offset as a function of angle of rotation.

When problems are observed with multiple head systems, a good tactic to use in order to identify the source of such problems is to treat each head separately as a single headed system. For example, with a dual headed system, acquire data for each head over 360° and apply the data analysis separately for each head.

As previously stated (see Section 4.1.1), the centre of rotation offset may not be constant with respect to angle. These second order effects may be observed on the plot of variations with angle, which should be small. They may, however, be ignored if they are taken into account by the reconstruction software, provided that they are reproducible. In this case, the centring test should be repeated in order to confirm reproducibility.

Interpretation of results

The interpretation of the results will depend on the extent to which the hardware and software of the tomographic system correct for errors in centring and therefore on the results of Section 4.3.4: Testing tomographic resolution in air. If the results of this test are unsatisfactory, the most likely explanation is the inaccuracy of the centring of the system. The interpretation of these two tests should be considered as a pair.

If the centre of rotation offset in X used by the system is accessible, this should be compared with that calculated using this test. For those systems that do not perform a centre of rotation offset correction, the value estimated by this test should tend to zero.

The curve of the offsets should be reasonably smooth and flat. The offset at the start and end angles should be very close. If considerable fluctuations exist, as measured by the standard deviation of the offset, in particular, if they are not reproducible, the system is likely to give poor clinical performance.

The plot of the centre of gravity along the Y axis for a source radially distant from the central axis gives a good indication of the tilt of the head or possible collimator hole angulation. If the head is not tilted, this Y axis offset should be independent of angle (the plot should be flat). It is important to realize that if the gantry supporting the head(s) is not accurately vertical, then

the axis of rotation will not be exactly horizontal. Thus, aligning the head(s) with a spirit level will not ensure that the detector surface is parallel with the axis of rotation (see Section 4.3.1, procedure steps (6) and (9)).

With multiple heads, it is important that the values of the X axis offsets measured for each head are the same, unless the reconstruction software specifically takes this into account. In addition, the values for the COGY should be the same for each head at each angular position. Thus, the plots for each head should be compared, or, if the results for one head are plotted as a continuation of those for another head, the data should be continuous. For example, the plot of the Y centre of rotation should be flat over the range of all head angles considered.

The centre of rotation offset should be independent of the position of the point source within the field of view. If this is not the case it may be an indication that the Y axis is not aligned with the axis of rotation.

Caution must be taken if the method for correction of circular rotation error is included in the reconstruction and is inaccessible to the user. In such a case, the raw acquired projection images may indicate a centring error that is, in fact, corrected for in the ensuing reconstruction.

Limits of acceptability

The mean value of the centre of rotation offset should be less than 2 mm, or it must otherwise be corrected. The centre of rotation offset estimated at the centre and for the edges of the field of view should all be within 2 mm of each other. For multiple head systems, the position of the Y = 0 axis, as well as the Y gain, should be the same for both heads.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.7. Test of slice thickness at the centre of the field of view

Purpose of test

To test the thickness of a tomographic slice at the centre of the field of view. To ensure that the spatial resolution along the tomographic Z axis is within acceptable limits.

Materials

A small point source of ^{99m}Tc, as used for the test of tomographic resolution in air (Section 4.3.4) and the test of centre of rotation offset (Section 4.3.6).

Procedure

The data collected for this test are the same as those acquired for the test of tomographic resolution in air (Section 4.3.4). Only the analysis of those data differs here and the same raw data may be used for this test.

(1) Place the point source in air within 1 cm of the centre of rotation, near the centre of the field of view.
(2) Set the radius of rotation to be approximately 15 cm, or, if this cannot be achieved, to be as small as possible. Use a circular orbit of rotation.
(3) Perform a tomographic acquisition using the normal matrix size and number of angles used clinically, collecting about 10 000 counts per angle.
(4) Reconstruct the data with filtered backprojection, using either a ramp filter or the sharpest filter that the system will permit. Do not sum transaxial slices together; a slice thickness of a single pixel must be used.

Data analysis

(1) Locate the slice in which the point source is most clearly seen (and is a maximum amplitude) and locate the pixel in which the maximum number of counts is observed. Note the (X, Y) coordinates of this pixel and note this maximum value.
(2) Record the number of counts at this same (X, Y) pixel position for all slices adjacent to and including the slice in which the maximum was found, such that all slices containing counts of more than 5% of the maximum are included. Generate a profile of the point source along the Z axis using these values.
(3) Calculate the FWHM of this profile (Section 2.3.8) and convert into millimetres using the pixel size as determined in Section 4.3.2.

Observations

This test is intended to be performed as a reference test and at half-yearly intervals.

It is particularly important that the point source be small and that only a single point source be employed. It is common when filling a syringe that activity remains in the needle used to fill the syringe, which will result in the presence of a second (albeit small) point source. While this does not affect the results obtained for the resolution test given in Section 4.3.4, it is likely to affect the results given here. The presence of such a secondary point source may be observed by inspection of any of the original (raw) projection images. If this is observed, change the needle for a clean one, or a blind cap. Reconstructions close to, but not including, the point source may contain significant streak artefacts, which should be ignored.

Interpretation of results

The FWHM for the slice thickness (here, as measured in air) should be the same as the normal transverse tomographic resolution in air for the distance corresponding to the radius of rotation.

Limits of acceptability

The slice thickness at the centre should be within 10% of the tomographic resolution as determined in Section 4.3.4.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.8. Test of variations of uniformity and sensitivity with angle

Purpose of test

To determine the variations in system sensitivity as a function of angular position of the detector.

Materials

A flood source of about 200 MBq (5 mCi) of ^{99m}Tc, which can be safely attached to the collimator face such that the system can be rotated. A ^{57}Co flood source could be used if the field uniformity is not computed for each angle.

Procedure

(1) Attach the flood source firmly to the camera so that it cannot shift when the system rotates.
(2) Perform a tomographic acquisition with at least about 10^6 counts per angle, for the normal matrix size used.
(3) For a system collecting data by continuous rotation, record the total rotation time. Repeat the test for both fast and slow rotation, for example, total rotation times of 4 min and 30 min.
(4) Repeat for any other heads.

Data analysis

(1) Find the total number of counts collected at each angle. Correct for decay of the radionuclide used if total acquisition time is significant, e.g. 60 min for ^{99m}Tc.
(2) Calculate the mean, standard deviation and maximum deviation from the mean.
(3) Perform the same calculations for a central region of interest.
(4) If suitable software exists, calculate the NEMA integral uniformity for each angular position. If such software is available, it might also be useful to compare each view on a pixel by pixel basis.
(5) For a continuous rotation system, perform the analysis for both a slow and a rapid rotation speed.
(6) If the speed of rotation (or time per acquisition) can be measured independently, this should be recorded.

Observations

This test is intended to be performed as a reference test.

Variations in uniformity and sensitivity as a function of angle can be caused by lack of magnetic shielding, changes in temperature and (for continuous rotation) mechanical drive problems.

Interpretation of results

Some systems show considerable variation in sensitivity and uniformity as a function of angle, probably because of the influence of the earth's magnetic field. These variations can cause considerable differences in uniformity, etc., as a function of angle, but are eliminated primarily by either good magnetic shielding of the photomultipliers, an appropriate energy correction, or both.

An alternative source of variation in sensitivity and uniformity with angle results from changes in temperature within the head as a function of angle. Check the deviations from the mean for the total number of counts and the central region of interest.

Limits of acceptability

The variation in sensitivity should be less than ±1% of the mean value. If it is greater than this, then the system is not performing satisfactorily. It may, however, be possible to perform an appropriate correction with software.

The results for continuous rotation should be similar for both rotation speeds. In particular, note if there is an apparent increase in counting time, corresponding to an increase in the number of collected counts for those angles when the camera head is rising, compared with the angles where the head is descending.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.3.9. Total performance test

Purpose of test

To verify that the system is performing adequately in a high count study. To estimate the contrast of objects of known size.

Materials

A total performance phantom, for example, the Data Spectrum ECT phantom (Jaszczak phantom) or the Carlson phantom or other such phantom as described in Section 4.1.4. It should have at least one region with uniform activity and one region with cold lesions to be detected. It is also desirable to have some estimate of resolution. The activity contained in the phantom should be about 400 MBq (10 mCi) of ^{99m}Tc.

Procedure

(1) Set up the total performance phantom, place it on the bed and carefully align it to be parallel to the axis of rotation. Alternatively, use a special holder to place the correctly aligned phantom along the centre of rotation.
(2) Acquire a tomographic study using the acquisition time in order to collect 800 000 counts for each projection, 120 projections, a matrix size of 128 × 128 and a 360° angle of rotation.
(3) Reconstruct the data with filtered backprojection, using either a ramp filter or the sharpest filter that the system will permit.

Data analysis

(1) Review each transaxial slice carefully, looking for ring artefacts, distorted cold spheres and rods.
(2) Add together 8 transaxial slices through the rod section, 5 slices through the uniformity section and 3 slices through the cold sphere section. Record the smallest rod section and the smallest sphere visible.
(3) Place a profile across the summed slices of the uniform region of the phantom. Measure the ratio of counts per pixel at the centre to counts per pixel at the edge. An estimate for the linear attenuation coefficient may be obtained from this ratio using:

$$\mu\ (\text{cm}^{-1}) = [\ln(\text{edge counts per pixel/centre counts per pixel})]/\text{radius} \quad (21)$$

The value for μ can then be used in the attenuation correction calculation. Also, measure the amplitude in per cent of any artefacts, as described in the test of tomographic uniformity (Section 4.3.3).
(4) By using appropriate profiles or regions of interest, measure the contrast of all visible spheres. Record the values at acceptance and compare values when the test is performed routinely.

Observations

This test is intended to be performed as a reference test, at half-yearly intervals and if a problem is suspected.

Different total performance phantoms have been designed for different purposes. In particular, some phantoms, such as the Jaszczak, have been designed to test the limits of performance of the system and must be used with

as many counts as possible, in circumstances that are non-clinical, as described in this test.

Other phantoms have been designed to mimic the clinical situation and are properly employed with much fewer counts and are not normally reconstructed with a ramp filter. The former type of phantom is much more difficult to image and is probably a better test of system performance, although it requires much more time to image. The latter type of phantom may not give a good indication of whether the system is performing optimally, but can give a good indication of how the system is performing clinically.

Interpretation of results

The images should be carefully inspected for artefacts and when these occur, this should be noted. If attenuation correction is used, the profile through the uniform section of the phantom should be flat. In general, the image quality should be visually assessed and the contrast of detected lesions noted.

For a further discussion of this test and for image examples taken under different conditions (see Section 3, Ref. [3]).

Limits of acceptability

Different systems will give different results and the same SPECT system will give varying results over time, especially with high count acquisition data. The presence of circular artefacts is a good indication of problems associated with uniformity and may indicate a need for recollection of the uniformity correction data.

Poor detection of lesions can be associated with a centre of rotation or energy window problem.

Any significant degradation in performance, for example, loss of visibility of small spheres or rods, between the reference test and a routine test needs to be investigated further.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

4.4. OPERATIONAL CHECKS

4.4.1. Check of routine function and centre of rotation offset

Purpose of test

To ascertain the proper function of a SPECT system and to ensure that the centre of rotation offset is minimal.

Materials

A point source of ^{99m}Tc, together with some means of suspending it in air within the field of view, for example, by using a long ruler, as used in the test of tomographic resolution in air (Section 4.3.4).

Procedure

(1) Using a spirit level, ensure that the camera is accurately aligned so that the head is parallel with the axis of rotation, i.e. that the head is not tilted.
(2) Suspend the point source in air at about 10 cm from the axis of rotation and within about 2 cm of the centre of the field of view, axially.
(3) Perform a normal tomographic acquisition using the normal digital matrix size used for tomography, collecting about 10 000 counts at every angular position. An acquisition consisting of 32 angles over 360° is adequate for this test.
(4) Generate a sinogram through the point source.
(5) Sum all the projection images into one image or observe the projection as a cine.
(6) Perform a centre of rotation check according to instructions provided by the manufacturer, if available.

Data analysis

(1) Examine the sinogram. It should be a smooth curve.
(2) Examine the summed projection image or the cine. The point source images should fall on a horizontal line and the cine should move on one horizontal line.
(3) Perform an analysis of the centre of rotation check according to instructions provided by the manufacturer, if available.

Observations

This test is intended to be performed as a reference test, at weekly intervals and if a problem is suspected.

The main purpose of this operational test is to do a quick check that the centre of rotation offset and head alignment are acceptable.

Interpretation of results

Any result that appears to be abnormal should be investigated further.

Limits of acceptability

If the results of the centre of rotation check are outside the values specified by the manufacturer, the test should be repeated. If the test remains abnormal, further action should be initiated.

Conclusion

Record whether or not the results confirm acceptable performance. If not, indicate the follow-up action taken.

5. CONSIDERATIONS FOR MULTIPLE HEAD SYSTEMS

5.1. INTRODUCTION

Scintillation camera systems are available in different multiple head (also termed multihead or multidetector) configurations, for example: dual head fixed 180°, dual head 90°, dual head variable angle, triple head fixed 120° and triple head variable angle.

This section addresses the special quality control aspects of these multiple head systems.

A number of the tests that are easily performed on single head cameras are difficult to perform on multiple head cameras because of the configuration of the heads or the difficult physical access to the heads. For example, intrinsic uniformity and spatial resolution measurements for some fixed dual and triple systems are difficult or impossible since the required source–detector distance

of five crystal diameters cannot be achieved. In such cases, the manufacturer's recommendations must be followed to test adequately certain aspects of these systems.

5.1.1. Multiple head camera planar tests

Section 2 describes the tests necessary for planar camera systems. When a planar system has a second (or third) detector head, essentially all the tests of Section 2 must be repeated for each head. In Section 4.1.3, the following parameters are listed as being applicable to the general scintillation camera (whether or not it is used for SPECT):

(1) Energy resolution (Section 2.1.6.2);
(2) Flood field uniformity (Section 2.1.6.3);
(3) Spatial distortion (Section 2.1.6.4);
(4) Differential ADC linearity (Section 6.1.2.3);
(5) Integral ADC linearity (Section 6.1.2.3);
(6) Spatial resolution (Section 2.1.6.1);
(7) Count rate response (Section 2.1.6.6).

In order to fully characterize a multiple head system, each of these parameters should be determined for each of the detectors. This will increase the amount of work and camera time correspondingly. All the tests described in Section 2.3 should be repeated for each head except for:

(1) Section 2.3.12: Test of basic computer timing:
This test is a system test performed with all detectors. Each head should be exposed equally to the source, with no intervening material between the source and detectors. Each head should collect about the same number of counts during this test. If the disparity is more than 10–15%, the discrepancy should be investigated by repeating the test with each head separately.
(2) Section 2.3.13: Test of computer timing in dynamic acquisition:
Extension to multiple heads is straightforward.
(3) Section 2.3.14: Test of ECG gated acquisition:
Extension to multiple heads is straightforward.

5.1.2. Multiple head tomographic cameras — non-tomographic parameter tests

5.1.2.1. System planar sensitivity

Tomographic systems can place special demands on the matching of detectors in a multiple head system. In particular, pixel size, detector and collimator alignment (centring) and system sensitivity for each detector are especially important. The test described in Section 2.3.9 (test of system planar sensitivity) must be repeated for each detector. The detectors should match within 3%. If they are not closely matched, artefacts can be seen in SPECT studies and these are associated with having detectors of significantly different sensitivities. The artefact usually manifests itself in the sinogram; one section of which will have reduced intensity, which will result in a corresponding area of reduced counts in the reconstructed slice.

5.1.2.2. System count rate performance

When testing the intrinsic count rate performance (Section 2.3.11), consideration should be given as to whether high count rate studies, for example, first pass cardiac studies, will be performed with the system. If such studies are contemplated and more than one detector is going to be used, then all the heads that are to be used in the high count rate acquisition should be activated and tested together. In some multiple head systems, the count rate performance of a single head is degraded when a second or third detector is used simultaneously. This is caused by delays in transferring data from the detectors to a common data bus in the computer. See Section 2.3.18 for further information.

5.1.3. Multiple head tomographic cameras — tomographic parameter tests

In Section 4.1.4 on performance characteristics, the following performance parameters are listed for tomographic SPECT systems:

(1) Slice thickness;
(2) Tomographic signal-to-noise ratio;
(3) Tomographic contrast;
(4) Tomographic uniformity;
(5) Tomographic (in-slice) resolution;
(6) Linearity of tomographic response;
(7) Quantitative accuracy in tomography;

(8) Precision of estimation of the centre of rotation;
(9) Tomographic sensitivity — slice and volume.

When determining these parameters of a tomographic scintillation camera system, separate tests involving each detector would not be required. Only when investigating abnormal performance would a decision be made to test these parameters for individual heads. It is usually possible to acquire the data from all detectors in a single acquisition. Then, at the time of reconstruction, any one, two or three detectors can be included in the reconstructed file. This is sometimes useful when troubleshooting difficult problems.

In Section 4.3 on acceptance and reference tests, there are nine tests used to check calibrations and to assess SPECT performance. The tests that would have to be repeated for each detector are those in Sections 4.3.1 (physical and mechanical inspection of the SPECT system) and 4.3.2 (test to determine the absolute size of a pixel). Some manufacturers may provide special fixtures, source holders and protocols so that pixel size may be conveniently measured and calculated for the multiple detectors.

5.2. TEST SCHEDULE

The test schedule for multiple head camera systems should follow the schedules established in Sections 2.2 for planar cameras and 4.2 for tomographic camera systems.

5.3. ACCEPTANCE AND REFERENCE TESTS

In the sections below, each of the tests of Section 4.3 will be discussed in relation to the testing of individual heads of a multiple head tomographic system. Planar parameters will not be discussed further. For all the tomographic acquisitions, either a $360°/n$ (where n is the number of heads) or 360° orbit may be used. A 360° orbit is generally preferred, as it allows each detector to acquire a full set of data that can be individually reconstructed. The images then become available for interpretation.

5.3.1. Physical and mechanical tests of the multiple head system

There are 15 steps for the procedure described in Section 4.3.1: Physical and mechanical inspection of the SPECT system, of which the following steps should be performed for each detector:

(2) Rotate all detectors as described.
(5) Check mechanical centring for all heads.
(6) Check that the Y axis of each head is parallel to the axis of rotation for all heads.
(8) Check all indicators for all heads.
(12) Check each collimator of each head for visible damage.

5.3.2. Absolute pixel size

The procedures given in Section 4.3.2: Test to determine the absolute size of a pixel, would have to be repeated for each detector in a system. The pixel sizes for the detectors should be within 5% of each other. It is extremely important that the pixel sizes in the X and Y directions for all detectors are matched. Manufacturers should supply software that will assist the user in maintaining confidence in the matching of the detectors with regard to this parameter. Independent checks of this software, especially on acceptance testing, are necessary.

5.3.3. Tomographic uniformity of the system

In Section 4.3.3: Test of tomographic uniformity of the system, a test for tomographic system uniformity is described that is usually performed with all detectors simultaneously. The only change to the procedure described in Section 4.3.3 is that the orbit may be 360º/*n*. With single head systems, circular acquisition orbits will produce circular (ring) image artefacts, so-called 'bullseye' artefacts, when there are uniformity problems. With multiple heads, these circles will take on the appearance of arcs (see Fig. 48). On occasion, for troubleshooting purposes, 360º orbits should be acquired to help isolate a problem in a single detector. Data from individual detectors may then be reconstructed and images of the tomographic uniformity phantom can be examined.

FIG. 48. Partial ring artefact from a dual head SPECT system. Dual detector SPECT system with detector heads at 180° from each other, 360° total angle of rotation (each detector rotates through 180°), circular orbit. SPECT acquisition of a uniform cylinder, no attenuation correction and no uniformity correction applied. Two images from the set of transverse slices are shown. Different non-uniform artefacts are seen in the transverse slices. In the left image, the centre of the innermost ring corresponds to the centre of rotation. Note that the phantom was positioned off-centre so that this ring artefact does not correspond exactly to the centre of the phantom. Other artefacts are semicircles (arrowed) centred around the axis of rotation and not full rings as would be expected from a 360° rotation of a single detector head SPECT system. In multiple head SPECT systems, each head will contribute to the non-uniformity of the resultant SPECT uniformity. Since the non-uniformity in each head will be different, the resultant pattern observed in the reconstructed images will not be full rings but partial rings (see Ref. [3]).

5.3.4. Tomographic resolution in air

In Section 4.3.4, a test is described that is usually performed with all detectors simultaneously. Either a 360º/*n* or a 360º rotation may be used. On occasion, for troubleshooting purposes to help isolate a problem in a single detector, individual detectors may be used to acquire and/or to reconstruct data from the point source. When performing this test, it should be borne in mind that the system resolution is dependent upon the distance from source to collimator. It is important to ensure that all heads are set at the same distance from the axis of rotation.

5.3.5. Tomographic resolution with scatter

In Section 4.3.5, a test that is usually performed with all detectors simultaneously is described. On occasion, for troubleshooting purposes to help isolate a problem in a single detector, individual detectors may be used to acquire and/or to reconstruct data from the tomographic phantom.

5.3.6. Centre of rotation and alignment of axes

This test takes on different names depending on the vendor of the multiple head camera system (examples of names include multiple head registration and Rotax). It is a very important test that is always part of the installation procedure and is performed periodically by the user at intervals recommended by the manufacturer (usually weekly or monthly on new systems). The manufacturer provides detailed instructions. Special fixtures to hold multiple sources are usually involved. Pixel size, head registration, as well as centring are determined and corrections calculated and applied using the software supplied by the manufacturer. All detectors are used in this test.

Section 4.3.6 describes a test that can be used if the test provided by the manufacturer is suspect or if the test described in Section 4.3.4 has failed. It is important to ensure that all heads are set at the same distance from the axis of rotation when performing this test on a multiple head system. On occasion, for troubleshooting purposes to help isolate a problem in a single detector, individual detectors may be used to acquire and/or to reconstruct data from the point source.

5.3.7. Slice thickness

The test described in Section 4.3.7 is performed only with all detectors activated. Make sure that all heads are set at the same distance from the axis of rotation when performing this test on a multiple head system.

5.3.8. Variations of uniformity of sensitivity with angle

The test described in Section 4.3.8 may have to be performed for each detector in a multiple head system. This test was found to be very important for detectors built before about 1990. Since then, greater attention has been paid to magnetic shielding for the PMTs. Some instances, such as a scintillation camera installation located near a magnetic resonance imaging scanner, may require that this test be performed. In such a case, the test should be repeated for all detectors.

5.3.9. Total performance test

Typically, the test in Section 4.3.9 would be performed with all detectors. On occasion, it may be informative to perform this test for individual detectors when troubleshooting system uniformity, resolution or contrast problems.

5.4. OPERATIONAL CHECKS

Routine operational tests are described in Section 2. These apply to each head of a multiple head camera. These operational checks comprise:

(1) Check of collimator and detector head mountings;
(2) Check of energy calibration of PHA;
(3) Check of flood field uniformity and sensitivity;
(4) Check of background count rate.

As indicated previously, these tests must be performed daily.

With instruments as complex as multiple head tomographic scintillation cameras, it is certain that failures will occur from time to time. The worst type of failure is one that manifests itself gradually and insidiously in the system. By carrying out routine checks of the SPECT calibrations (centre of rotation and uniformity), it is possible to verify quickly, proper SPECT system performance.

Section 4.4.1 should be performed on multiple heads as well as on single head systems at regular intervals, such as on a weekly basis.

Section 4.3.9 is another routine performance test that is used to verify that the system is free from uniformity artefacts and yields reproducible images. The total performance check should be performed at quarterly or half-yearly intervals. If the system is used with different radionuclides, the total performance check should be performed with different radionuclides, perhaps by employing two phantoms, one for short lived radionuclides and one for long lived radionuclides. The phantom need not be sophisticated or complicated. Any tightly sealed jar of water greater than 15 cm in diameter will suffice to prove that reconstructed system uniformity is adequate.

All routine tests should have the instrument configured as it would be used for patients. This would normally require that all detectors be used for these operational tests.

These tests and additional examples are discussed further in Section 3 of Ref. [3].

6. CAMERA–COMPUTER SYSTEM

6.1. TECHNICAL ASPECTS OF THE SYSTEM INTERFACE, DIGITAL ACQUISITION AND PROCESSING SYSTEMS

6.1.1. Introduction

Where a data processing system is in use, it should be remembered that poor performance of the scintillation camera–computer interface can render the results from even a high quality imaging device completely useless. In addition, failures with this part of the system can be much more difficult to detect than faults with more conventional equipment. While such problems can cause a loss of image quality, an associated loss of accuracy of quantitative data may be less obvious.

For the purposes of this section, after considering how the scintillation camera–computer interface works (Section 6.1.2), quality control of the interface will be subdivided into four subject areas, namely:

(1) Tests of static performance (Section 6.1.3);
(2) Tests of dynamic performance (Section 6.1.4);
(3) Tests of special functions, e.g. gated acquisition (Section 6.1.5);
(4) Tests of the computer software (Section 6.1.6).

Thus, two sections will be devoted primarily to considerations of the interface, while the two last sections will be concerned with what happens before and after the interface. It is difficult to isolate tests of the interface from conventional tests of the scintillation camera. In fact, it has been suggested that certain tests originally devised for the interface may prove better tests of overall scintillation camera performance than some originally chosen specifically for that purpose. To this end, one of the aims of this section is to suggest quantitative tests, making use of a computer, by means of which performance indices can be defined. This would enable objective rather than subjective statements to be made about performance.

A major problem in quality control/assurance is the use of subjective tests (e.g. examining the image of the flood field uniformity), making subjective assessments and then failing to take any action. It is suggested that objective criteria be used, for example, deciding that a scintillation camera must not be used if a given performance parameter is outside a preset tolerance value. This objective approach requires the definition of action thresholds.

In the USA, protocols with objective criteria were developed for factory testing of equipment by the NEMA [8] to determine if published specifications are met. The NEMA performance standards are recognized throughout the world for the testing of nuclear medicine equipment.

This section is concerned primarily with acceptance and routine testing. The use of a logbook is essential. Unexpected events tend to happen, especially with computers, which must not be ignored. Firstly, 'what happened' should be logged with as much detail as possible. Then, if it is possible, efforts should be made to find out why the unexpected event occurred. Much trouble can be eliminated by this procedure by following up seemingly small errors; potentially major problems can be identified early and appropriate action taken. Thus, while specific tests are proposed in this section, quality control of computer systems involves an attitude of mind, refusing to take things for granted and being watchful for potential problems, pursuing them patiently, using a certain amount of imagination in finding an explanation for what is often seemingly inexplicable.

6.1.2. The scintillation camera–computer interface

In general, all modern scintillation camera systems are connected to some form of computing system; purely analogue systems are becoming increasingly rare. A computer is required for handling dynamic studies and for tomography. However, in many centres, the processing and display of nuclear medicine studies are handled by the scintillation camera–computer system. The reporting and archiving are done in a picture archiving and communication systems computer network.

There are many variations in the details of signal processing of scintillation camera signals. These are discussed, to some extent, elsewhere in this publication under the topics on spatial distortion correction, energy correction, etc. In general, the camera electronics must find the location of the interaction of an event within a detector by processing the output of a number of signals coming from photomultipliers attached to the scintillating crystal. Thus, while this operation of the scintillation camera itself may be analogue, digital processing of these signals is also common. Here, digitization of the initial analogue signals can occur at different stages of the signal processing chain, including encoding the output of every photomultiplier or a cluster of photomultipliers directly and then performing the position calculation in a purely digital manner. The electronics within a scintillation camera head is thus extremely variable in design.

Nevertheless, some form of interface between the camera head electronics and the external computing system is required. Often, this interface is not accessible to the user because it is tightly integrated into the system. In other cases, the interface will be obvious because it is supplied by a third party, such as a computer vendor or an interface vendor other than the camera manufacturer.

In any case, the purpose of this interface is to capture, for every event that it is desired to record, a digital value that represents (at least) the coordinates of the detected event. This digital value may be stored directly (list mode) or used to form digital images directly (frame mode).

In a purely analogue camera, the X and Y outputs, essentially the same as those used to drive the analogue display, are fed into ADCs. The conversion needs to be triggered (the time at which the conversion is to take place needs to be established) by a signal that is also available on most cameras as an output signal and is usually termed the 'gating' signal. It is a digital signal (taking only logical values 0 and 1) and often follows transistor–transistor logic (TTL). Figure 49 shows how the trigger signal must occur at precisely the correct time to have the ADC measure the X or Y position signal. Thus, in the simplest system, the change of state of the gating signal is used to indicate to the ADCs the point in time at which to perform the digital conversion (determination of the digital value corresponding to the amplitude of the analogue signal).

In practice, many variations can be found. The gating signal may not correspond to the expected voltage values that are anticipated for a TTL signal.

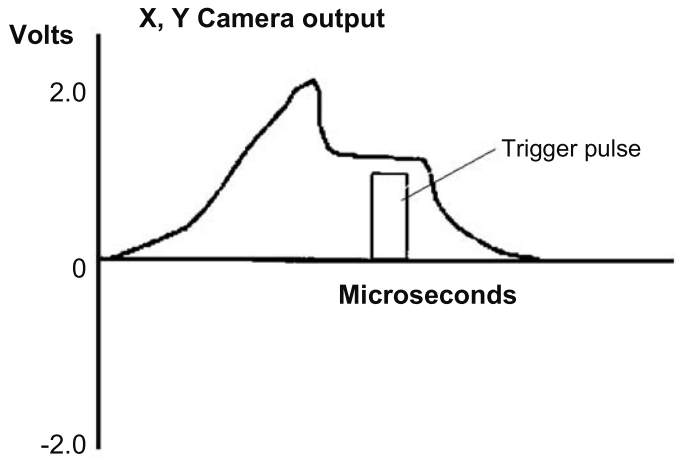

FIG. 49. The relation between the camera analogue position signal and the trigger pulse.

The time for conversion might correspond to a rising edge or a falling edge, or be delayed in some manner. The rise time might be inappropriate, or overshoots or undershoots might be observed. Many of these problems can be managed by including some signal conditioning in the interface, in particular, by the addition of sample and hold amplifiers.

In particular, the gain and offset of these voltage levels need to be controlled carefully. The gain is that component which directly affects the size of the image stored digitally; the offset is that component which shifts the image, in other words, centres the image in the display or in the computer matrix.

The values of the voltages that are found on the X and Y outputs themselves may be very variable, ranging from millivolts up to tens of volts. The form of the signal (the shape of the pulse) is likely to be variable, with significant amplitude decrease over time. The polarity is particularly important. Many systems are designed such that zero voltage for the X and Y signals corresponds to a position close to the centre of the field of view of the camera. The advantage of this is that the average voltage level is zero over some reasonable integration time. Many systems will produce voltage pulses that will be either positive or negative, depending on position. Some systems indeed produce two outputs, for both X and Y, positive and negative outputs, which are normally termed X+ and X– (and correspondingly Y+ and Y–).

As a further complication, the output signal may not be a simple voltage pulse, but could be a current pulse instead and needs to be handled appropriately. In any case, all output signals need to be appropriately matched and loaded.

One important variation occurs when the gating signal is either missing or considered unsuitable. An example of the latter is when energy information is required in addition to positional information, for example, for external energy correction. In this case, the analogue Z or energy signal must be used if it can be located. Energy windowing can be performed by gating the whole process by a signal from the camera's PHA, or by performing the PHA operation externally to the camera.

List mode data with energy information are also found in which the data are stored in the computer as triplets of values, that is, X, Y and Z (energy) information for every event. A purely digital interface, where the digital values of the X, Y (and perhaps Z) signals are produced internally within the camera and are available externally, is in essence a simpler device. A parallel connection is made between the camera and the external device, where again some signal conditioning is required to match voltage levels and load. This serves to synchronize the system, for example, select the appropriate time at which the transfer should happen. Normally this will involve a 'handshake',

that is, an exchange of signals of the type 'ready to transfer', 'transfer', 'transfer completed'.

In addition to capturing X, Y and, in certain cases, Z energy information, the interface has other functions and signals that need to be controlled. The timing of the acquisition must be managed so that the total acquisition time, or the acquisition time for one frame, can be determined with sufficient precision, for example, to the nearest millisecond. This is particularly important for ECG gated studies where each heartbeat is divided up into a number of frames, which are then summed.

Thus, an additional feature of most scintillation camera interfaces is the capability to handle the signals coming from an ECG trigger. Typically, this is performed by capturing the digital (TTL) signal that is produced by a conventional ECG trigger device, where such a pulse is produced when the R wave is detected. As before, the polarity, rising or falling edge, and shape of this pulse must be taken into account. An alternative is to capture the analogue ECG signal directly, which is subsequently processed by the interface to produce the desired gating.

6.1.2.1. Corrections

Many of the conventional corrections carried out in the scintillation camera head can be performed either by the interface, or in the computer system attached to the head, and there are several differences.

Processing in the head allows analogue or digital or hybrid processing and is usually fixed in firmware, although correction maps need to be established, normally in an independent acquisition step, and then downloaded to the head. The use of special purpose hardware, such as digital signal processing chips, is not uncommon.

Processing in the interface is normally entirely digital. Again, special purpose hardware in the interface (digital signal processing chips, etc.) may be used to improve speed. The computer code is often easily reprogrammable, but again, correction maps need to be acquired independently. These maps may be stored directly in the interface, or in memory available on the host computer.

Processing in the host computer can be divided into two types: list mode and frame mode. List mode processing, which might be performed on the fly (during acquisition) but which is much more commonly a post-acquisition operation, is somewhat time consuming and may involve delays between acquisition and the availability of data to display. Within the limits of the precision of the list mode data (number of bits), anything that can be performed in the interface can be performed post-acquisition. An advantage of post-processing is that several different correction schemes may be employed and

compared. Frame mode correction can also be used. It is more limited in the types of correction that can be performed, but is generally much faster. Energy correction (and scatter correction) can be performed if multiple energy frames exist (e.g. frames for different energy windows). Spatial distortion correction can be performed with limited accuracy, depending on the matrix size used. Counts are moved from one pixel to another; the greater the matrix size, the smaller the error that will result. Only matrix multiplication uniformity type correction can be performed and should only be used to remove residual variations in sensitivity and not uniformity caused by other problems such as energy variations.

6.1.2.2. The interface, how it works and what can go wrong

Before starting to devise tests for the interface itself, it is worth considering how it works and what kinds of error can occur. These, in fact, depend on how the interface has been designed and there are many variations. In general, with respect to static image quality, the major problem that is likely to occur is that of poor differential linearity, rather than poor integral linearity, terms which are discussed below.

The extent to which such errors can occur is, in particular, a function of which type of ADC is used. There are at least three major types in use: Wilkinson ramp, sequential approximation and flash. In addition, various refinements such as bit randomization may also be employed.

The basic principle of a Wilkinson ramp ADC involves starting at some time zero, at which time a clock is started and a ramp voltage is generated that is compared with the input voltage which is to be digitized. A comparator decides when the ramp is greater than the input voltage and the clock is stopped. The number of pulses recorded from the clock represents the digital value of the input signal.

A sequential approximation ADC operation can be summarized in the following set of steps:

(1) Set the most significant bit in a register;
(2) Generate the voltage corresponding to the number in the register;
(3) Compare it to the input reference signal;
(4) If it is greater, reset the current bit;
(5) If there are any bits left to test, set the next lower bit and go to step (2).

Thus, at the end of n operations, where there are n bits in the register, a digital value corresponding to the input signal is generated. The value of n for scintillation camera–computer ADCs is often 12 or more, but generally truncated to 8. A randomized ADC follows the same principles, but the values

of the least significant bit(s) are randomized in some manner to improve the differential linearity. This corresponds to 'injecting' noise into the system.

In a flash type of ADC, instead of using a single comparator, a separate comparator is used for each voltage level and very fast conversion times can be achieved. The problems of differential and integral linearity are similar to those in the sequential approximation ADCs. This type of ADC is rarely used in nuclear medicine since it is expensive and of value only when very fast conversion times are essential.

6.1.2.3. Differential and integral linearity

When a pulse is encoded, it is given a numerical value and it is assumed that the distribution of numerical values is constant (see Fig. 50). In fact, the voltage at which bit value changes occur is not necessarily that expected and thus deviations from the line of identity can and do occur. The absolute value of the difference between the real voltage and the value it is believed to be is termed the integral non-linearity. It increases with voltage amplitude. It tends to be worse for the Wilkinson ramp ADC than for the sequential approximation ADC. However, the slope of the response curve determines the bin size, that is, the probability of an event being included in any given bin and thus recorded as having a particular value. The slope of the curve is directly related to the size of boxes along the Y axis, which directly reflects the differential non-linearity. In Fig. 50, it can be seen that considerable variations occur. In practice, these give rise to vertical or horizontal stripes in scintillation camera images, or in two dimensions, a tartan pattern.

6.1.3. Static tests

Static tests are those used to check the performance of the scintillation camera interface when collecting a static image. They are primarily concerned with uniformity, linearity, etc., and are subdivided into acceptance and routine tests as follows:

Acceptance tests

(a) Test of interface uniformity;
(b) Test of interface spatial distortion, resolution and integral linearity;
(c) Test of interface differential linearity.

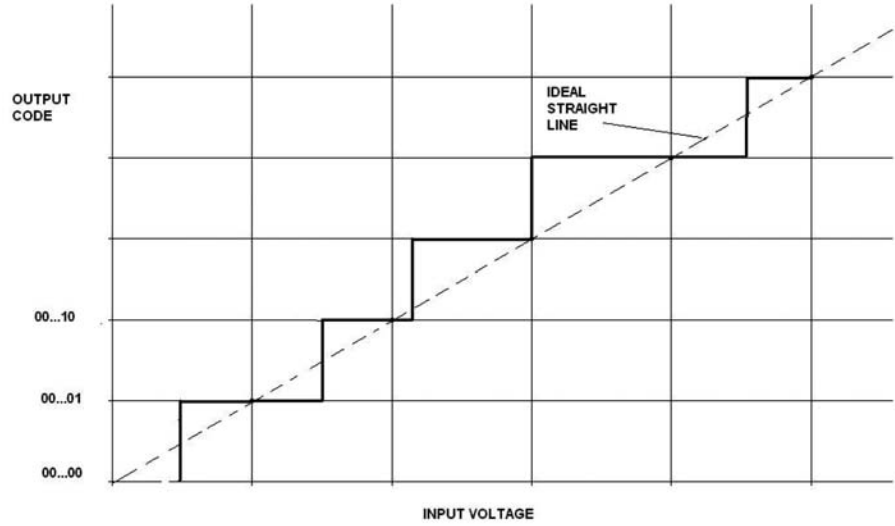

FIG. 50. ADC linearity. The ideal values fall on the straight line — the line of identity. In this example, only the second code falls on the ideal line. Also, the analogue bins are larger than others, leading to poor differential linearity and ultimately undesired lines in the scintillation camera image.

Routine tests

Test (a) above should be repeated on a daily basis.
Test (b) above should be repeated on a monthly basis.

6.1.3.1. Test of interface uniformity

The test of interface uniformity is intended to determine whether the digital image of a uniform source of radioactivity is uniform. While it is simple to perform, it is probably not as good a test as the use of a coarse OHTP as described in Section 6.1.3.2. The test checks whether the computer interface does, in fact, function, checks whether or not the image is centred in the computer frame and provides an indication of whether the differential linearity of the interface is acceptable. Obviously, it is important that the image of a flood source collected on the computer is uniform, as errors may distort quantitative measurements and reduce image quality. However, a number of other potential areas of malfunction can be readily detected with this test.

A point source or flood source used for the conventional scintillation camera quality control test of uniformity is used (see Sections 2.1.6.3 and 2.4.3). This test provides a quick quantitative analysis of the daily routine check of the scintillation camera uniformity. An appropriate protocol is as follows:

(1) Ensure that the collimator normally used for this test (the collimator most commonly used) is in place and that the camera orientation is normal.
(2) Collect a digital image in a 64 × 64 or 128 × 128 matrix. The number of counts should be about 5 million, depending on the field of view size.
(3) Note the start and finish times of the computer collection and the camera collection and check the number collected by the computer.
(4) Display the resulting image.
(5) Record the value of the pixel with the maximum count (C_{max}) in the image within the (UFOV) and record, if possible, the mean count (C_{mean}) within the UFOV.
(6) Find the value of the pixel with the minimum count (C_{min}) also within the UFOV.
(7) An estimate of the NEMA integral uniformity [8] can now be calculated as $(C_{max} - C_{min})/(C_{max} + C_{min})$. This value should be plotted.
(8) As a routine test, compare the image with the most recent reference image. This may be performed as follows. The two images should be normalized to the same mean value. This will normally be approximately true if they contain the same number of counts. The two images should be subtracted and the resulting image inspected. If possible, this subtraction image should be normalized by multiplying it by 100 and dividing by the mean of the reference image that will convert it to an image of percentage differences between the two images. This image is a very sensitive indicator of change. The most recent reference image that is appropriate to use will depend on the system. One possibility is to use the image from the preceding day. An alternative is to use that obtained when this test was performed at acceptance. The latter is susceptible to error because of lateral shift of the image over time. The former is insensitive to gradual slow change of the system. If both types of error are occurring, it may be appropriate to compare the new uniformity image with both types of reference image.
(9) Plot a vertical and a horizontal profile.
(10) Note the pixels where the profiles reach 30% of the mean value, or if the mean, C_{mean}, is not known, those values defining the top edge, bottom edge, left hand edge and right hand edge may be used for calculating the size and centre of the image when the normal scintillation camera image is a circle inscribed within a square. These values should be recorded.

It should be noted that all of the above operations can be performed by a single, special purpose computer program, as can the further operations suggested below. However, all the preceding operations can also be performed using conventional software, although some difficulty may be experienced defining the UFOV. With special purpose software:

(1) Generate a map of the percentage excursion of the image from the mean value. This means setting all values outside the UFOV to zero, multiplying all values within by 100 and dividing by C_{mean}.
(2) Generate a histogram of excursions from the mean, i.e. the percentage of pixels that lie within ±5% of the mean, within ±10%, etc. Statistical tests can be performed on such histograms.
(3) Calculate the coefficient of variation, being the standard deviation of all pixels within the UFOV, divide by the mean and subtract the expected pixel error, i.e. the coefficient of variation due to Poisson statistics, which equals $1/\sqrt{C_{mean}}$.
(4) Store the data. When performed as an acceptance test these data are critical since they will be used as reference data for the future. Depending on the system, it is recommended that reference images also be stored on a monthly basis to provide short term reference images.
(5) Visually inspect the raw data and, when available, the map of percentage excursions. Check for counts outside the field of view of the camera. Check for a central hot spot. In particular, after increasing the contrast of the image, check for horizontal and vertical stripes that are evidence of problems of differential non-uniformity.

The estimate of NEMA integral uniformity should be compared with that obtained at acceptance when the system was performing satisfactorily. If the value of the NEMA uniformity is greater than approximately twice that at acceptance, it may be assumed that a problem exists. The first action should be to rotate the scintillation camera and repeat the test. If the non-uniformity rotates, then the problem lies with the camera or with the flood. If the non-uniformity remains fixed then the problem lies with the interface.

At acceptance testing, this test should be repeated with the finest acquisition matrix that is available (e.g. 512 × 512 or 1024 × 1024) to check the higher order bits in the ADC. At least 10 million counts should be collected and any image structure, such as straight light or dark lines in the image indicating poor differential uniformity, should be noted.

6.1.3.2. An alternative test using the OHTP

This test is intended to be a reference test to be performed at acceptance and at half-yearly intervals.

This test uses a coarse OHTP and measures the camera and camera–computer interface linearity. Special software is required, as described below. The hole spacing of such a phantom is important. It has been demonstrated that a loss of information occurs if the hole spacing is coarser than about 2 cm. Too fine a sampling, using, for example, a hole spacing of less than 7 mm, means that errors in the calculation of FWHM in particular will occur since the images of individual holes cannot be completely isolated. It is also important that the holes be aligned in some regular manner with respect to the PMTs themselves. This is important in that, for example, maximal spatial distortion occurs between PMTs and minimal spatial distortion over PMT centres. Thus, information that is more meaningful can be collected if the hole spacing is fixed with respect to PMT positions and spacing. An appropriate protocol is as follows:

(1) Remove the collimator and attach the OHTP and lead mask ring (if available). (The lead mask is intended to limit the field of view to that normally obtained with the collimator in position.) Place a small ^{99m}Tc source on the axis of the collimator at a distance of at least 1 m, such that the overall count rate is less than 20 000 counts/s.
(2) Collect an image at the finest matrix size available (e.g. 256 × 256, 512 × 512 or 1024 × 1024), with at least 1 million to 4 million counts.
(3) Using a suitable software package designed for the task, identify the centres of gravity of the various holes that lie in a regularly spaced manner. It is possible to calculate an index of distortion.
(4) Determine also the FWHM of the images of the various holes.
(5) Note the variation in peak counts over the images of the various holes and record the integral of the counts corresponding to each hole.
(6) In this way, it is possible to determine the spatial distortion, resolution and point source uniformity over the whole field of view.

Variations in position of the point sources can be caused by spatial distortion of the camera, or by changes in the integral linearity of the camera. Quantitative measures of the mean resolution and its variation and mean integral of counts for each hole position and its variation should be noted and compared with those values found at acceptance.

6.1.4. Dynamic tests

The dynamic tests are those used to check the performance of the scintillation camera interface when the system is being used in dynamic mode. For appropriate tests to check the correct functioning of the computer, see Sections 2.3.12 and 2.3.13.

6.1.5. ECG gated acquisition

The purpose of this test is to check that data acquisition is triggered correctly from the ECG signal and that the ventricular volume curve is clinically valid (see Section 2.3.14).

6.1.6. Software tests

6.1.6.1. Introduction

The software that is used as part of the acquisition procedure is a major component in producing clinical results in diagnostic medicine. Such software has become increasingly important in the handling and manipulation of data and in the interpretation of the results of the investigation. In addition, results from prior studies must be capable of being reliably retrieved, as patient management involves the comparison of results at some given time with those obtained previously. An example of this is monitoring the size of a tumour. Finally, given the almost universal use of software as part of the instrumentation for acquiring, manipulating and interpreting clinical information, the user interface involved should be sufficiently robust such that human errors can be minimized. Thus, software quality assurance concerns a number of different aspects: a robust user interface, reliable data storage and retrieval, acquisition and correction of image data and production of results that are both accurate and precise. There are four major types of error that may be the result of software faults: (i) failure (the information is lost), (ii) reliability (random error is introduced and precision is poor), (iii) accuracy (bias is introduced) and (iv) 'fuzziness' (the meaning is unclear and can be misinterpreted).

At present, there is no satisfactory method of ensuring the correct performance (or even specification) of nuclear medicine software; the complexity of the software is such that exhaustive testing is impossible [18]. Thus, the only possible valid tactic is to divide the overall problem into components. A practical procedure would be to divide software testing into distinct stages (or levels using the ISO model [19]). The first stage would be

that of verifying accurate data acquisition, for example, by using phantoms. A second stage would then be to use simulated data for testing processing algorithms, where the known ('true') values are compared with observed (computer) results. A third stage would be to generate and thereafter test systems using clinically validated real patient data, as in the program [20]. A fourth stage would be to stress the system using clinically extreme and intermediate data to test for failures. Overall, a method for assessing clinical performance as a function of software components must be developed and methods for ensuring quality assurance in the process of software production, such as ISO9001 [21], need to be established and controlled.

6.1.6.2. General advice

A well-known expression states that, "while the number of known bugs is finite, the number of unknown bugs is infinite". Thus, the potential for a data processing system to produce errors is almost unlimited and care must be taken at all times to ensure that the indicated results are reasonable. However, there are some guidelines and tools that may be helpful and useful in limiting the number of errors and in checking a system to verify that it is reasonably well behaved. If the basic checks on the interface are reasonable and the timing tests for dynamic acquisitions and gated studies are satisfactory, the next set of tests concerns simple types of data manipulation. For example, a good, simple test involves defining three regions of interest for a dynamic study: one large, and two dividing the large region of interest into two equal parts. In order to do this, the time–activity curves must be generated and the curves from the small regions of interest must be added together. It is then necessary to check that the result is identical to that from the large region. Likewise, the values at various points within a matrix must be checked against the values printed out on a profile through the matrix, taking care to use the correct convention for defining coordinates. Some systems term one corner the point X = 1, Y = 0, while others refer to the same point as X = 1, Y = 1. In addition, matrix elements may be displayed at the top left, bottom left, or in some other manner. It is important to check the computer output and not to accept it at face value.

In general, there are two useful types of data that serve to validate computer programs: validated clinical data and simulated data. In both cases, inputs are created for which the result is known, in order to check the determined values against the true value. This type of test is especially useful after a software upgrade so that the previous validated values can be checked against values obtained with the new software.

It is recommended that a set of validated clinical data be established or obtained, if possible, for each type of clinical procedure. Whenever the system is changed or modified in a significant manner, it should be possible to test the new software on known cases. The use of simulated data can also be helpful. For example, a program that generates a matrix of values from 1 to 4096 for a 64×64 matrix can be used to test the display, the profile generation program and programs performing matrix manipulation, etc. More complex simulations, such as those that can be used for testing cardiac programs, may also be useful. An example of such a simulation would be when the frames from a simulated cardiac study can be obtained by generating four ellipsoidal chambers varying in size as a function of time, blurring them with the collimator response function, and adding Poisson noise and background to simulate other organs. In particular, it is strongly recommended that a program for generating simulated tomographic acquisitions be used for testing tomographic software. The program for generating such data is likely to be approximately 50 lines of a suitable computer language. Unfortunately, such validated, simulated acquisition software is not readily available.

Whenever the system is upgraded, an acceptance test should be performed to the extent possible. Each of the clinical procedures in use, or each procedure that is likely to be used, should be tried and any differences between what is performed and what is described in the documentation should be recorded. If a set of validated data exists, the programs should be tried out on these data and the answers should be checked. If the answers are incorrect, it is necessary to determine why.

There is a general trend towards standardization in software generation, in the use of software (and the user interface) and in the generation of standard results. The use of physical phantoms provides little help in these areas, and software phantoms and the collection of large series of validated clinical test data sets are fundamental.

A more systematic approach

Firstly, it is necessary to identify the various system components that need to be tested. These can be divided into the following groups of functions:

- Interface to the scintillation camera/acquisition device;
- Acquisition protocols;
- Low level processing;
- Clinical protocols;
- Complex image processing;
- Quality assurance software;

- Links to other systems;
- Display and hard copy;
- Archiving;
- Safety/noise;
- Overall clinical performance.

These topics are considered individually in the following sections. It should be noted that these items, with the exception of the last one, are all components of any complete system. However, the last item, overall clinical performance, includes tests of the complete system comprising all the individual components. Such tests are therefore 'holistic' and although they provide little information about what kinds of faults may exist, they are essential for ensuring that the complete system also functions as specified or, more accurately, as desired, with all its interacting components.

In each section, one example of a test in that area is included, as well as an example of a corresponding problem that has been observed in actual practice.

The interface

Tests should be provided to check for the capability of the hardware interface (under software control) to provide good data. In addition to the hardware tests previously specified, these tests should check for:

- Absence of artefacts (e.g. stripes);
- Accuracy of time information (e.g. dead time/count loss);
- Capability to work with different matrix sizes;
- Stability;
- 'P scope' (persistent scope) functions;
- Correct registration of orientation;
- Corrections (energy/spatial distortion);
- Any other special functions.

An example of a specific test is to check that the image stored in the computer is oriented correctly. When a point source is placed in one quadrant of the scintillation camera field of view, the position of the point source should be determined on the image obtained directly on the camera (if available). Otherwise, the position of the point source should be determined in relation to that obtained by the computer. The labelling of the image in the computer should be checked and the process repeated for other quadrants. If available, the labelling should be checked for other rotations of the scintillation camera

image to ensure that the image acquired is correctly orientated and labelled for all acquisitions.

An example of such a problem has been observed when an update of a well-known manufacturer's software inadvertently created 'mirror images' in the computer.

Acquisition

While the interface tests check the hardware of the acquisition system, further tests are required to ensure that the overall acquisition system is functioning correctly. These additional tests include checking:

- Frame timing (see Sections 2.3.12 and 2.3.13):
- List mode data handling (if list mode is available):
- ECG gated studies (and bad beat) handling (see Section 2.3.14);
- Correctness of associated data;
- Identification of patient information;
- Appropriate handling of errors;
- Correction of errors;
- Handling of operator requests (injection started, scan aborted, second phase initiated, etc.).

A simple test of ECG gated acquisition (as described in Section 2.3.14) is to use a point source with the camera, with an actual person acting as the ECG trigger. To emulate bad beats, the ECG leads can be 'shaken'. The resulting data should be uniform in time. The number of bad beats can normally also be checked. Testing for the accuracy of the trigger itself requires a more 'physiological' phantom.

Many systems reject the beat following the bad beat, but include the bad beat itself since the event is only detected after the bad data have already been included within the summed images.

Low level image processing

Having ensured that the acquired data are appropriate, low level image processing functions must be checked, for example:

- Smoothing and filtering;
- ROI generation;
- Time–activity curve generation;
- Statistical analysis;

- Use time information;
- Arithmetic (frame and curve);
- Curve fitting;
- Robustness;
- Patient archive.

An example of such a test would be that of time–activity curve generation, for example, using known, simulated data for which the results are known. Such data should include different frame time intervals, to ensure that these are handled correctly, and high count rates to verify what might happen if an 'overflow' were to occur. This test must be accompanied by a test of the accuracy of positioning the ROI required to generate the time–activity curve. Many systems show sharp jumps in time–activity curves when different framing rates are used.

Clinical protocols

While the testing of the previous items can generally be performed to a tight specification and protocol, testing the clinical protocols is more difficult. Such clinical protocols include:

- Cardiac protocols;
- Kidney protocols;
- Bone protocols;
- Oncology protocols;
- Thyroid/liver/other protocols;
- Non-standard studies.

These are considered in more detail in the next section. In addition, tests need to be performed with respect to:

- The protocol language itself;
- User maintenance and modifications.

An example of such a test would be a check of the value of the ejection fraction as one of the cardiac protocols. The most satisfactory method is to use either simulated data, where the true results are exactly known, or to use validated clinical data, such as those provided under the COST B2 program. The user establishes the value of the ejection fraction as determined by the clinical software on the system, then compares it with the expected value.

Reproducibility is also an important check. For fully automatic algorithms, it is interesting to shift the data slightly, or to add additional noise.

From one manufacturer to another, there may be a difference of up to ±50% in the ejection fraction obtained from the analysis of the same clinical data.

Complex image processing

Such clinical protocols often include the use of more complex image processing functions. The variability of different systems and the lack of specific knowledge of what is being done often make such testing difficult. Such software includes:

- Automatic ROIs;
- Functional images;
- Factor analysis;
- Routine tomography;
- Advanced tomography;
- Attenuation correction/scatter correction;
- Quantitation;
- Model fitting;
- New software functions (unspecified).

The testing of automatic ROI generation can be performed using software phantoms. However, such phantoms should include difficult (testing) cases to try to establish what conditions cause these algorithms to go wrong. A good example is to include very small and very large images of organs. Reproducibility certainly needs to be investigated, as previously discussed.

Automatic ROI algorithms typically fail in about 5–10% of all cases.

Quality assurance software

Specific sets of tests need to be performed for the quality assurance software itself. It is normally assumed that the quality control software, provided by the manufacturer to quantify the detector performance tests, follows the correct algorithm, for example, NEMA tests. This is often not the case. Tests should be performed to check:

- The NEMA tests, uniformity, resolution, etc.
- Correctness of interpretation of such results.
- Trend analysis (if available).
- Expert systems (if available).

The use of a method such as ISO9001 (and variants) is important. An example of such a check is to compare the results for the computed NEMA integral full and central field uniformity for flood fields where these values have already been established and validated.

Links to other systems

Networking is increasingly important. Checks need to be provided with respect to:

- Patient demographic information;
- Scheduling;
- Patient records and billing;
- Recording of research information;
- Security (hacking);
- Connections to hospital information systems;
- Connections to archival and retrieval systems (picture archiving and communication systems);
- Connections to other nuclear medicine systems (e.g. using Interfile or DICOM);
- Upgrades;
- Networking standards, e.g. throughput and data transfer;
- Network management and error checking.

The most important check in this list is that conducted with respect to patient demographic information. If this is retrieved though a network, the numbers of unlabelled or mislabelled studies should be investigated and the number of studies where the image data are missing should be ascertained. The timing information with respect to the transfer of such information should also be tested. Systems have been observed where up to 30% of patient studies were mislabelled.

Hard copy and display

Quality assurance needs to be performed on devices connected to the computer, in particular on the display. Tests should be performed with respect to:

- Technical performance (e.g. speed);
- Uniformity;
- Dynamic range;
- Readability of text;
- Life (of hard copy);
- Cleanliness.

An example of a simple visual check is to compare the readability of text overlays on the original display and on the hard copy. Likewise, the colour rendition of a colour display and the interpretation of the images themselves should be verified after being converted to the hard copy. Most colour printers do not give a good rendition of a monitor display and can give rise to false interpretation. For further information on this topic, see Section 7.

Archiving

Nowadays, archiving is increasingly being performed by a separate system, such as a file server with one or more large hard disks. Parameters to assess are:

- Throughput and time to store;
- Compression (if used);
- Time to access and retrieve;
- Reliability and robustness;
- Capability to correct errors;
- Lifetime of medium (if possible);
- Portability (can it be connected to other systems);
- Flexibility of retrieval;
- Capacity (and overflow control);
- Backup and retrievability.

An example of a check of archiving would be to assess the time to store a given study while the overall system (network) is quiet and when the system is busy. The data should also be retrieved in both cases to ensure that no data loss

has occurred. The test needs to be performed on an archive file and repeated to confirm reliability.

One well-known manufacturer's system occasionally failed to store individual frames of a study when busy, generating a fleeting (and hence unnoticed) error message, resulting in lost frames in the stored study.

Safety

Some general tests required for all such equipment also need to be performed, including:

- Electrical safety;
- Thermal safety;
- Noise generation;
- Heat generation;
- Capability to clean;
- Security.

Tests of monitor quality, for example, flicker and positioning, are increasingly important with respect to the health and comfort of the users. The systems must be able to tolerate a hospital environment.

Overall clinical performance

The previous tests are all concerned with one specific aspect of the system. The overall performance of the system needs to be checked with respect to the following areas:

- User interface;
- Receiver operating characteristic curve tests (clinical accuracy, specificity, etc.);
- Reporting;
- Reliability (in presence of abnormal cases);
- Additional features (voice, etc.);
- Audit;
- Follow-up;
- Research;
- Teaching.

The most important test in this case is to review the results from specific nuclear medicine procedures and compare the results with the clinical diagnosis (clinical audit). Abnormal sensitivity and specificity are of interest, since they may indicate improper usage of the system or faulty components. Dramatic changes in clinical performance have been observed with different clinicians, with obvious implications.

Conclusions

Errors in software are an important source of clinical unreliability in the use of nuclear medicine procedures. Software is notoriously difficult to test; exhaustive testing is believed to be impossible because of the high level of complexity of the system to be tested. Thus, it is essential to test small components of the system according to well-defined protocols using appropriate test data.

Example of clinical software documentation

The following is an example of the items that should be included with clinical software documentation:

- A version number.
- Name of program and date of version.
- Name of the author, the source and the 'background' of the software.
- A certificate of acceptance by appropriate authorities and contact information of the person/institute undertaking the testing and/or approval.
- Introductory text for each program (or routine), to serve as an information leaflet for users, as well as a guide to the more detailed documentation which follows.
- A summary, describing in a condensed form the clinical purpose of the program(s) and the aim and the field of application (a description of what is being studied and what and how it is evaluated).
- Clinical information such as: what degree of patient cooperation is necessary, preparation of the patient (pre-medication, hydration, etc.), which projections are used, changes in procedures according to age and sex, and radiopharmaceutical data such as chemical form, radionuclide, activity, method of administration and requirements regarding purity (chemical, radiochemical, radionuclidic).

- Physical information (the acquisition device to use): scintillation camera, multidetector system, etc. Special requirements on spatial and energy resolution, uniformity, linearity, sensitivity, dead time, use of multiple windows, etc. Type of collimator used, physiological triggers. The summary should also contain information concerning system hardware and software necessary to run the program, such as the acquisition requirements (number of counts, timescale, matrix size, zoom factor, etc.).
- Personnel: interaction required during acquisition and analysis. Demands on the competence of the operator.
- A set of sample results produced by the program using the supplied test data, with detailed operating instructions.
- List of references.
- Information about software distribution medium and format (source, object, code, etc.).
- Authorization and eventual restrictions regarding use and copying.
- Warnings about misuse and legal liability.

6.2. QUALITY ASSURANCE OF IMAGING PROCEDURES

6.2.1. General introduction

The goal of the diagnostic nuclear medicine procedure is to differentiate between normal and pathological conditions. The results of the procedure should therefore provide images and quantitative data that are suitable for classifying disease in a correct and reproducible way. From the instrumentation viewpoint, the clinical procedure is a combination of the scintillation camera and patient set-up, the data acquisition, the data processing and the final display of results. It is essential to realize that all aspects of the clinical procedure are linked. In order to achieve consistent and correct results, each part of the procedure must be carefully monitored and a standard clinical protocol must be adhered to.

This section more fully addresses quality assurance related to the clinical software (sometimes referred to as applications software) used to process the acquired images. Such software provides objective parameters of specific organ functions and generates functional diagnostic patterns, for example, for cardiac studies and for renal studies. The clinical software may be supplied as an integral part of the purchased scintillation camera–computer system, or it may be a part of an independent nuclear medicine computer system.

The user expects that the clinical software has been developed to provide valid quantitative data for a well-defined objective and that this software has been well tested. The responsibility for this lies with the manufacturer or producer of the software. The software must have been carefully tested in clinical settings in a large number of studies in order to ascertain its validity for the objective. Some clinical studies must be supplied with the released clinical software as sample studies with known results for the user to test the software in their own environment. The software must also be accompanied with full documentation.

The clinical software requires acceptance testing and, thereafter, regular monitoring. The first responsibility of the user, therefore, is to perform quality assurance assessment on the clinical software before actually putting it into clinical use. This involves becoming fully familiar with the documentation and with all aspects of the clinical procedure. The objectives and assumptions of the software must be well understood. The software itself must then be tested with a set of sample clinical studies, or a database of studies with known results, in order to ensure satisfactory results. In order to maintain quality assurance once the software is accepted and put into routine clinical use, the user is responsible for monitoring the software results. This requires user training, audits of results using the software, and inter- and intra-observer comparison tests.

The user must be aware at all times that the potential of a system to produce errors is almost unlimited. It is vital never to accept blindly the results from clinical software, but to remain open at all times to inconsistencies and errors, and to take care that results are reasonable. Owing to the diversity of clinical software, it is not possible to give exact protocols for quality assurance. The following guidelines are intended to help limit errors, to help check that the system is well behaved and to help the user gain confidence in the generated results.

6.2.2. Library of clinical studies

When the purpose of a nuclear medicine procedure is to provide quantified physiological parameters, different software should produce the same and reproducible results within a defined margin of error. The same is true of the same software used by different users. A particular software package that offers fully automatic processing (without operator intervention) must produce identical results for the same clinical study. In order to test the clinical software, a library or database of clinical studies is required, one that will allow the studies to be processed with different clinical software or versions of software, their results compared and a reference set of results obtained for each study. This bypasses the data acquisition part of the clinical procedure and

concentrates on the processing. Whenever new clinical software is purchased or when software is modified, the 'new' software can be tested and compared with these reference data. The studies should include, if possible, clinically validated normal and abnormal studies, which help establish normal and abnormal ranges.

An example is the gated blood pool study analysed with clinical software that determines the left ventricular ejection fraction. The user establishes the ejection fraction value as determined by the clinical software installed on their system and compares that value to the value expected by the same version of software or by different software. Reproducibility is an important check. It should be noted that the ejection fraction might vary up to ±50% between different clinical software packages. Modification of a software package may produce a change in ejection fraction values. The user must always be aware of possible differences or changes in values and must always establish and/or confirm the normal range of values to be adopted. Several levels of clinical studies can be considered for different purposes.

6.2.2.1. Sample clinical data from the manufacturer or supplier of software

The manufacturer or software supplier must provide sample clinical data with reference results for each clinical software package. These studies are required to check that the clinical software has been installed properly and that it is being used properly. The sample of clinical data must include at least one normal and one abnormal study. When an updated version of software is supplied, a new set of reference results for the same sample of clinical data must be provided.

6.2.2.2. Reference clinical data

It is recommended that a library of studies with reference data be established for each clinical procedure in regular use. These serve as an extensive set of reference studies that are available for the (re-)evaluation of the software. The studies may be clinical studies provided by other departments, or they may be the results of clinical studies collected from one's own department. Normal and abnormal studies must be included.

A set of studies could be obtained through a computer users' group or through others using the same clinical software. Pooling studies may help with the set-up of normal and abnormal reference results, when the studies are clinically validated according to strict criteria characterizing normal and pathological conditions. However, pooling of data also requires strict adherence to standard procedural methods.

6.2.2.3. International databases

Validated clinical databases of studies, such as those promoted under the COST B2 program, are valuable for comparing software results against known data. The advantage of such databases is that they have been tested on a wide basis in a number of departments and with different software packages, which may include the one the user is using or is about to use.

Such studies will most likely be in a standard file format, such as Interfile or DICOM. Before using such studies, the user must be able to 'read' this format and be able to convert it into the native file format of their own computer. The user is cautioned though that conversion may not be effortless and a study file header may require some editing.

6.2.2.4. Database results that define normal ranges

Some clinical software packages include a database of normal results that have been developed at one or more institutes. Care must be exercised in using these databases.

It is necessary to be certain that the results from the clinical software conform to those expected from the software. This can be checked by processing a sample set of 'normal' studies used for the database. This can only partially test specificity but gives no indication of sensitivity. A sample set of such data should be supplied with the clinical software.

It is necessary to know details of the patient population that contributed to the normal database and to determine whether the patients examined in one's own department are matched to the database population (e.g. in age, size, weight). Details of the data population must be supplied.

If the decision is taken to use the database of normal results, care is required to use the same clinical procedure (e.g. for SPECT myocardial perfusion studies, a database supplied for supine patient positioning cannot be used with prone patient positioning).

6.2.2.5. Paediatric databases

Quantifying image data acquired from children may be difficult and may even fail, for example, owing to small ROIs, small organ size that affects automatic ROI edge detection and patient movement. The user is cautioned in using clinical software designed for adults for processing paediatric studies. If a large percentage of patients are paediatric, then special efforts must be made to obtain software that has also been tested on a paediatric population. This area needs more attention.

6.2.2.6. Mathematical simulations

Data from mathematical simulations are data of known form and data on statistical noise distribution generated by a computer. Such data are useful in verifying specific algorithms because the exact result is known. For example, a simulation based on the physiological model of the left ventricle can be helpful for testing specific algorithms of the cardiac software that determines the ejection fraction. Simulated data can be used in conjunction with, but not to replace, actual in vivo clinical data.

6.2.3. Test of clinical software

Purpose of test

The purpose of this test is to verify that the results from the implementation of the clinical software are as expected and are reproducible. The guidelines in Section 6.2.4 provide more details regarding specific situations that require a test of the clinical software.

Materials

The materials required are a set of clinical patient studies suitable to be used with particular clinical software. For example, a set of 10 studies (or more) that covers the range of normal and abnormal results, which could be supplied by the software provider and which originate from an international database or from one's own department.

Procedure

(1) Ensure that the 10 test studies are available for data processing in one session and that the persons who will perform the test are familiar with the processing procedure and documentation accompanying the software.
(2) Process the studies by employing each person who will use the software, preferably in one session.
(3) Document the results and include a hard copy of the final display of results, the date and the person who processed the data.
(4) Twice repeat steps (2) and (3) on other, separate occasions, so that each person has processed the test studies three times.

(5) Perform Bland–Altman [22] plots of the quantitative results for each pair of results, in order to evaluate the intra- and inter-observer reproducibility. A Bland–Altman plot is made by plotting the difference between the means on the Y axis against the average of the means on the X axis. It is a sensitive test to determine the validity of one test compared with a second test or a new test.

Interpretation of results

(1) Examine the documentation display and hard copy of the study and check that it conforms with that expected (e.g. study identification, numerical data, labelling of images, absolute or relative colour scaling, colour scale used, ROIs).
(2) Examine the Bland–Altman plots of the quantitative results and check for acceptable variability over the whole range of abnormal and normal values. Ensure that there is no difference in variability at the extreme ends of the range of results, or a systematic offset.

Limits of acceptability

(1) For fully automatic clinical software, the results from the same clinical studies should be identical, regardless of who processed the data and regardless of the computer system on which the processing took place.
(2) Intra-observer variability, i.e. the same person processing the same set of clinical studies on different occasions, should produce <3% variability in results.
(3) Inter-observer variability, i.e. different persons processing the same set of clinical studies, should produce <5% variability in results.
(4) If the limits are exceeded, the reasons should be investigated, follow-up action taken and the test repeated.
(5) If the results and display do not conform to that indicated by the software supplier, then this must be reported to the supplier for follow-up action. The software should not be used clinically before the problem has been resolved.

Conclusion

(1) This test should be performed by each person who will use the software for the first time, before that person is permitted to use the software routinely.

(2) This test should be performed by each person who infrequently uses the software before using the software.
(3) Record the date and document follow-up action taken.

6.2.4. Practical quality assurance guidelines

Some practical guidelines are given below for various situations where the user is confronted with a new or revised clinical procedure using new or updated clinical software. The general test given in Section 6.2.3 is applicable for testing software and for assessing its acceptability.

6.2.4.1. Learn and understand the documentation

When clinical software is to be used for a particular clinical procedure, the user must first understand the objectives of using the software and the exact procedure requirements in order to use that software correctly. This includes learning about the requirements for the radiopharmaceutical, the patient preparation and study set-up, the data acquisition method, the requirements for data analysis and display, the algorithms and assumptions of the software, and for the use of any normal database results. The documentation supplied with the software should supply this information.

This also applies when an updated version of existing software is to be used. The documentation should be studied carefully to learn what changes have been made to the existing software.

6.2.4.2. Initial tests of software with clinical sample data

Before using the new software or updated software routinely, the software should be tested using sample clinical studies supplied by the manufacturer. These sample data must always be available and must include at least one normal and one abnormal example. These studies must be supplied with the appropriate values and images. It is helpful to have a knowledgeable representative of the manufacturer present during these tests, especially when using the software for the first time. Any discrepancy between one's own results and the documentation should be noted and reported.

If a database of validated clinical studies with reference results exists, the clinical software should be tested using this database and the results examined; if differences are noted, reasons should be sought (see Section 6.2.4.5).

6.2.4.3. New data acquisition protocol

New clinical software may introduce a new acquisition protocol that replaces an existing one. It is important to learn how the new software works and how its requirements differ from the old protocol. For static, gated SPECT and whole body studies (all of which can be repeated without a second injection of the radiopharmaceutical), a study should be acquired and the data processed using both new and old protocols and the differences determined. This will not be possible with dynamic studies.

6.2.4.4. Use of a new or different computer system or software

All users are confronted at some time with the task of replacing an old computer system with a new one, or with implementing new applications software. For all routine studies, it is essential that differences in procedures and software results be evaluated between the new and old software in order to conserve continuity in results. Different approaches to acquiring test data could be considered, depending on the situation:

(1) Acquire data simultaneously on both old and new computers and evaluate and compare results.
(2) Use a set of studies from the old system and process with the new system, then compare the results. One problem could be 'reading' the studies with the new system and differences in acquisition protocols.
(3) Apply the same database of clinical studies on both systems to check the processed results.

Results to be expected from upgrading the computer system with:

- *No change in the clinical software.* No changes should be expected. Identical results should be obtained for the same input data when no user interaction is required and a maximum of 5% variation can be expected when user interaction is required.
- *New software from the same manufacturer.* A maximum variation of 5% can be expected. The user must be particularly aware of a systematic offset in the results.
- *New software from a new manufacturer.* A database of studies must be tested. For software with the same function (e.g. ejection fraction determination), a comparison between the old and new situations is required. The user must be very careful to re-establish a normal range of values (see Section 6.2.4.5).

6.2.4.5. Normal ranges

Normal values and ranges should be established, preferably with a suitable database of clinically validated studies, if available. This is one of the most difficult aspects of the quality assurance procedure. If such a database is not available or suitable, then it is necessary to build one that has a 'normal' population. Any study used to determine a normal value must be carefully archived for future use as reference data and for re-evaluation with new or modified software. The clinical criteria originally used to categorize the study as normal must also be archived for reference purposes.

Once a range of normal values has been established, or a new normal range has been established, transfer this knowledge to the clinical users of the system. Confirm this range periodically by reprocessing the same normal set of studies (e.g. for training and continuing education). This serves as a quality control check to ensure that a change in processing method has not crept in over time.

6.2.4.6. Training and responsibilities

A person should be designated to receive special training in the use of the clinical software. This person, the 'specialist', should then be responsible for instructing others in the use of the software. This may include initially testing the software. This person should assist in drawing up a local guideline for using the new software and can help with monitoring the use of the software and the software results.

New personnel must be properly instructed and trained in the use of the software. A record of the persons trained to use particular software should be maintained and only those persons trained should be allowed to use that software in routine clinical practice. It is possible to conceive of a password system whereby access to the clinical software for routine purposes can only be gained by trained persons.

6.2.4.7. Record logbook

A logbook should be maintained which records all software testing activities, problems and solutions, changes in procedures and actions taken. New software or software updates, including the software name, version and the date implemented, should be recorded. The logbook may serve as a basis for audits.

6.2.4.8. Audits

Audits should be undertaken on a regular basis. Such audits should consider the technical problems associated with software as well as clinical software results. It is essential that the results of audits be communicated to all those involved.

Clinical audits

The purpose of a clinical audit is to certify that results from software are as expected and are consistent. Examples of clinical audits are:

(1) *Retrospective review of results.* A retrospective review of software results with diagnosis: Did the results correspond with the diagnosis? If not, then action is required. Abnormal sensitivities and specificities are of interest since they may indicate improper use of the system or faulty components. This is also particularly useful when a systematic change has been made in the procedure method (e.g. acquisition or processing parameters, colour tables).
(2) *Inter- and intra-observer comparison.* Reprocessing a set of reference studies within the department to determine inter- and intra-observer variations in the results and to determine the reproducibility with respect to previous results from the same reference studies.

The data set used at the installation of the software could serve as a set of reference studies, or as another reference data set with known results. A useful method of comparing results from two analyses of the same data is the Bland–Altman method [22], whereby the mean value of two results is plotted against the difference in these two results. Such a plot gives an excellent insight into variations between two measurements and any offsets between two measurements over a wide range of values (e.g. for low, medium and high ejection fractions). A linear regression comparison is not sensitive and is inappropriate for these comparisons.

Technical audits

The purpose of technical audits is to review problems, find solutions and achieve improvements. They are useful for following up long term, outstanding problems, so that they are not forgotten. The logbook should contain all necessary information and it is, therefore, an invaluable source of information. Feedback and follow-up of audits is essential. The results of audits must be

regularly discussed with all involved, e.g. at routine departmental meetings. The main points should be documented in the minutes of the meetings.

Examples of technical audits are:

(1) *Problems and solutions.* A regular review and documentation of all problems and their solutions, including the frequencies of problems, helps to monitor software and operator performance.

(2) *Software failures/errors.* When a software failure or error occurs, it is necessary to establish under what conditions it occurred and whether it could recur. A review of the failure of the software to generate results in the way expected must be made regularly and communicated to all involved and to the software supplier in order to obtain an improvement and solution. An example of software failure is when automatic edge detection fails. A review requires understanding the circumstances under which the software fails and assessing whether this is due to improper use or to a software error. Examples of improper use are not using the correct pixel size (zoom factor), incorrect positioning of ROIs, using the software for a purpose other than that intended (e.g. software for left ventricular ejection fraction program used to process the right ventricular ejection fraction). It should be noted that problems encountered may not necessarily relate to an error with the software itself, but to another part of the procedure, for example:

- The technical performance of the study (e.g. timing/time sequence of the study, labelling of the radiopharmaceutical, injection technique, collimator selected, patient positioning);
- The data acquisition (e.g. matrix size, pixel size, images/time per frame for a dynamic study, count statistics);
- The data analysis and display (e.g. ROI selection, background ROI positioning, colour scale).

(3) *Software modifications and updates.* A mechanism to review the effects of software modifications and updates with all persons involved should be established, not only by discussion but also by actual demonstration and training in the use of the new software.

6.2.4.9. Each clinical procedure

(1) Follow the standard procedure, especially for patient positioning, data acquisition, data analysis and display.

(2) Note any divergent methods with the patient study documentation. Non-standard studies must not be included in establishing one's own library of reference studies.
(3) Record ROIs selected with the study results (e.g. make a hard copy).
(4) If quantitative data do not agree with visual images, then this must be investigated.
(5) For quantification that relies on a standard amount of radioactivity measured in a radionuclide activity calibrator, make sure that the calibrator quality control is performed and is acceptable.

6.2.4.10. Procedure manual

It is important to develop and to maintain a departmental clinical procedure manual that describes the full details of the clinical procedure and the use of the clinical software. This manual should be updated at least annually and whenever modifications have been made. This manual should be an integral part of the quality handbooks of the department and should always be available for reference. All users must be familiar with its contents.

6.2.4.11. Change in clinical procedures

It may be necessary to make a systematic change in a procedure with respect to acquisition parameters (e.g. matrix size, pixel size) or processing parameters (e.g. change limits between which parameters are calculated, change filter type and/or filter parameters, introduce a new parameter or new fitting algorithm, use another colour table and/or colour characteristics (logarithmic versus linear scale)).

If this is the case, then:

(1) Assess the effects on quantitative results and on the images, using reference studies.
(2) Edit the departmental clinical procedure manual.
(3) Transfer this knowledge to other users, including nuclear medicine physicians.
(4) Introduce the modification into routine practice.
(5) Note the modification in the logbook together with the date implemented.
(6) Perform a clinical audit to investigate the effect of changes made (see Clinical audits in Section 6.2.4.8).

It should be noted that a modification to a procedure may have profound implications in the interpretation of studies and should never be made on an ad hoc basis.

6.2.4.12. Continuing education

A plan for the review of methods and any changes should be conducted on a periodic basis. Clinical software data analysis should involve all those who participated, including the nuclear medicine physicians. A record should be kept of which software application was reviewed, who took part and when the continuing education took place. A regular plan for continuing education in the use of clinical software should be established.

6.2.4.13. Communications with other users

It may be useful to form or join a local users' group of the computer system being used. This is often an excellent forum for discussion of common problems (and solutions), with feedback to other users of the same system, as well as to the manufacturer, software supplier, or local vendor.

Within the group, sample studies could be circulated for processing a specific clinical software package and a comparison of results made. Knowledge of the variation and reproducibility of results is essential for all users of the same software, but might also be useful on a wider scale for presentation at a regional nuclear medicine meeting.

7. DISPLAY: HARD AND SOFT COPY

7.1. INTRODUCTION

The display system includes the computer monitor (soft copy) and the devices and components responsible for producing the paper or film output (hard copy). The discussion here is limited to nuclear medicine systems that have a digital computer associated with them. Purely analogue camera systems are not discussed in this publication.

There are numerous devices used to produce soft copies and hard copies, and the number and type of devices will continue to increase as technological innovation continues. Not all such devices are suitable for use in imaging. It is

important that the users realize the limitations of the various devices. The tests described in this section will be helpful in characterizing displays with respect to their suitability as nuclear medicine display systems. The tests described here will be especially useful for monitoring the performance of these devices at the time of acceptance testing and as operational tests to ensure the consistency of their performance.

7.1.1. Basic principles

The variety of display devices all share common parameters, which include:

- Brightness;
- Contrast;
- Grey scale mapping as a function of counts;
- Spatial resolution;
- Spatial linearity (barrel and pincushion distortion);
- Focus;
- Stability.

Testing of all of these items, if done with compulsion and care, could be tedious and time consuming if detailed quantitative answers are needed. However, what is desired are performance tests that demonstrate that the display systems are working correctly.

Early in the development of computer displays this problem was recognized by persons working with computer displays and medical images. It was formally addressed by the Society of Motion Picture and Television Engineers (SMPTE) in 1983. At that time, the SMPTE defined a test pattern that permitted qualitative and quantitative testing of diagnostic displays. This definition has been updated several times in the intervening years. The latest revision is known as SMPTE Recommended Practice RP 133-1991: Specifications for Medical Diagnostic Imaging Test Pattern for Television Monitors and Hard-Copy Recording Cameras [23]. The test pattern is shown in Fig. 51.

To quote from the SCOPE section of the SMPTE Recommended Practice [23]: “This practice describes the format, dimensions, and contrast required to make diagnostically significant measurements of the display and camera system resolution for both digital and analogue monochrome signal sources. The practice provides users of medical diagnostic systems with a comprehensive test pattern for day-to-day operational checks and adjustments

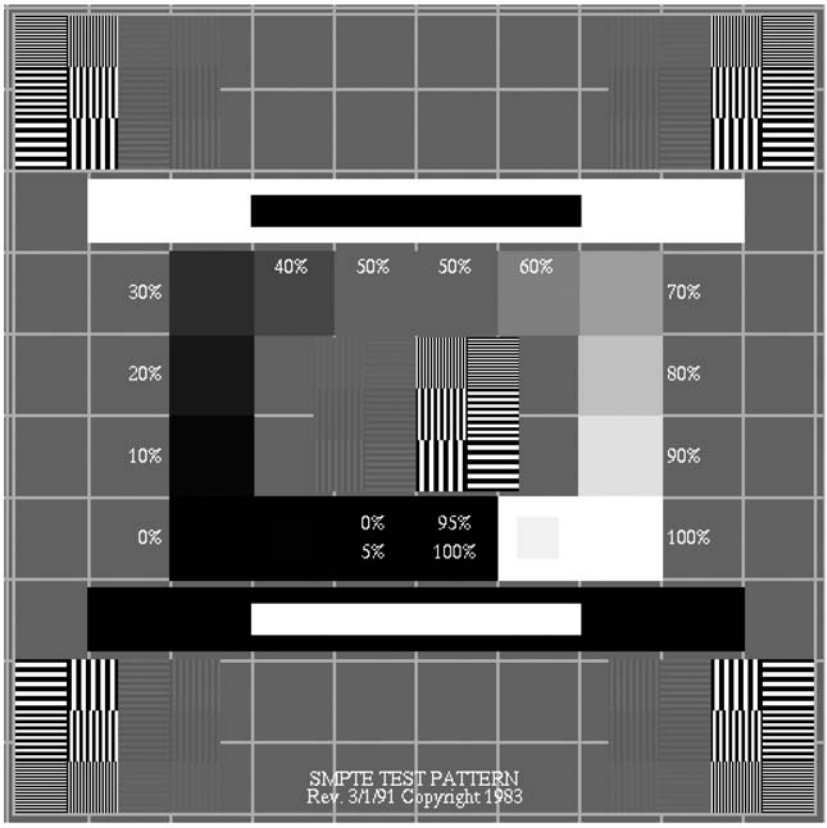

FIG. 51. The SMPTE test pattern showing the basic elements used in display quality control testing.

of focus, brightness, and contrast, resolution response, mid-band streaking, uniformity, and linearity of viewing monitors and hard-copy recordings."

It should be noted that the test pattern was initially designed for certification of monochrome systems. However, it can be utilized equally well for colour displays and monitors in the red/green/blue or enhanced mode of operation.

In the following section, some suggestions are discussed as to how to use the digital test pattern as an aid to acceptance testing and periodic quality assurance. The SMPTE publication suggests other, more quantitative and thorough uses of the test pattern to interested persons.

7.2. ACCEPTANCE AND REFERENCE TESTS

Electronic versions of the SMPTE test pattern are available free from several sites on the internet. This should allow most nuclear medicine users to import a digital version into their nuclear medicine system and display it on the system's monitors and print it on film. All of the tests described below assume that the nuclear medicine imaging system being tested has imported the digital SMPTE test pattern so that it can be displayed and printed.

7.2.1. Test of brightness and contrast

Brightness and contrast of a monitor or minimum and maximum optical density of film can be easily verified in a qualitative but important use of the test pattern.

Purpose of test

To test the brightness and contrast of a display monitor and the optical density of film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

Display the test pattern on the monitor, or make a film or print of it.

Data analysis

The section of the test pattern reproduced in Fig. 52 should be noted.

When properly displayed on a general purpose display, both the 'square within the square' items should be visible, indicating that the full range of possible grey scales are viewable. This is usually achieved by setting the brightness and contrast controls of the monitor (for soft copy use) or the exposure controls of the camera (for hard copy film uses). Several trial adjustments will be required to allow the full range of display contrast to be displayed.

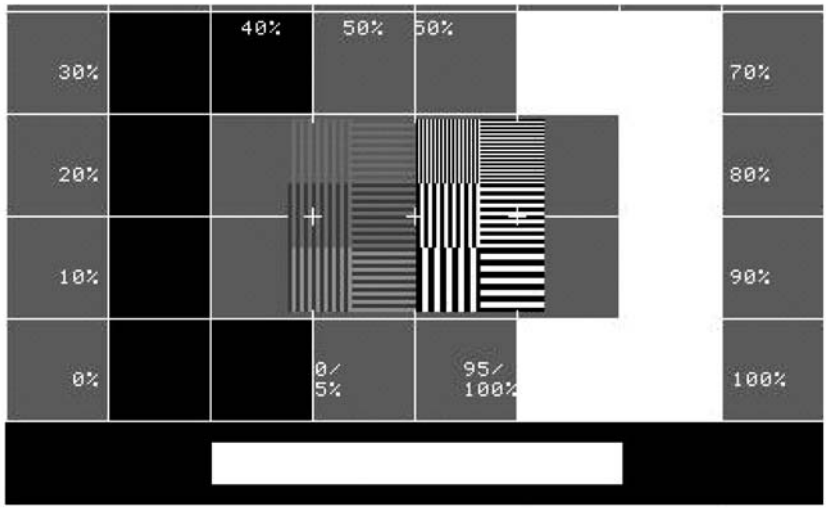

FIG. 52. A portion of the test pattern showing the section used in the brightness and contrast control settings. At the low end of the scale there is a 0% and 5% intensity square within the square which should be visible (the square to the left of the 0/5% square). At the high end of the intensity scale, the square within the square includes the 100% and 95% intensity levels (the square to the right of the 95/100% square). These should also be visible during routine operation of the display. For a good quality reproduction of this pattern see Ref. [3].

7.2.2. Grey scale linearity

Purpose of test

To test the grey scale linearity of a display monitor, film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

Display the test pattern on the monitor or make a film or print of it.

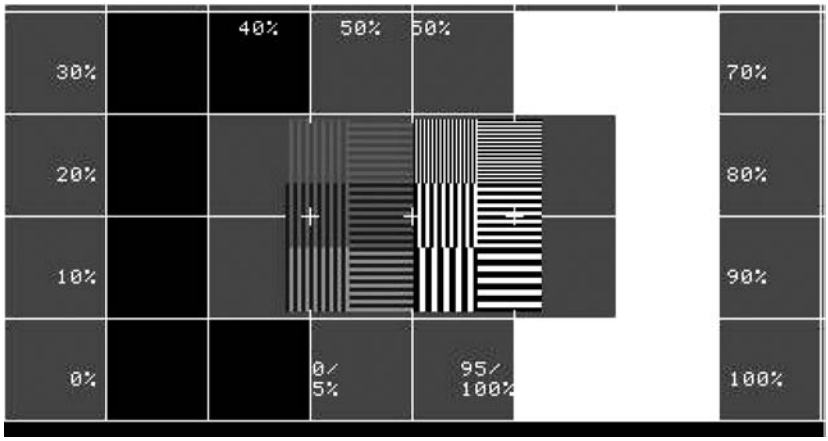

FIG. 53. The central portion of the test pattern showing the total range of contrast from 0% to 100% in 10% increments. Each section should be uniform and of a distinctly different shade than its adjacent section.

Data analysis

Figure 53 is taken from the central portion of the SMPTE pattern.

The 10 steps moving from black to white are related to the counts in a linear fashion and should all be perceived as different grey levels.

7.2.3. Spatial linearity

Purpose of test

To test the spatial linearity of a display monitor, film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

Display the test pattern on the monitor or make a film or print of it.

Data analysis

The 1010 major grid (see Fig. 52) will aid in the determination of display linearity. It should appear to be straight and uniformly spaced on the display. Quantitative measurements can be made if desired. All 100 squares are intended to be the same size. Some non-linearity is to be expected. Differences of more than 10% should be corrected.

7.2.4. High contrast spatial resolution

Purpose of test

To test the high contrast spatial resolution of a display monitor, film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

Display the test pattern on the monitor or make a film or print of it.

Data analysis

Figure 54 shows the area used in observing the high contrast resolution properties of the display system.

7.2.5. Low contrast spatial resolution

Purpose of test

To test the low contrast spatial resolution of a display monitor, film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

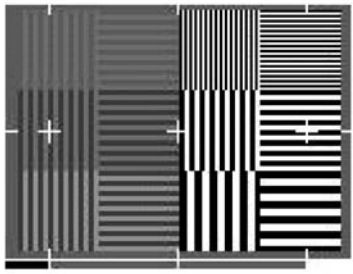

FIG. 54. A section of the pattern showing the high contrast resolution area. There are five such areas in the full SMPTE test pattern. Modulation is 100%. Appearance should be similar at all five locations.

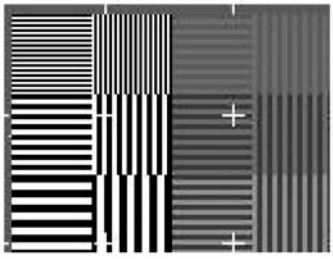

FIG. 55. The low contrast resolution section of the test pattern. There are five such sections located in the test pattern. There are three sections with 1%, 3% and 5% contrast. These patterns are sensitive to various types of imaging noise.

Procedure

Display the test pattern on the monitor or make a film or print of it.

Data analysis

Figure 55 shows the patterns of the different sections. These patterns are sensitive to imaging noise.

7.2.6. Effect of large contrast changes

Purpose of test

To test the effect of large contrast changes on a display monitor or on the optical density of film or on the appearance of a print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

(1) Display the test pattern on the monitor or make a film or print of it.
(2) Invert the black and white colour table and display the test pattern on the monitor or make a second film or print.

Data analysis

Figure 56 shows the white on black and black on white performance of the display system. There should be no streaking, smearing or banding in a properly performing display system.

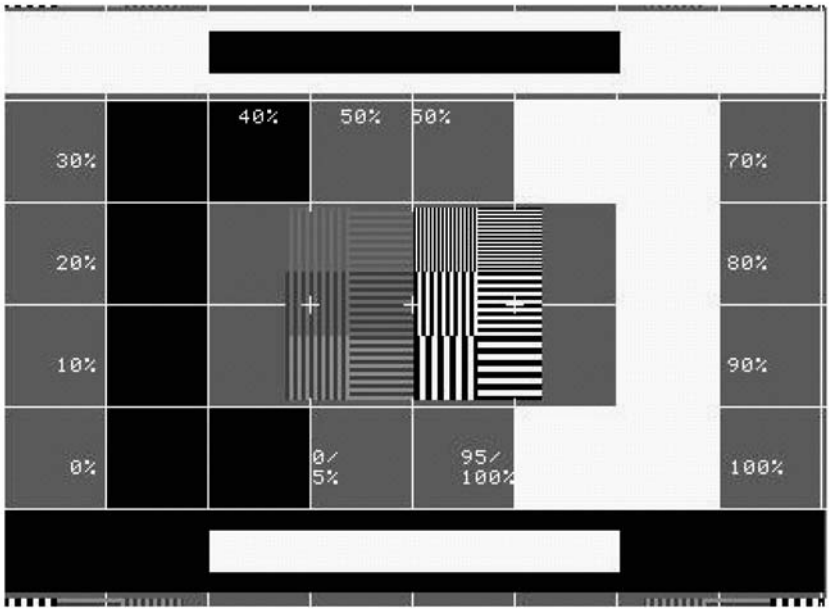

FIG. 56. Black on white section (top horizontal bar) of the test pattern used to check the performance when changing full white to full black. White on black section (bottom horizontal bar) of the test pattern used to check the performance when changing full black to full white.

7.3. OPERATIONAL CHECKS

7.3.1. Check of the display

Purpose of test

To test the general performance of a display monitor, film or print.

Materials

- A display monitor, multiformat film output or printer.
- The SMPTE test pattern displayed on the monitor, film or print.

Procedure

Display the test pattern on the monitor or make a film or print of it.

Data analysis

A comparison should be made with the reference prints made at the last full display check. Changes based on the checklist below should be noted and corrections made if any changes are detected. The following checklist for operational test of the display should be followed:

- Full range of greys displayed.
- Both 5% and 95% squares within the squares are visible.
- The 1010 grid is uniform and consistent across the display.
- The high contrast resolution pattern is unchanged.
- The low contrast resolution pattern is unchanged.
- The image is free from streaks and smears.

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GLOSSARY OF TECHNICAL TERMS

attenuation. When a beam of photons passes through absorbent material, the likelihood that the photons will experience interactions depends on their energy as well as on the thickness and properties of the absorbing material. These interactions reduce or attenuate the beam intensity.

attenuation correction. In SPECT, attenuation reduces the apparent concentration of radiotracer in deep tissues in tomographically reconstructed images. This can lead to errors in interpretation as well as quantification. Attenuation correction compensates for this effect.

backprojection. This is the process used in reconstruction which allocates counts in the reconstructed image at each voxel in proportion to the recorded counts on the projection, defined by the geometry of detection. In the simplest case, assuming a parallel hole collimator, each voxel will be allocated counts from a projection pixel, defined by a line drawn at right angles to the projection that passes through the voxel.

bar phantom. A phantom (test pattern) consisting of lead bars of varying width and separation, set in Perspex or other plastic material. The bars can be arranged either parallel to each other across the entire phantom, or, more usually, into quadrants of parallel bars. Bar separation and width is then different for each quadrant. Bar phantoms are primarily used for measuring spatial resolution.

centre of rotation. This defines the point that should correspond to the exact centre around which the detectors rotate. This point should correspond exactly to the centre of each image in the X direction, recorded at each angle. If the centre of the projection matrix does not correspond to the physical centre of rotation of the collimated detector, a loss of spatial resolution in the reconstructed images will occur. This error is termed the centre of rotation offset.

collimator hole angulation. A parallel hole collimator should be constructed so that the holes and septa are exactly perpendicular to the surface of the crystal. Any difference in this angle is referred to as a collimator angulation error. Any such angulation error will lead to an error in defining the positional origin of a detected photon at depth.

converging collimator. Similar to the parallel hole collimator, except that the holes are angled to converge to a focal point at some distance in front of the collimator. This collimator can be used to obtain a magnified image of a small organ.

cut-off (*or* critical) frequency. The shape of a filter is defined by some mathematical function, with the value 1 at zero frequency and lower values at progressively higher frequencies. The cut-off or critical frequency is a parameter that defines the shape of the function, a lower cut-off frequency defining a curve that drops to zero faster, resulting in a smoother image. In the case of the Butterworth filter, the cut-off frequency defines the point at which the amplitude reaches half the maximum value.

dead time. In radiation counting devices, the time required to resolve individual detected events is termed dead time. Pulses produced by radiation detectors all have a finite time duration such that if a subsequent pulse arrives before the first has been processed, only the first will be counted.

default PHA window. This is the energy window width recommended by the manufacturer.

Digital Imaging and Communication in Medicine (DICOM). This is a group of standards for transmitting and storing information in medical imaging. The standards include definition of file formats and a protocol for network communications.

energy spectrum. A plot of the number of photons detected as a function of their energy.

energy (*or* PHA) window. The energy window or PHA window sets the limit of photon energies to be accepted. All photons with energies falling outside this window are rejected.

field of view. This is the physical area as seen by the collimator.

filtered backprojection. A computationally efficient algorithm for reconstructing tomographic slices from projection profiles. Filtered backprojection is the most commonly used method for SPECT image reconstruction.

full width at half maximum (FWHM). Refers to resolution measurement (e.g. spatial, energy resolution). Spatial resolution 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 of a single gamma emitting radionuclide. The spread is due to various degrading effects and is measured by the full width of the profile at a point which is half the maximum height of the profile.

hydration. The NaI(Tl) scintillation crystal is hygroscopic and requires a hermetic seal to prevent the absorption of water. If this seal is broken, exposure of the crystal to a humid atmosphere can result in a surface discolouration (yellow) that impairs the transmission of light to the PMT array.

iterative reconstruction. This general term applies to a number of reconstruction algorithms that involve a repetitive process of comparison to find the best estimate of the activity distribution that matches the measured projections. Common examples of iterative reconstruction include maximum likelihood expectation maximization and ordered subsets expectation maximization.

line source. A thin line (such as a capillary tube) filled with radioactivity, used to measure spatial resolution. The diameter of the line source should typically be less than or equal to 1 mm.

multiple head SPECT system. This is a SPECT system with more than one detector head. Each detector head acquires a section of the total rotation. The data acquired from all the detector heads are used to reconstruct the tomographic images.

photomultiplier tube (PMT). Electronic tubes that produce an electrical pulse in response to being stimulated by weak light signals, such as the scintillations produced by a gamma ray interacting with a NaI(Tl) crystal.

pulse height analyser (PHA). Analyses the size of the energy signal and produces an output only if the size of the energy signal is within the range specified by the predefined energy windows.

reconstruction. This is the process of obtaining tomographic images from a set of projection images.

ring artefact. This is an artefact in the transverse reconstructed images caused by a localized non-uniformity in the detector or collimator, or a non-linearity in the computer interface.

scatter correction. Scattered photons that are detected by the scintillation camera have slightly less energy than the primary photon (gamma ray) and can carry erroneous positional information. Detector systems do not have sufficient energy resolution to eliminate all of these scattered photons. This scatter creates a background signal that blurs fine detail in projection images and contributes to quantitative inaccuracy. Scatter corrections are methods of reducing these effects. Although there are many scatter correction methods, most use algorithms to estimate the distribution of scattered gamma rays in projection data and then subtract this estimated distribution.

scattered photon. A photon that has undergone one or more interactions, by which it has lost energy and changed direction from the original path.

sensitivity. The fraction of the emitted photons detected by the collimated scintillation camera (system sensitivity). It is specified for each collimator in units of counts/second/MBq.

single photon emission computed tomography (SPECT). Cross-sectional slices of the radionuclide distribution in the patient are generated by taking images all around the patient (projections) and then reconstructing these images with a computer into cross-sectional slices.

sinogram. The image formed by placing projection values in sequential rows (i.e. arranging pixels corresponding to projection position versus projection angle) is termed a sinogram, since the projections from a single point describe a perfect sine wave when plotted in this form.

smoothing. An operation that involves spreading values across neighbouring pixels; the averaging effect reduces statistical noise but degrades image resolution. Smoothing is a filtering operation often achieved by convolution.

sodium iodide (NaI) crystal. The most commonly used detector in nuclear medicine is the sodium iodide crystal to which has been added a small amount of thallium, NaI(Tl), which allows the crystal to produce light in proportion to the energy deposited when a photon interacts in the crystal.

spatial resolution. This is the system's capability to distinguish between two, small, closely spaced radioactive sources. It is usually expressed in terms of FWHM.

uniformity. A measure of how uniform the observed counts across the field of view are when the detector is irradiated with a uniform distribution of radiation.

voxel. If a digitized 3-D volume rather than a digitized 2-D image is considered, each digital value within the volume can be considered to occupy a small volume (e.g. a small cube), element, or voxel. It is possible, therefore, to refer to planar projections as having pixels, but each reconstructed slice as having voxels, which also have a thickness corresponding to the spacing between adjacent slices.

ABBREVIATIONS

CFOV	central field of view
COG	centre of gravity
COST B2	European Cooperation in the Field of Scientific and Technical Research
CPU	central processing unit
DICOM	Digital Imaging and Communications in Medicine
ECG	electrocardiogram/graph
FWHM	full width half maximum
FWTM	full width tenth maximum
ICR	input count rate
IEC	International Electrotechnical Commission
ISO	International Organization for Standardization
NaI(Tl)	sodium iodide activated with thalium
NEMA	National Electrical Manufacturers Association
OCR	observed count rate
OHTP	orthogonal hole test pattern
PHA	pulse height analyser
PLES	parallel line equal spacing
PMT	photomultiplier tube
PSF	point spread function
$R_{-20\%}$	input count rate for 20% count loss
ROI	region of interest

SMPTE	Society of Motion Picture and Television Engineers
SPECT	single photon emission computed tomography
TTL	transistor–transistor logic
UFOV	useful field of view

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The objective of this publication is to provide professionals in nuclear medicine centres with quality assurance procedures for the scintillation camera, computer system and digital image display. It is intended to be a resource for medical physicists, technologists and other healthcare professionals who are responsible for ensuring optimal performance of imaging instruments, particularly SPECT systems, in their respective institutions. It may also be useful to managers, clinicians and other decision makers who are responsible for implementing quality assurance/quality control programmes in nuclear medicine centres.

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