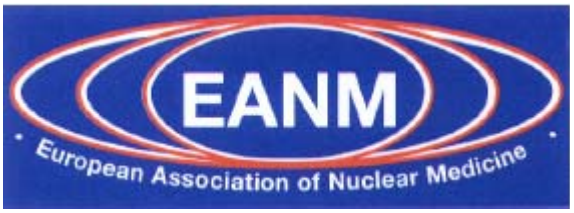

Draft Work

Guidelines on Quality Control for Nuclear Medicine Instrumentation

EANM Working Group Authors: Dr M Lyra[1], Dr R Klett[2], Dr W B Tindale[3]

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December 2007

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I Introduction

The purpose of this document is to provide nuclear medicine practitioners (physicists, technologists, physicians) with general guidelines in performing and interpreting quality control (QC) of all instruments of a Nuclear Medicine Department (counting devices, imaging systems) with a special focus on new, sophisticated, digital technologies. The aim is to help in achieving a high quality standard of diagnostic imaging, which will maximise the impact of nuclear techniques in diagnosis.

These guidelines recommend the necessary actions to ensure that the Nuclear Medicine instrumentation is kept in proper condition and that appropriate results are provided for patient care. Thus they will assist practitioners in obtaining accurate information with the smallest radiation burden and minimum clinical risk to the patient. As such, they are general recommendations on good practice which should be applicable across European boundaries.

As these QC guidelines cover both non-imaging and imaging equipment, they provide a general reference for routine quality control procedures required to ensure that Nuclear Medicine equipment is functioning correctly. With good quality control procedures, problems associated with the systems will be detected before they impact on clinical patient studies.

This document is divided into sections covering QC of the following instrumentation: non-imaging systems, planar gamma cameras / whole body scanners, SPECT gamma cameras, gamma camera PET (GCPET) systems, PET scanners and hybrid PET/CT systems. It also incorporates a section on attenuation correction for all tomographic scanners and has a short contextual section on image quality control.

Rationale, methodology, performance acceptance and an indication of the frequency of the recommended quality control procedure are included. As the frequency of tests depends on the equipment, criteria are provided for selecting appropriate test frequencies. In certain countries specific legislation exists which deals with quality assurance. If this demands tighter limits on acceptable performance and test frequency, these country-specific regulations must be observed instead of the values given in this guideline. When considering the frequency of the test, one must take into account the manufacturers' recommendations and also past experience as to the reliability of any given

system. Additional testing may be necessary if there is any concern about the function of the system or if there has been any maintenance undertaken or adjustment made which may impact on the performance.

Advances in gamma camera design and in the embodied CT systems, over the last 5 years, have improved all aspects of image quality, particularly for tomographic imaging. As system complexity increases, it becomes more important that all Nuclear Medicine workers are able to recognize the various types of aberrations or artifacts that can occur in Nuclear Medicine imaging systems. An understanding of the potential impact of system performance aberrations is a prerequisite for implementing an effective quality control program. These guidelines attempt to provide background information for performing particular procedures.

A quality control programme consisting of calibration and performance tests must be designed in each Nuclear Medicine Department; it should include regular testing procedures to ensure proper operation on a daily basis and attention should be given to the daily QC results that may indicate system deterioration. The programme should be reviewed when new techniques are introduced.

It is imperative that quality control procedures are carried out in a consistent manner (for example, background, collimation, orientation, radioisotope, activity, energy window width, attenuation correction technique etc) and the quality control settings and results are recorded to enable meaningful comparisons to be made over time.

Proper record keeping greatly facilitates detection of gradual deterioration of performance over an extended period of time, by analyzing the results for degradation and initiating corrective action when necessary.

Specific quality control procedures vary between manufacturers and models, thus it is impractical to provide detailed procedures covering all types of equipment.

Initial performance testing of imaging equipment should be performed upon installation and should be completed before clinical use. Acceptance testing should be more comprehensive than periodic performance tests and should be consistent with current acceptance testing practices. Acceptance tests can usually be carried out according to the international recommendations such as those published by NEMA (National Electrical Manufacturer Association), and the technical standards published by IEC (International Electrotechnical

Committee) as well as special manufacturer's instructions. Acceptance testing is not within the scope of these guidelines and the reader is referred to the international recommendations mentioned above.

A baseline set of quality control results should be recorded, after a thorough evaluation of the system at installation and acceptance testing, to serve as a reference for the life of the equipment. These can be used as a basis for developing detailed protocols for individual systems and models of equipment.

It is important to recognize that there will need to be a financial investment in instrument quality control tools, such as phantoms. This investment is modest when compared to the value of the services performed for a patient.

Nuclear Medicine practitioners are encouraged to call on the advice of experienced Medical Physicists to draw up detailed QC protocols for their specific equipment based on quality assurance references and the guidelines presented here.

As advances in medicine occur at a rapid rate, the review and update of guidelines, such as these, should take place at regular intervals and should be considered to be part of the quality assurance process.

II Background and Procedures

A Non-Imaging Systems QC

1 Dose Calibrator QC

1.1 System description

Dose calibrators are ionisation chambers. They are used for measuring the activity of radionuclides and radiopharmaceuticals administered to the patient. The ionisation chamber may cover an energy range of 60 keV to 2 MeV to allow measurement of the full range of medically-relevant radionuclides, including the higher energy nuclides used for PET. Because radiations of different types and energies produce different amounts of ionisation, equal activities of different radionuclides generate different quantities of current. Isotope selectors provided on the dose calibrator are the feedback resistors to compensate for the differences in ionisation produced by different radionuclides so that equal activities produce the same reading.

1.2 Dose Calibrator quality control tests

1.2.a Background

1.2.a.i Rationale

Contaminations in the chamber and in the surrounding space or electronic drift can influence the background reading and measurements taken will be systematically either too high or too low. Therefore, contaminations should be eliminated, if possible, and zero offsets and general background should be set to zero with the controls provided on most dose calibrators.

1.2.a.ii Methodology

A measurement without an activity source has to be done. To check for contamination of the chamber, background should be measured with and without the sample holder. The reading must be compared with a reference value.

The reference value has to be determined at installation as the mean value of multiple background measurements over several days.

1.2.a.iii Acceptable Performance

The actual background value should read between 50% and 200% of the reference value.

1.2.a.iv Frequency

Daily. If very low activities are measured it should be carried out before each activity measurement.

1.2.b Constancy

1.2.b.i Rationale

Changes in calibration or malfunction of the dose calibrator over time, e.g. because of electronic drift, can result in readings either too high or too low relative to the real activity measured.

1.2.b.ii Methodology

For constancy check, a source with a long half – life (e.g. Co-57, Cs-137) is measured.

The necessity to check constancy in all isotope selector positions or only in the commonly used depends on the used system (e.g. digital system: only one isotope selector position is necessary).

1.2.b.iii Acceptable Performance

The variation of the displayed activity must not exceed ±5% relative to the reading predicted from the decay of the check source. Larger discrepancies should be investigated and if necessary, the dose calibrator must be recalibrated, repaired, or replaced.

1.2.b.iv Frequency

Daily

1.2.c Linearity

1.2.c.i Rationale

The linearity test indicates the dose calibrator's ability to measure the activity accurately over a wide range of values.

1.2.c.ii Methodology

The test is carried out by assaying a very high-activity Tc-99m source over a time period of about 10 half – lifes, and plotting the activity against time. The timing of measurements should be selected to ensure a reading at each power of ten (e.g. with a period of 12 hours).

As an alternative, the different activities can be achieved by dilution.

1.2.c.iii Acceptable Performance

The measured activity should agree with the calculated activity of Tc-99m indicating the linearity of response in the range of activities of interest. If the error of any measurement exceeds ±5% relative to the regression line, correction factors must be determined or the calibrator must be repaired.

1.2.c.iv Frequency

Half-yearly.

1.2.d <u>Accuracy</u>

1.2.d.i Rationale

Accurate measurement of the activity administered to the patient is important because activities which are too large will result in unnecessary radiation exposure to the patient. For diagnostic procedures, an administered activity which is too low may prolong the study time or result in sub-optimal images and, for therapeutic procedures, a sub-optimal therapeutic effect may result.

1.2.d.ii Methodology

Accuracy of the dose calibrator can be determined directly or indirectly.

For a direct accuracy check, an official reference source has to be measured for each radionuclide.

To check the accuracy indirectly, a comparison with a directly calibrated dose calibrator of the same design is possible. In this case, a comparative measurement of at least three sources has to be performed. Usable radionuclides, for example, are Am-241 for the low energy region, Ba-133 for the middle energy region, and Ra-226 for the high energy region.

1.2.d.iii Acceptable Performance

For direct testing, the variation of the displayed activity must not exceed ±5% relative to the reading predicted from the reference source.

For indirect testing, the reading of both dose calibrators must not differ more than ±5%.

1.2.d.iv Frequency

Accuracy should be checked upon installation or after repair (normally done by the manufacturer).

1.2.e Geometry

1.2.e.i Rationale

Variations in sample volume or other geometric configurations can affect the accuracy of measurement, particularly in low-energy radiations. Therefore, if different samples are often used, a correction factor for the different geometric configurations has to be determined.

1.2.e.ii Methodology

For each radionuclide, the same activity must be measured with the different geometric conditions used. The readings must be compared with the reading of the geometric conditions, which are used normally.

1.2.e.iii Acceptable Performance

If the difference of various readings of the same activity exceeds ±10% because of different geometric configurations, a correction should be carried out.

1.2.e.iv Frequency

The geometry correction factor should be determined upon installation or after repair.

2 Well counter and NaI(Tl) – Probe QC

2.1 System description

The well counter consists of a thallium activated sodium iodide NaI(Tl) detector with a central bore and associated electronics such as a photomultiplier tube (PMT), preamplifier, amplifier, pulse – height analyser (PHA), and scaler – timer. Placing a radioactive sample in the central bore of the detector increases geometric efficiency (almost 99%) and hence counting efficiency of the counter. The well counter is shielded with about 8.5 cm thick lead to reduce background from cosmic rays, natural radioactivity, or background activity in the work area.

NaI(Tl) – probes consist of a detector, about 5 cm in diameter and 5 cm in thickness, and other associated electronics, as in well counters. One of the differences between NaI(Tl) – probes and well counters is that the first requires a collimator, which limits the field of view. The collimator is a 20 to 25 cm long cylindrical barrel made of lead and covers the detector as well as the PM tube. The efficiency of a probe varies inversely with the distance between the detector and the radioactive source.

2.2 Quality control procedures

2.2.a <u>Energy window setting</u>

2.2.a.i Rationale

Incorrect energy window setting decreases the sensitivity of the system and leads to reduced readings. Therefore, the photopeak has to be centred within the energy window.

2.2.a.ii Methodology

The energy window setting has to be controlled by visual inspection. If visual inspection of the energy spectrum is not possible, the right energy window setting can be controlled by determination of the measurements sensitivity.

2.2.a.iii Acceptable Performance

Any difference of the energy window centre should be corrected

2.2.a.iv Frequency

Daily for each used isotope

2.2.b Background level measurement

2.2.b.i Rationale

High levels of background radiation may arise from a contamination of the counter or probe, the floor, or the walls. Other possible sources are “hot” patients or other unshielded radiation in the proximity of the measurement system. Where the measurement sample has low level radioactivity, background radiation can lead to significant errors and decreases the sensitivity of the well counter or probe. Therefore, measured counts must be background – corrected.

2.2.b.ii Methodology

The background level measurement should always be carried out with the same geometric conditions. The measurement time should be 1 minute. The count rate must be compared with the reference value.

The reference value has to be determined at installation as the mean value of multiple background measurements over some days.

2.2.b.iii Acceptable Performance

The actual background value should read between 50% and 200% of the reference value.

2.2.b.iv Frequency

Daily for each used isotope

2.2.c Constancy

2.2.c.i Rationale

Over time, the sensitivity of well counters and NaI(Tl) probes can be diminished because of voltage drift. To detect such a sensitivity reduction constancy of the system must be checked.

2.2.c.ii Methodology

Constancy should be checked using a long – lived source, such as Cs-137 or Co-57. Because count rate is influenced by geometric variations, measurements have to be performed under the same conditions. As an alternative, measurements sensitivity can be determined.

2.2.c.iii Acceptable Performance

The variation of the measured count rate must not exceed ±5% relative to the reading predicted from the decay of the check source.

2.2.c.iv Frequency

Weekly.

If the variation is out of range, energy calibration (see below A2.2.e) should be performed on non digital systems. Chi-square statistics (see A2.2.f) should be performed on all systems.

2.2.d Sensitivity Measurement

2.2.d.i Rationale

If the absolute activity of the measured sample is necessary, the sensitivity measurements for the γ– ray energy of interest must be determined using a standard of the radioactive sample with known activity. This correction factor can be applied to the count-rates of samples of unknown radioactivity when counted at the same settings as the standard to give the absolute activity.

2.2.d.ii Methodology

The count-rate of a probe with known activity should be determined using a measurement time of one minute. The result should be given in **counts / minute * MBq**. Since the count-rate varies with sample size, sensitivity should be given for a 1 ml sample.

Measurements sensitivity should be determined for each isotope used. For isotopes with a short half-life the radioactive decay between the timings of the dose calibrator and probe measurements must be considered.

If measurements sensitivity is determined instead of constancy check, only the isotope, which is used with the highest frequency, needs to be assessed.

At installation, a reference value for each used isotope must be determined.

2.2.d.iii Acceptable Performance

The variation of the determined measurement sensitivity must not exceed ±5% relative to the reference value.

2.2.d.iv Frequency

Half-yearly.

If measurement sensitivity is used instead of window setting or constancy, the frequency of these tests has to be considered.

If the variation is out of range, energy calibration (A.2.2.e) should be performed on non digital systems.

2.2.e Energy calibration

2.2.e.i Rationale

It is essential that the dial settings of the discriminators on the PHA are calibrated so that the dial readings can be read in units of keV. This calibration is called the energy calibration and is necessary for non digital systems.

2.2.e.ii Methodology

The calibration is carried out by using a radioactive source with known energy of the γ–rays, e.g Tc-99m with a 140 keV photopeak. The discriminator level is set at 140 divisions and a small energy window, smaller than the full width of half maximum (FWHM) of the photopeak, is used. Starting from low values, the high voltage and the amplifier gain are increased in small increments until the observed count rate reaches a maximum.

2.2.e.iii Acceptable Performance

not applicable

2.2.e.iv Frequency

At installation and after repair.

On non digital systems also if variation of constancy and sensitivity is out of range.

2.2.f Chi – square statistics

2.2.f.i Rationale

The readings of repeated measurements must not exceed statistical variations. This can be checked by chi-square statistics.

2.2.f.ii Methodology

For a minimum of ten measurements (n) of the same source, the mean value ($\overline{N}$) of the measured counts (N_i) must be calculated. Each measurement should reach a minimum of 10,000 counts. Using these values, chi-square can be calculated:

$$X^2 = \frac{\sum_{i=1}^{n} (Ni - \overline{N})^2}{\overline{N}}$$

2.2.f.iii Acceptable Performance

The value of x^2 has to be within 4.2 and 14.7.

2.2.f.iv Frequency

The chi-square statistics should be determined at installation, after repair and if variation of constancy is out of range.

3. Hand-held gamma probes

3.1 System description

Hand-held gamma probes are small, transportable NaI(Tl)-probes which are especially used for the detection of sentinel lymph nodes.

3.2 Quality control procedures

For these probes only background level measurement and constancy check are necessary. If the constancy check results in a count rate outside the acceptable values, energy calibration must be carried out.

B Planar Gamma Camera QC

1 System Description

Gamma cameras detect radiation from the entire field of view (FOV) simultaneously and therefore are capable of recording dynamic as well as static images of the area of interest in the patient.

1.a System components

The gamma camera consists of several components: a collimator, a detector, typically a NaI(Tl) crystal, multiple PM tubes, a preamplifier, an amplifier, a PHA, a X-, Y- positioning circuit, and a display or recording device. Gamma cameras are used as single detector systems, multi detector systems, and whole body systems.

1.b Performance parameters

The performance of a gamma camera is affected by several parameters including spatial resolution as a combination of intrinsic resolution, collimator resolution, and scatter resolution as well as sensitivity and uniformity. Quality control procedures are described for these parameters. In addition, the final result of an acquisition depends on contrast, matrix size, and time per frame. Therefore, selection of appropriate acquisition parameters has an important bearing and consideration needs to be given to the following:

1.b.i Contrast

Contrast of an image is the relative variation in count densities between adjacent areas in the image of an object and gives a measure of detectability of an abnormality relative to normal tissue. Several factors affect the contrast of an image, namely, count density, scattered radiation, size of the lesion, and patient motion. For a given imaging setting, a minimum number of counts needs to be collected for reasonable image contrast and contrast is improved with increasing administered activity or increasing imaging time.

Background in the image increases with scattered radiation and thus degrades the image contrast. Narrow PHA window settings can reduce the scatter radiation, but also sensitivity. Therefore, as a compromise, a 15 – 20% PHA window centered on the photopeak of interest is most commonly used in routine imaging.

To reduce patient motion, the patient should be restrained or rested in a comfortable position.

1.b.ii Matrix size and time per frame

In a static study, the choice of a matrix size depends on the field of view of the imaging system, the extrinsic resolution and the expected signal to noise ratio in the data. Normally, a pixel size of 2 to 3 mm is considered appropriate for good image resolution. Thus, for large FOV gamma cameras, one would need a 256 x 256 matrix to obtain the above pixel size. The time per frame depends on the count rate in the image and should be long enough to get an appropriate contrast.

Data in dynamic studies can be collected in a sequence of several phases and the time per image can be different in each phase. The choice of time per frame for a given study depends on the kinetics of the radiotracer through the organ of interest. Especially in high kinetics, time per frame is short, resulting in low count rates. Therefore, to get adequate image statistics, a matrix size of 64 x 64 or 128 x 128 is used.

2 Quality Control procedures

2.a Visual Inspection

2.a.i Rationale

A visual inspection may reveal obvious defects, which may compromise the safety or the imaging efficacy of the system. Of particular importance is the visual inspection of collimators. Signs of denting or scratching may indicate mechanical damage to the collimator, and surface stains may be a sign of possible contamination. Both of these may produce artefacts such as cold or hot spots on planar images and rings on SPECT images.

2.a.ii Methodology

A general visual inspection of collimators and for any other defects, which may compromise patient or staff safety (e.g. frayed or damaged electrical cables, mechanical faults in the camera or scanning table) should be carried out.

2.a.iii Acceptable Performance

If signs of new dents, scratches or stains are recognized, a background level measurement and possibly an extrinsic uniformity check should be performed before a suspect collimator is used for patient imaging.

If any other defects are detected, the equipment should not be used until it is established that it is safe to do so.

2.a.iv Frequency

Before any patient study and whenever collimators are changed

2.b <u>Background level measurement</u>

2.b.i Rationale

Background radiation, if it is of sufficient intensity, has the potential to compromise any type of imaging. Even moderately elevated background levels have the potential to seriously degrade intrinsic uniformity or other intrinsic measurements. High levels of background radiation may arise from a contamination of the camera, especially the collimator, the floor, or the walls. Other possible sources are “hot” patients or other unshielded radiation in the proximity of the imaging system.

2.b.ii Methodology

The background level measurement should be carried out with the same collimator and in the same camera position, using the energy window which is most frequently used for imaging. The count rate (measurement time one minute) must be compared with the reference value. The reference value has to be determined at installation as the mean value of multiple background measurements over some days.

To simplify the daily procedure, a reference background level can be performed for each collimator.

An increased count rate indicates a contamination or an unshielded radiation source in the proximity of the camera which has to be eliminated before imaging can proceed.

2.b.iii Acceptable Performance

The actual background value should read between 50% and 200% of the reference value.

2.b.iv Frequency

Daily

2.c Photopeak and window setting

2.c.i Rationale

Incorrect energy window setting can degrade uniformity, reduce sensitivity or can increase the scatter contribution to the image. Particularly in older cameras, the photopeak can change due to slight variations in high voltage, photomultiplier drift, changes in temperature and other factors.

2.c.ii Methodology

Peak settings should be checked and adjusted in a consistent manner and the settings should be recorded to detect long term drift. Sudden changes in peak setting indicate a possible fault in the camera and should be fully investigated and rectified if necessary before the camera is again used for clinical studies.

It is important to check the energy window settings for all radionuclides used on the gamma camera for the day. Proper peak settings for one radionuclide (e.g. Tc-99m) do not necessarily mean that the window settings for other radionuclides are correct. In particular, if a change in the peak setting for one radionuclide is detected, it is likely that the settings for other radionuclides also need to be adjusted.

Using a point source, peaking should preferably be performed intrinsically (without collimator), to reduce scatter and to ensure that an average peak for the whole field of view (FOV) is obtained. The count rate should not exceed 20k cps.

2.c.iii Acceptable Perfomance

If the discrepancy of photopeak and window centre is greater than ±3% of window centre, the camera must be recalibrated before imaging can proceed.

2.c.iv Frequency

Daily

2.d Sensitivity Measurement

2.d.i Rationale

The sensitivity depends on the geometric efficiency of the collimator, the detection efficiency of the detector, PHA discriminator settings, and the dead time. A deviation of the sensitivity from the reference value indicates faults in the crystal, in the electrical camera units, or a degradation of the energy resolution.

2.d.ii Methodology

To measure the sensitivity, a source with a γ-energy <200 keV should be used. The collimator, the source-collimator distance, the geometrical conditions, and

the energy window must be always identical. If a Tc-99m source is used, the decay between the timings of the dose calibrator and camera measurements must be considered.

Using high-energy collimators, sensitivity of a point source depends on the position of the source at the collimator because of the septal thickness. Therefore, if sensitivity measurement is carried out using high-energy collimators, a surface source should be used instead of a point source.

A reference value has to be determined at installation.

2.d.iii Acceptable Performance

The measured values of the sensitivity should not exceed ±5% of the reference value.

2.d.iv Frequency

Weekly

2.e <u>Uniformity</u>

2.e.i Rationale

Interpretation of clinical images taken with the gamma camera rely on the assumption that differences seen are due to differences in the patient only and not differences introduced by the gamma camera. A large number of possible problems in the gamma camera can degrade uniformity. It is thus a good general QC test of proper performance of the camera. Uniformity defects can be quite marked and focal, such as during a failure of a photomultiplier tube, or there can be general degradation of uniformity across the field of view due to inappropriate spatial linearity or energy corrections. Further QC tests may thus be required to detect the cause of the observed non-uniformity.

2.e.ii Methodology

Uniformity can be checked either without collimator (intrinsic) or with collimator (extrinsic). Intrinsic uniformity is simpler to perform and does not require a flood tank or sheet source. For planar imaging, only intrinsic uniformity check is necessary.

To detect gradual deterioration in uniformity, it is important that uniformity measurements are carried out in a consistent manner (e.g. same orientation, same number of counts, etc.) and records are kept to allow comparisons over periods of weeks or even months.

Regular analysis of uniformity by a computer can facilitate detection of gradual deterioration prior to any visible change. Two different uniformity parameters can be quantified: integral uniformity and differential uniformity.
Integral uniformity is a measure of the maximum pixel count deviation in the field of view:

$$\textbf{Integral uniformity} = 100 * ((Max - Min) / (Max + Min))$$

Differential uniformity is a measure of the maximum deviation of a limited area designed to approximate the size of a photomultiplier tube, usually 5 x 5 pixels.

$$\textbf{Differential uniformity} = 100 * ((Max_{area} - Min_{area}) / (Max_{area} + Min_{area}))$$

The calculation should be done for the X- and the Y-directions independently and the maximum change represents the differential uniformity.
To perform an intrinsic uniformity check, a source (typically Tc-99m) should be suspended > 5 x FOV of the gamma camera away from the detector.
To perform an extrinsic uniformity check, a flood source is placed on the detector with a collimator attached. The flood source must exceed the field of view on all sides by a minimum of 20 mm. The overall total thickness (bottom, fluid, cover) must be at least 80 mm. The variation of the fluid thickness in the whole source must not exceed ±1%. As an alternative, a Co-57 sheet source is usable.
The count rate should not exceed 20 kcps. A measuring time should be selected, that guarantees a count density > 5000 counts per cm². The camera should be peaked and a consistent set up has to be used (e.g. same distance of source, same orientation, same radionuclide, same collimator). Significant background radiation from other sources should be eliminated.
Pronounced non-uniformity in the image should be checked visually. Windowing may be used to highlight non-uniform areas if the study is collected on computer. If possible, a computer-based quantitative analysis should be used.

2.e.iii Acceptable Performance

The non-uniformity should not differ from that in the reference image, which is prepared at installation.
If a quantitative analysis is used, intrinsic integral uniformity should be < 8%

2.e.iv Frequency

Weekly

2.f Spatial Resolution

2.f.i Rationale

Spatial resolution refers to the ability of the camera to image two separate objects as they are moved closer and closer to each other in space. Because the images of two sources will blur together at a separation approximately equal to the FWHM of the images of the single source, one method of estimating spatial resolution of an imaging system is to determine the FWHM of a source.

2.f.ii Methodology

For periodic evaluation of the spatial resolution, a semi quantitative method using bar phantom is employed. Bar phantoms consist of four sets of parallel lead bar strips arranged perpendicular to each other in four quadrants. The widths and spacing of the strips are the same in each quadrant but differ in different quadrants. The thickness of lead should be sufficient to stop photons of a given energy for which spatial resolution is being estimated.

To determine the spatial resolution, the bar phantom is placed over the detector of the gamma camera. A flood source of equivalent dimension containing sufficient activity (usually, 200 to 400 MBq) is placed on the top of it and an image is taken. The image is visually inspected and spatial resolution is estimated from the smallest strips or spacing distinguishable on the image. If the image is taken with a collimator (extrinsic) the spatial resolution depends primarily on the collimator resolution.

An improved method is based on the use of a line source. The line source is placed in the field of view and an image is obtained. The FWHM value of the resulting line in the image gives the spatial resolution. For routine quality checks, the use of a bar phantom is sufficient.

2.f.iii Acceptable Performance

The smallest width and spacing of the strips that is distinguishable on the image should be < 4 mm for intrinsic measurement and < 6 mm for extrinsic measurement.

2.f.iv Frequency

Half-yearly

2.g Linearity

2.g.i Rationale

Non-linearity diminishes the spatial resolution of the gamma camera and leads to artefacts in whole body scans and SPECT.

2.g.ii Methodology

Linearity is checked using the procedure of uniformity determination and using a bar phantom that consists one set of parallel lead bar strips. The strips should be 1 mm thick and 30 mm one from the other. The strips should be rotated by 45° to the X- and Y-direction.

Using special programs, the deviation of the strips from a straight line can be determined. This deviation is expressed in mm, and like uniformity, it is measured for both integral non-linearity (maximum deviation) and differential non-linearity (point-to-point deviation). Linearity can be checked as intrinsic linearity (without collimator) or extrinsic linearity (with collimator). For extrinsic linearity check a flood tank is necessary.

2.g.iii Acceptable Performance

If non-linearity is estimated visually, the actual image should not differ from a reference image prepared at installation.

Limits of quantitative non-linearity: Up to now, general accepted limits do not exist. The value of the resolution is suggested as a practical useable value.

2.g.iv Frequency

Half-yearly

2.h Whole body scan Resolution

2.h.i Rationale

To avoid loss of resolution in the scanning direction during whole body scans, the relative physical position between bed and detector has to be accurately synchronised with the electronic offset applied to the image data to form the whole body image. Both mechanical problems and drift or inappropriate adjustment of image offset or size can cause a loss of resolution for whole body scans.

2.h.ii Methodology

To perform whole body scan resolution measurements, multiple sources separated by equal distances should be placed on the bed (e.g. 6 sources with a distance of 30 cm). The detector should be placed as close as possible to the

bed. A whole body scan should be collected at all speeds used for patient scanning. Finally the distance of sources at the whole body image should be compared with the real distance on the bed.

2.h.iii Acceptable Performance

There should be no difference between the distance at the image and the real distance on bed.

2.h.iv Frequency

Half-yearly

C Single Photon Emission Tomography (SPECT) QC

1 SPECT acquisition system

The term SPECT stands for Single Photon Emission Computed Tomography, (sometimes shortened to SPET).

1.a System components

A SPECT acquisition system essentially comprises a rotating gamma camera, which may have one, two or three detectors. The principle of the SPECT technique is to acquire a number of images at small angular steps around the object. This creates a series of 2-D projections or projection images, which are then processed using a reconstruction algorithm to yield estimates of the 3-D distribution of radioactivity in the object. The acquisition and processing of the data is controlled by the nuclear medicine computer system.

For single detector systems, the gantry allows full 360-degree rotation; this is generally possible with most of the multi-detector systems, although with such systems 360 degrees of data can be acquired with a reduced arc of rotation for each detector.

1.b Performance parameters

The performance of a SPECT system is affected by all of the factors which influence that of a planar gamma camera. The performance requirements for SPECT are more stringent, requiring tighter control to achieve acceptable tomographic image quality. The selection of appropriate acquisition parameters has an important bearing on the final result and consideration needs to be given to the following:

1.b.i Acquisition matrix size

To preserve resolution in the digital image matrix, the linear pixel dimension must be not greater than half the resolution of the camera at the position of the object of interest. Typically this will require a matrix size of 64 x 64 or 128 x 128 for higher resolution studies. It is not uncommon to apply a digital zoom during acquisition of tomographic images (for example, when imaging the heart) when a matrix size of 64 x 64 is used, in order to achieve an appropriate pixel size. In

such circumstances, users must be aware of the potential effects of any truncation of the data.

1.b.ii Angular sampling and number of projections

Angular sampling must be sufficiently fine in order to yield an accurate reconstruction of the object distribution. The accuracy achievable is dependent upon the spatial resolution (r) of the camera. The relationship between the required number of projections (N) over 360 degrees and the spatial resolution is given by $N = \pi D(r/2))$ where D is the field size.

1.b.iii Acquisition mode

Projection data are generally acquired in either step and shoot mode or continuous mode. Continuous rotation is the most efficient in terms of data collection, but is at the expense of some blurring as data are acquired over a range of angles for each projection. It is preferable for fast rotations (for example, dynamic SPECT using cameras with slip ring technology) and in conditions where the stepping time between views is more than 10% of the imaging time per projection.

1.b.iv Time per projection

The Signal to Noise Ratio (SNR) improves with the number of counts collected and therefore to improve image quality the acquisition time per projection should be as long as practicable to give adequate counting statistics. In practice the total acquisition time of a clinical SPECT study is limited by the length of time the patient can remain still. Generally no longer than 30 – 45 minutes is recommended. This limits the time per projection, according to the number of projections and acquisition mode selected. Shorter acquisition times may be needed if the radioactive tracer shows physiological redistribution, in order to minimise artefacts in the reconstructed data. Appropriate choice of starting angle may also assist in minimising redistribution artefacts (for example, bladder-filling artefacts can be reduced by commencing the acquisition sequence with the posterior projection). Multi-detector systems have increased sensitivity compared with single detector systems. This can be exploited either in terms of a shortened total acquisition time or an improved SNR.

1.b.v Collimation

The quality of the SPECT image is governed by its SNR, which can be improved either by increasing contrast (through improved spatial resolution) or by decreasing noise (through improved sensitivity). In general, a high-resolution

collimator is preferred, as higher resolution imaging requires relatively fewer acquired counts for an equivalent SNR. The choice of collimation is a balance and will be application dependent.

1.b.vi Radius of rotation

For parallel hole collimation, this should be as small as practicable, consistent with maintaining reproducibility, in order to maximise spatial resolution.

1.b.vii Reconstruction

The reconstruction of the acquired data to create transaxial slices is user-definable in terms of technique and parameters. Most manufacturers offer both Filtered Backprojection (FBP) and Iterative Reconstruction (IR). The choice of filter for FBP is not an exact science. It is dependent on the noise and frequency content in the data, which will in turn be dependent on the underlying radioactivity distribution, the activity, the equipment performance and the acquisition parameters. It is recommended that filter selection is application-dependent but that it is not varied between patients undergoing the same examination. IR is computationally intensive but has the advantage of removing some of the image artefacts associated with FBP and incorporating corrections for physical effects such as attenuation, scatter and depth-dependent resolution. A thorough evaluation of the influence of reconstruction parameters on image quality is recommended to ensure appropriate usage and correct image interpretation.

2 SPECT Quality Control procedures

2.a Detector Head Tilt

2.a.i Rationale

For accurate tomographic imaging it is essential that the detector(s) is aligned correctly [parallel to the axis of rotation] and that the angular readouts displayed on the gantry provide a true representation of detector head position. If this is not the case, errors will be introduced into the reconstruction due to an incorrect geometrical configuration.

2.a.ii Methodology

The position of each detector when aligned in a horizontal position as defined by the gantry angular readouts should be checked in the X and Y directions using a spirit level.

2.a.iii Acceptable performance

Any deviation in detector position from the horizontal implies that there is detector misalignment or just an error in the angular readout, which should be investigated.

2.a.iv Frequency

This check should be combined with the centre of rotation measurement on a monthly basis. For clinical tomographic imaging, a visual check should be made that the detector position is set to the required angle prior to each acquisition.

2.b Centre of Rotation (COR)

2.b.i Rationale

During the reconstruction of tomographic projection data, it is assumed that the centre of each image corresponds to a single fixed point in space. If there is some lateral movement in the detector head position, for example caused by gantry sag, then this will introduce errors into the position of the centre of the image and cause artefacts in the reconstructed data.

It is inevitable that the mechanical rotation of heavily shielded large field of view detectors will not be accurate about a single point and as such internal corrections must be applied to the acquired data to minimise the effect of any lateral shift.

Testing is commonly performed by the acquisition of tomographic data for a point source, and the subsequent analysis of the stability of the image.

2.b.ii Methodology

Gamma camera manufacturers routinely include standard protocols for the acquisition and testing of the COR offsets. These may include the use of multiple point sources and/or acquisitions at multiple radii. The most common set-up for testing the COR is to draw up 20-40 MBq of Tc-99m in less than 0.1 ml in a stoppered (needle-free) syringe. This is then placed near the centre of the field of view and at approximately 10 cm off the central axis. Data is acquired in a 128x128 matrix with approximately 64 projections over an orbit of 360º and with a radius of 20 cm. Approximately 10,000 counts should be acquired per projection.

Qualitative assessment of the COR can be performed by creating a sinogram of the acquired data and looking for discontinuities in the data. Quantitative assessments involve the calculation of the centre of gravity for each frame and

looking at the variation of this over the course of the acquisition. The y-axis position should be constant. In the x-axis this should follow a smooth sinusoidal pattern of motion.

The COR must be tested for each different gantry set-up (e.g. 'H-mode' and 'L-mode') and rotation and for each set of collimators used for SPECT, since these will result in different stresses on the gantry and will commonly have different COR corrections applied.

2.b.iii Acceptable performance

Variation in the x and y axes should be less than 2 mm.

2.b.iv Frequency

COR corrections should be incorporated into a monthly QA programme for the SPECT gamma camera system.

2.c Detector Alignment

2.c.i Rationale

Multi-detector systems will either have a fixed detector orientation or allow the detectors to be located at different relative gantry positions e.g. for a two-detector system they may be facing each other (in 'H-mode') or at right-angles (in 'L-mode'). This allows for greater flexibility of scanning, particularly for SPECT imaging if one wishes to be able to scan over an arc of 360º and also 180º. The integrity of these detector orientations is essential to ensure that data is reconstructed appropriately.

2.c.ii Methodology

Acquisition parameters and source set-up are similar for COR corrections (detailed in section C 2.b), but with the source placed close to the central axis (rather than at 10 cm off-axis). A SPECT acquisition is then performed using one detector only, set at the minimum axial radius. A second acquisition is then performed using the second detector at the maximum radial distance and with the source position unchanged.

The only difference between appearance and position of the point source in corresponding projection images should be related to the loss of resolution with radial distance. This can easily be checked by subtracting the corresponding projection images from one another and confirming that each resultant image corresponds to a concentric ring shape. This should be performed for each of the detector orientations used for the system.

2.c.iii Acceptable performance

If there is any variation in appearance of the images from that of a concentric ring then this should be investigated.

2.c.iv Frequency

Head alignment should be incorporated into an annual QC programme for the gamma camera system.

2.d High Count Field Uniformity

2.d.i Rationale

The system uniformity requirements for tomographic imaging are more stringent than for planar imaging since any areas of non-uniform response will be reinforced by overlaying data from multiple projections during the reconstruction process. Focal areas of non-uniformity will result in ring artefacts in the transaxial slices of the reconstructed data set (with a radius corresponding to the off-axis distance of the non-uniformity).

When testing the response of the system for use in SPECT a higher count uniformity image must be acquired to reduce the statistical noise in the image. This image is used for the measurement of uniformity and also commonly for the creation of a uniformity correction map.

Modern gamma cameras typically include a uniformity correction map that is applied to the acquired images to minimise any fluctuations in sensitivity across the FOV and reduce artefacts in the reconstructed images. These maps are only intended to provide the final correction for a well functioning system and not to balance out non-uniformities introduced by sub-optimal function of some other part of the system e.g. failure to correct for linearity errors. One should be aware for any local system whether the QC images are displayed with or without the application of the uniformity correction and where possible the uncorrected images should be reviewed.

2.d.ii Methodology

As discussed in the section on planar gamma camera QC [section B.2.e] uniformity may be measured either intrinsically or extrinsically. If an image is to be used for the creation of a uniformity correction map then this should be acquired extrinsically for each collimator used on the system and using a uniform flood source of appropriate energy.

For assessment of uniformity, an image should be acquired of a uniform flood source covering the field of view of the system [see section B.2.e]. The image should contain 40 million counts, with a 256x256 matrix, which is suitable for qualitative image review. For calculation of integral and differential uniformity, it is usual to compress the data into a 64x64 matrix

2.d.iii Acceptable performance

The integral uniformity and the differential uniformity should be less than 4% and 3% respectively when the uniformity correction map is applied.

2.d.iv Frequency

High-count uniformity should be incorporated into a monthly QA programme for the gamma camera system.

2.e Tomographic Resolution

2.e.i Rationale

The resolution of a SPECT imaging system is worse than that attainable for a comparable system performing planar imaging due to the influence of additional factors that do not impact on static imaging, including the COR corrections. As such, a measurement of the resolution represents an overall test of SPECT image quality rather than an investigation of an individual system parameter.

2.e.ii Methodology

SPECT resolution phantoms are commercially available, typically incorporating multiple capillary line sources lying parallel to the central axis of the imaging system and allowing measurement of resolution at multiple off-axis positions. Whilst such an approach represents the most thorough quantitative assessment of the system it is not considered necessary as part of a routine gamma camera system QC. The following gives a simplified quantitative approach, adapted from NEMA 2001 and the American Association of Physicists in Medicine (AAPM) report 22.

A capillary line source should be suspended close to and parallel to the central axis of the camera, away from scattering material. A 120 frame SPECT acquisition is performed with a 128x128 matrix and a zoom of 2, acquiring at least 100 kcounts per image. The data should be reconstructed using a ramp filter alone and profiles plotted across the resultant image in both the x and y direction to find the full width half maximum (FWHM).

2.e.iii Acceptable performance

The resolution of the system for SPECT should not be more than 10% worse than that for planar imaging at a similar set-up distance.

2.e.iv Frequency

SPECT resolution measurement should be performed as part of the acceptance testing and baseline assessment for the gamma camera system and in the event of concerns about SPECT image resolution. In the case of the latter, measurements of rotational uniformity should be undertaken (see C.2.f).

2.f Rotational Uniformity

2.f.i Rationale

As discussed above (C.2.d) the uniformity response of the detectors is an important factor in the acquisition of SPECT data. It is necessary that this uniformity of response is consistent throughout the rotation of the gantry and that it has not just been optimized for a fixed position; e.g. face-up or face down. Reconstruction processes assume uniformity of response throughout the gantry rotation and any variation will introduce artefacts into the reconstructed data.

There are a number of factors which may affect image quality at different gantry angles. These include the impact of magnetic fields on photomultiplier tube (PMT) performance and the slip in position of components within the detector head. PMTs are sensitive to variations in external magnetic fields; therefore as the gantry rotates and the angle of the tube to the field varies, this has the potential to affect performance. Modern gamma cameras typically incorporate good shielding of the components from external magnetic fields.

2.f.ii Methodology

The rotational uniformity can be assessed using a lightweight flood source attached to the front face of the collimated detector. A 30 frame SPECT acquisition is then performed acquiring 10 million counts per projection into a 64x64 matrix. (This acquisition will probably need to be performed overnight.) Alternatively, with appropriate manufacturer's software, it can be undertaken using an uncollimated detector with a point source positioned at the centre of rotation. This is a simpler technique in practice, but does require the availability of software to correct for the non-uniform activity distribution from a point source positioned a finite distance from the detector.

Qualitative assessment may be performed by reviewing cine of the acquired projection data. Use of a colour scale with multiple (greater than 10) discrete colour steps may emphasise non-random areas of uniformity variation

Quantitative assessment may be performed for each projection image as for planar uniformity measurements [section B.2.e].

Alternative quantitative methods have been proposed in AAPM reports 22 and 52.

2.f.iii Acceptable performance

Rotational uniformity should be as expected for equivalent count planar images acquired using the available sheet source.

2.f.iv Frequency

Rotational uniformity should ideally be incorporated into an annual QC programme for the gamma camera system. It should be undertaken as part of an investigation when there are concerns about SPECT image resolution.

2.g Reconstruction Phantom Studies

2.g.i Rationale

As already identified there are a number of parameters which will impact on the overall quality of the images acquired using a SPECT system. One method to ensure that the images are of a consistent quality is to acquire suitable phantom data, incorporating objects for the qualitative assessment of resolution and uniformity. The most widely used is the Jaszczak ECT phantom.

The phantom is imaged under idealised conditions with signal to noise ratios (SNRs) much higher than those normally found with patient studies. While the phantom studies do not reflect true patient data, the improved quality of the images allows a more effective qualitative assessment and the identification of any changes in performance before they are likely to impact on routine clinical investigations.

2.g.ii Methodology

The phantom should allow clear identification of multiple objects in the image. Additionally it should include some objects that are on the limit of clear identification in the reconstructed images. (If all objects within the phantom are clearly resolved then it may take some significant deterioration in performance before the impact is noticed in the image.) There should also be a uniform transaxial area to allow better identification of ring artefacts.

While the methodology used will reflect the phantom available and the need for idealised data, it is expected that 120 projections and an image matrix of 128x128 will be required and that the total acquired counts will be in excess of 30M.

Data should be reconstructed using commonly used protocols to ensure that the reconstruction process is not subject to change (e.g. inadvertent changes to the use of default filters).

2.g.iii Acceptable performance

Any variation in the reconstructed image quality from reference data should be investigated.

2.g.iv Frequency

Reconstruction phantom studies should be incorporated into a six-monthly QC programme for the SPECT gamma camera system.

D Positron Emission Tomography (PET) QC

Positron tomographs provide static, dynamic, or gated images of the distribution of positron-emitting radionuclides within the body by detecting pairs of photons produced in coincidence by the annihilation of a positron and an electron. The two photons interact with the scintillator crystals and are detected within an electronic time window. The straight line along the centres of 2 detectors that are operating in coincidence mode is the Line of Response (LOR). Photomultiplier tubes connected to the detectors convert the light that is produced in detectors into an electrical signal. Positron-emission tomographic (PET) images are produced by reconstruction from the coincidence pair data.

1 Multi-headed gamma cameras for PET (GCPET)

1.1 System Description

Positron emission tomography (PET) imaging can be undertaken on a dedicated PET camera or a gamma camera modified for coincidence detection. Whilst the former technique is regarded as the gold standard, gamma camera PET (GCPET) is used in some centres where dedicated PET is unavailable. A GCPET system is a tomographic gamma camera with a minimum of two detectors which are capable of detecting 511 keV photons in coincidence. Triple headed systems have also been employed although these are less common. The principle of the technique is based on the coincident detection of the two high energy photons by the gamma camera detectors. In a manner similar to dedicated PET, the position of the photon interaction in each detector defines a line along which the source of positron emission is located.

SPECT cameras have also been used with ultra-high energy collimators to detect single 511 keV photons arising from the positron emission process. This is an inherently low resolution technique which has poor sensitivity and which has been mainly limited to cardiac applications. It is not in common use and will not be discussed further here.

1.2 Performance characteristics of GCPET

1.2.a Sensitivity

Sensitivity is limited by the detection efficiency of sodium iodide for high energy photons and the count rate capability of the gamma camera detectors. In uncollimated mode (which is the normal acquisition mode for GCPET), the detectors may be subjected to count rates of the order of million counts per second or more, if an acceptable number of true coincidences are to be recorded. This demands a much higher count rate capability per detector than for dedicated PET (where many more detectors are employed), requiring the use of modified electronic circuitry to minimise dead time and suppress pile-up effects. A thicker crystal is typically used to increase the stopping power of the detector, although this must be balanced with the use of the system for routine non-GCPET imaging. Even with these modifications, the sensitivity of GCPET is an order of magnitude less than that for dedicated PET in 3-D mode.

1.2.b Spatial Resolution

In contrast, the spatial resolution of GCPET systems is, in principle, comparable with that found in dedicated PET systems. It is better than the spatial resolution obtainable when the camera is operated in SPECT mode, as there is no degradation due to collimator performance. Whilst spatial resolution in GCPET is theoretically limited by the intrinsic resolution of the detectors, in practice, the ability to resolve objects is also highly dependent on the counting statistics and the contribution of scatter and randoms in the image.

1.2.c Attenuation

Attenuation correction (see section E) using transmission imaging is frequently employed to reduce attenuation artefacts in GCPET and to improve detectability. Transmission sources are generally X ray sources or single photon emitting radionuclide sources.

1.2.d Acquisition parameters

The performance of GCPET is dependent on the acquisition parameters selected. In particular the user can select to acquire in what is commonly known as 3-D or 2-D mode. In the former, the axial acceptance angle is at a maximum,

limited only by the use of lead side shields to reduce acceptance of out-of-field events. The latter configuration uses one-dimensional 'collimation' in the axial direction through the use of lead septa, which reduces the axial acceptance angle. The choice is governed by the specific application and is a trade-off between data quality (reduction of scatter and random events) and sensitivity for true coincidence events. The other main user-selectable acquisition parameter relates to the energy window used. The inclusion of 'Compton coincidences' as well as coincidence events falling within the photopeak window serves to increase sensitivity but at the expense of increasing detection of scatter and randoms. The choice is application and system dependent and requires knowledge of the performance characteristics in the different configurations.

1.3 GCPET Quality Control Procedures

A GCPET system will generally be used for both SPECT and PET acquisitions at different times and will therefore need to satisfy the requirements of both SPECT and PET QC programmes. Several of the SPECT QC routines are important for correct operation in GCPET mode (for example, centre of rotation and rotational variations in detector responses). Specific GCPET QC should include the parameters/measurements listed below.

Note that in most cases the details of how the measurements are to be undertaken, and their frequency, will be specific to each manufacturer and will be dependent on their own designs for optimising high energy performance. Tests will usually require the use of positron sources. Longer lived small volume sources, such as Na-22 and Ge-68/Ga-68, are often suitable for such purposes. Where relevant, QC tests should be undertaken using the acquisition parameters which will be used for clinical purposes.

1.3.a Photopeak position and Energy Resolution

1.3.a.i Rationale

In most systems operating in coincidence mode the energy window adjustment will be automatic under software control. However it is important to ascertain that the main energy window is correctly positioned over the 511keV photopeak and that the energy resolution is within required limits.

1.3.a.ii Methodology

Typically, a low activity, low scatter positron source will be required. The full width at half maximum (FWHM) of the energy spectrum of data acquired in coincidence mode should be calculated and compared with baseline values.

This is a test of overall detector integrity, with out of limits performance indicating possible mismatch of photomultiplier gains or inconsistencies in light collection.

1.3.a.iii Acceptable Perfomance

Energy resolution for a NaI(Tl) crystal should better than 15%.

1.3.a.iv Frequency

It is recommended that photopeak position and energy resolution is checked daily[1].

1.3.b <u>Uniformity of detector response</u>

1.3.b.i Rationale

Acceptable uniformity of response at the energies employed in SPECT imaging does not necessarily imply acceptable uniformity when the camera is used in GCPET mode. Uniformity tends to be a less significant parameter for GCPET than for SPECT since, without the use of parallel hole collimators, different points on the detectors will detect events occurring along the same line of response. Differences in count rates and processing circuitry mean that uniformity must be independently assessed in high energy mode.

1.3.b.ii Methodology

Some systems allow both singles and coincidence floods to be acquired; in all cases comparison with reference values is essential.

1.3.b.iii Acceptable Perfomance

Uniformity of the system should be comparable with that for planar imaging on the system, although this will vary between systems depending on whether uniformity correction is applied.

1.3.b.iv Frequency:

Measurements should be carried out daily[1] or weekly according to manufacturer specifications.

[1] Refers to each day the camera is used for GCPET

1.3.c Count Rate Performance

1.3.c.i Rationale

The peak coincidence rate and singles rate and the activities at which these occur provide an indication of the performance and integrity of the coincidence electronics.

1.3.c.ii Methodology

This is usually assessed using a decaying high activity source at acceptance testing and forms the basis of acquisition optimisation for future clinical studies. Some manufacturers provide simplified software tests of coincidence electronics, which can be carried out using low activity long lived sources as part of regular QC procedures.

1.3.c.iii Acceptable Perfomance

Measured count rates will vary greatly with set-up (e.g. acceptance angle). Comparison should be made with performance at acceptance.

1.3.c.iv Frequency

Measurements should be carried out bi-annually.

1.3.d System sensitivity

1.3.d.i Rationale

Variations from baseline values in system sensitivity can be an indication of malfunction in GCPET electronics. Such faults would not necessarily be evident from planar or SPECT QC tests.

1.3.d.ii Methodology

A simple measure of system sensitivity can be obtained from a tomographic acquisition in coincidence mode of a low activity low scatter point source. Acquired counts in a standardised reproducible acquisition are related to the known activity of the point source and compared with baseline values.

1.3.d.iii Acceptable Perfomance

Sensitivity for an unattenuated point source should 2000cps-per-MBq or better for a NaI(Tl) crystal, although this will depend on the crystal thickness and set-up (e.g. acceptance angle and energy window). A 19mm thick crystal will have a sensitivity of over three times greater than that of a 13mm crystal.

1.3.d.iv Frequency

This is a useful daily[1] QC test that can easily be performed. More detailed phantom sensitivity measurements (see below D.1.3.g) are recommended at monthly or bi-monthly intervals, depending on the stability of the system.

1.3.e Spatial resolution

1.3.i Rationale

Measurements of spatial resolution give information about the ability of the system to localise events in the crystal and the mechanical integrity of the system when rotating, as well as the reconstruction process.

1.3.e.ii Methodology

Provided the point source is small enough (~1mm), the same acquisition used for system sensitivity can also be used to assess system spatial resolution. The FWHM of the reconstructed point source in the axial, radial and tangential directions provides the parameter for baseline comparison.

1.3.e.iii Acceptable Perfomance

The resolution should be 5mm or better for a NaI(Tl) crystal.

1.3.e.iv Frequency

Measurements should be carried out bi-annually.

1.3.f Attenuation correction (AC)

Quality control of transmission imaging used for GCPET AC will be dependent on the technique used. In many cases this will be similar to that required for AC for SPECT imaging. Refer to section E and manufacturers' information.

1.3.g System calibration

1.3.g.i Rationale

Where GCPET systems are used for quantitative measurements, it is necessary to establish calibration factors, which convert counts/second/voxel in the reconstructed image to Bq/cm^3. Whilst such factors are established at acceptance testing, it is important that they are monitored over time to ensure that they continue to accurately reflect the current performance of the camera.

1.3.g.ii Methodology

Calibration factors are determined from measurements made using a phantom based on NEMA specifications filled uniformly with an accurately known quantity

of a positron emitting isotope. Count rates should be such that they represent conditions under which the calibration factors would be used in the clinical situation.

1.3.g.iii Acceptable Perfomance

Measurements should be compared with the results of acceptance testing. (There will large variation between systems depending on the AC method used and the crystal thickness.)

1.3.g.iv Frequency

Frequency of calibration factor determination is dependent on the stability of the system.

It should be noted that the reconstructed images from the uniform phantom could be used to provide an assessment of image uniformity.

2 State-of-the-art dedicated PET scanners (PET)

2.1 System Description

A state-of-the art PET camera consists of several full-ring detectors crystals of BGO (bismuth germanate orthosilicate), LSO (lutetium orthosilicate) or GSO (gadolinium orthosilicate). Block detectors, created by partially cutting a large block of scintillator material, have been designed and used in PET cameras enabling the use of small detectors with accurate localisation capability, with a reduced number of photomultiplier tubes (PMT) compared with conventional scintillation detector technology.. The block detector design reduces the dead time compared with that of the gamma cameras. Many block detectors are contained in a PET scanner and are arranged in 18-32 arrays in full rings. A PET detector array usually consists of tens of thousands individual crystals.

Variations on this basic design include the partial ring dedicated PET scanner and the dedicated PET scanner with six position-sensitive sodium iodide (NaI) detectors. The partial-ring rotating PET scanner was developed to reduce the cost of dedicated PET scanner. It has about 45% of the PET crystals of the corresponding full-ring system, but it employs 3D acquisition and reconstruction to compensate for the loss of efficiency.

The following section of the guidelines relates only to the dedicated scanner with full rings of BGO, LSO or GSO detectors.

2.1.a Two-dimensional (2D) and three-dimensional (3D) acquisition

In a full-ring PET scanner, lead or tungsten ‘septa’ are normally used to separate adjacent rings of detectors and coincidences are only recorded between detectors within the same ring or lying in closely neighbouring rings. This is referred to as two-dimensional (2D) acquisition mode.

PET scanners also have the capability to retract the interplane septa from the field of view and acquire coincidence data at all possible angles in three dimensions. This is referred to as three-dimensional (3D) acquisition mode and results in a substantial increase in detection efficiency.

Compared with 2D PET, 3D PET imaging has slightly worse axial resolution and a significantly higher contribution of scatter and random coincidences, but 3D PET has much better sensitivity than 2D PET imaging.

2.1.b PET crystal properties

The main characteristics that determine the suitability of a PET crystal detector are: stopping power, linear attenuation coefficient. (cm^{-1}), index of refraction, light yield, light decay time, ruggedness (non-fragile), non-hygroscopic properties

2.1. c Coincidence detection

Coincidence events in PET fall into 4 categories: true, scattered, random and multiple.

2.1.d Image Reconstruction

The method of image reconstruction employed can markedly influence image quality in PET studies. Standard vendor-provided reconstruction algorithms, use either filtered back projection or the manufacturer's recommended iterative reconstruction algorithm with an appropriate filter. (see also section C.1.b).

2.2 Performance parameters

All positron emission tomographs (PET) should be tested on installation and monitored at least annually to ensure that they are functioning within the manufacturer's specifications and accepted performance standards. Additional performance monitoring may be necessary in certain situations of major maintenance.

Performance tests can usually be carried out according to the international recommendations. The parameters can be measured using phantoms containing positron emitters (such as cylindrical Perspex containers) according to standard protocols defined by the National Electrical Manufacturers Association (NEMA), [NEMA NU-2-2001, NU-2-1994] , the International Electro technical Commission (IEC),[IEC 61675-1, 1998, IEC/TR 61948-3, 2005] and the German Standard DIN 6855-4, ["Qualitatsprufung nuclearmedizinischer Messysteme- Teil 4:Konstanzprufung von Positronen-Emissions-Tomographen PET"]. Such standards allow the performance of a PET scanner to be evaluated under well-controlled conditions and also provide a basis for comparing different PET systems' designs or different modes of acquisition on the same Positron Emission scanner.

In the section below, the most important and common performance parameters and QC tests for full ring PET systems are described.

2.2.a Spatial resolution

There is a fundamental limit to the spatial resolution that can be achieved with PET, due to the physical processes of positron annihilation Beyond this, the main factors that determine spatial resolution of current generation PET scanners are the crystal material and the size of individual detector elements. The intrinsic resolution of the detectors (which is related to detector size) contributes significantly to the spatial resolution of the system. It should be noted that intrinsic resolution is best at the center of the FOV and deteriorates toward the edge of the FOV. The PET spatial resolution can be defined in axial, transverse radial and transverse tangential directions.

2.2.b Sensitivity

Sensitivity refers to the proportion of emitted annihilation pairs that are detected as true coincidences and contribute to the final image. The higher the counting efficiency, the better the instrument makes use of the available photon flux and the lower the noise level in the images. Sensitivity is commonly expressed as volume sensitivity in units of cps/Bq/cc (or kcps/μCi/cc). Manufacturers normally use this unit as a specification for the PET scanners.

Overall sensitivity depends on the geometric efficiency, detection efficiency, pulse height analyzer (PHA) window settings, and the dead time of the system. Increasing the diameter of the ring decreases the solid angle subtended by the source at the detector, thus reducing the geometric efficiency and in turn the sensitivity. Sensitivity increases with increasing number of rings of the tomograph.

2.2.c Noise Equivalent Count Rate

In addition to the true coincidences, which carry useful image information, random and scatter coincidences are also recorded during a PET scan, The cause a loss of contrast in the image, resulting in reduced lesion detection. The concept of Noise Equivalent Count Rate (NECR) has been introduced in order to provide a single figure of merit for different relative trues, randoms and scatter contributions.

2.2.d Scatter Fraction

Scattered radiations add noise to the reconstructed image, and the contribution varies with different PET scanners. The scatter fraction (SF) is parameter that is often used to compare the performances of different PET scanners. The lower the SF value the better the performance of a scanner and better the quality of images.

2.2.e Count Rate Performance/ Count Rate Losses and Random Coincidences

Count rate performance describes the response of the imaging system when the amount of radioactivity within the field of view is increasing. To characterize the count rate behavior of a PET scanner at high activity, random events, noise equivalent count rate, and dead time loss need to be determined as a function of activity. Emission data will be corrected for randoms, dead time, and scatter using vendor-provided algorithms.

2.2.f Sensitivity to out of field of view activity

For PET cameras operated in 3-D mode, activity outside the direct axial field of view (FOV) can contribute significant counts. Where events are detected which originate from outside of the FOV, the singles, randoms and scattered events are increased (contributing to an increase in count rate) but there is no increase in the true events detection.

3. PET Quality Control Procedures

3.a Blank scan

3.a.i Rationale

The Blank scan ensures the integrity of the detector system, that is that the detectors have not drifted since the last normalization (see section D.3.c). The blank scan also serves for attenuation correction (see section E.2.a).

3. a.ii Methodology

The blank scan QC procedure involves a volume phantom used to assess the current camera performance against standard or reference QC data.

The uniform phantom filled with a positron emitter should be positioned both horizontally and vertically in the center of the field-of-view of the PET camera. Data are acquired to enable a tube-by-tube efficiency calculation by comparing the Daily QC scan to the Standard QC scan. Tube efficiencies that are outside the expected efficiency range (90%-110%) are flagged.

Alternatively, sinograms are obtained using a long lived Ge-68 or Cs-137 source mounted by brackets on the gantry which is rotated around the scan field without any object in the scanner. All detectors are uniformly exposed to radiations to produce homogeneous detector response and hence a uniform sinogram. A malfunctioning detector pair will appear as a streak in the sinogram.

Significant differences from the reference data prompt the need for re-normalization, detector set-up or replacement. Manufacturers generally include software for the acquisition and testing of the blank scan and usually a report is generated by the system, which assesses the efficiency of the detectors.

The number of counts acquired in the blank scan should be the same as for the standard or reference QC scan. Also, it is very important that the bed and phantom are placed in the same position as they were placed for the Standard QC scans during normalization.

3a.iii Acceptable performance

The daily-acquired blank sinogram is compared with the standard (reference) blank sinogram obtained during the last setup of the scanner. The difference between the two sinograms is characterized by the value of the average variance, which is a sensitive indicator of various detector problems. It is expressed by the square sum of the differences of the relative crystal efficiencies between the two scans weighted by the inverse variances of the differences. The sum divided by the total number of crystals is the average variance. If the average variance exceeds 2.5, recalibration of the PET scanner is recommended.

A chi-square calculation is performed between the Standard QC and the Daily QC volume phantom scans

- If chi-square is>2.5 but <7, a PET normalization procedure should be performed as soon as it is convenient. Patient scans can still be performed; however, the test scans' images must be checked for obvious artifacts.

A single detector block among other well-tuned blocks may not result in a chi-square >7.

- If chi-square is >7, re-normalization or service on the scanner is needed. Patient scanning is not advisable.

3a.iv Frequency

It is recommended that this quality control procedure be executed daily to verify system integrity before scanning.

3.b Transmission scan

See section E

3.c PET Normalization

3.c.i Rationale

Normalization corrects for non-uniformities in images due to variations in the gain of photomultiplier tubes (PMT), location of the detector in the block and the physical variation of the detector; that is, the normalisation procedure ensures the existence of an adequate correction of the changes in efficiency among the crystals of the detectors. The normalization correction in PET is similar to the flood field correction applied in SPECT.

The sensitivity of the line of response (LOR) relative to the mean is affected both by the geometry of the scanner and the LOR position. Additionally, the block detectors themselves vary in efficiency, as the PMT gains are not all exactly the same and may change with time; and the scintillation crystals are not all identical. In the process of normalization, the individual correction factors for each LOR are referred to as Normalization Coefficients (NCs). The most straightforward way of obtaining a full set of NCs is to perform a scan where every possible LOR is illuminated by the same coincidence source. Since NCs can change with time and should be measured as part of routine quality control, NCs are modelled as the product of intrinsic crystal efficiencies and a small number of geometric factors. Any given crystal efficiency is a factor for many NCs, and, if the geometric factors are accurately known, the number of unknowns is reduced from the number of LORs to the number of crystals.

3.c.ii Methodology

This test is carried out by using a rotating rod source of a long-lived radionuclide (normally Ge-68) mounted on the gantry parallel to the axis of the scanner or using a standard volume phantom containing a positron-emitter at the center of the scanner. The activity used in the source is usually low to avoid dead time loss.

Data are acquired in the absence of any object in the field of view. This exposes all detectors uniformly. The multiplication factor for each detector is calculated by dividing the average of counts of all detector pairs by each individual detector pair count (i.e., along the LOR). These factors are saved and later applied to the corresponding detector pairs in the acquired emission data of the patient.

3.c.iii Acceptable Performance

To have better statistical accuracy in individual detector pair counts, several hours of counting are necessary and overnight acquisition of data is often made. Fine gain calibration of all detectors in the PET system will be performed regularly in conformance with the manufacturer's recommendation, followed by recalculation of the sensitivity normalization factors for the scanner.

3.c.iv Frequency

During installation, the PET system should be normalized. Following that, the normalization should be performed monthly as part of regular maintenance, or if need is indicated by the daily quality control results. Normalization should also be performed following a service of the detectors or electronics.

3.d PET Crystals' Efficiency / Plane Efficiency

3.d.i Rationale

Crystals' Efficiency test is performed to ensure that all detectors are working properly. The setup scan changes the detector gains to modify any possible drifts.

The Plane Efficiency test compares the variations in uniformity of images between planes.

3.d.ii Methodology

A standard phantom containing a positron emitter is used. The phantom in positioned in the center of the PET scanner field of view. A Crystals Efficiency scan is acquired; the reconstructed images are checked for any non-uniformity. A bad detector indicates a decreased activity in the image, and requires the adjustment of PMT voltage and the discriminator settings of Pulse Height Analyzer (PHA).

After the Crystals' Efficiency calibration is completed, the plane efficiency scans are acquired by keeping the standard phantom/positron emitter at the center of the field. The scans are compared, and the computer, using multiplication factors to average the plane's responses, corrects inter-plane efficiency variations to average

the planes' responses and produce uniform images.

3.d.iii Acceptable performance

PMT gains must be adjusted to ensure all outputs are within 8% deviation of the target outputs

3.d.iv Frequency

This can be performed weekly; but if the daily quality control data are within the acceptable limits, the crystals efficiency calibration and plane efficiency could be performed monthly.

3.e Coincidence Timing

3.e.i Rationale

The Coincidence Timing Calibration adjusts the time delay of events from each crystal to obtain the same mean delay for all crystals in the system.

In a PET camera, each detector registering an incident photon, generates a timed pulse. There will usually be some time difference between two timing pulses arising from a coincidence event due to the finite time resolution of the detector and the discriminator system. So if a timing pulse is generated on one channel at time t, a coincidence will be recorded if there is a timing pulse on the other channel between $t - \tau$ and $t + \tau$. (τ is known as the coincidence resolving time of the system. The value of τ must be carefully chosen. If it is too small compared to the time resolution of the detection system, true coincidences will be missed. If it is too large, more random coincidences will be counted without significant increase in the number of true coincidences.)

These pulses are then combined in coincidence circuitry, and if the pulses fall within a short time-window, they are deemed to be coincident.

3.e.ii Methodology

PET manufacturers include software programs (sometimes available as an extra cost option) for the coincidence time calibration so that parameters such as coincidence timing and time window channels are definable by the user. Data acquisition required a point source, usually Na-22, placed within a scatter cylinder and laid axially in the field of view. The point source is placed on end inside of the Scatter Cylinder and centred in the field of view. Acquisition and calibration are generally under the control of software provided by thr manufacturer and may be system specific.

3.e.iii Acceptable Performance

Refer to manufacturers' information

3.e.iv Frequency

Weekly.

3.f PET Energy Shift (Discriminator Threshold)

3.f.i Rationale

In most systems operating in coincidence mode the energy window adjustment will be automatic under software control. It is important, though, to ascertain that the main energy window is correctly positioned over the 511 keV photopeak and that the energy resolution is within required limits.

PET energy shift is the average difference in energy peak location of the current reading from the baseline. This routine procedure determines the minimum energy required to trigger an event, calibrates the lower threshold window to achieve a proper ratio of coincidence to single events and trigger channel single count to total single counts. It provides stable and uniform system operation.

3.f.ii Methodology

As this calibration is sensitive to gain changes, if a transmission source is mounted on the system, it must be removed and stored away from the gantry.

A Quality Control source holder is inserted inside the field of view and a point positron source of low activity 3,7 MBq (100 μCi) is placed and adjusted in the center of the Field of View.

A vendor's software program for Discriminator Threshold calibration will be run. The system calculates which trigger channel is the most out of calibration and adjusts only that one trigger channel. The results are stored in the system.

3.f.iii Acceptable Performance

Success is achieved when the worst-case channel is only 2% out of tolerance.

3.f.iv Frequency

Weekly.

4 PET/CT System QC.

4.1 PET/CT system basics

In this system a CT scanner is combined with the PET components of a full-ring BGO, LSO or GSO scanner. This allows attenuation correction of PET images using CT data and the co-registration of functional images from PET and morphological images from CT. In a PET/CT scanner, the PET and CT tomographs are housed in a single gantry or separated contiguous gantries. The CT tomograph is usually in the front of the gantry, and the PET tomograph in the back. State-of-the-art PET/CT scanners usually have a bore size of 70 cm and an axial length of 100 cm. This large bore allows the use of immobilization devices, such as body molds and head casts, and accommodates large patients. PET/CT scanners can be used either as a dedicated PET scanner or as a dedicated CT scanner. The CT scanner is usually a multi-slice device with axial or helical acquisition modes and different rotation speeds, and the PET scans can be acquired in 2-D or 3-D mode.

4.1.a Image acquisition and processing

A PET/CT imaging protocol usually calls for acquisition of a CT scout scan first, followed by a CT scan and a PET scan. The CT scout scan serves as an anatomic reference for the PET/CT scan and is used to define the starting and ending locations of the actual CT and PET acquisitions. The CT scan is acquired over the range that is defined on the scout scan. Upon completion of the CT scan, the bed is automatically moved to position the patient in the field of view of the PET scanner. The patient is positioned so that the PET scan matches the same anatomic extent imaged during the CT acquisition.

4.1.b CT Acquisition Parameters

CT scans are usually obtained using 100–140 kVp at various amperages, depending on the imaging protocol. Other parameters which may be varied depending on the protocol include slice thickness/pitch and tube rotation speed. Subjects are generally imaged supine at suspended maximal inspiration with arms elevated over the head to minimize beam hardening artefacts.

4.2 PET/CT Quality Control Procedures

4.2.a CT Check-up (X-ray Tube Warm-Up)

A CT Check-up procedure guarantees optimum image quality. It should be performed at the start of each day, if the system has been idle for two or more hours and prior to running any QC test or patient examination.

If an error occurs during the check-up, the check-up must be performed again and if the problem persists, the service must be contacted.

4.2.b CT parameters measurements

4.2.b.i Rationale

Image quality in computed tomography depends on the technical characteristics of the radiological equipment, as well as the parameters of the examination. Pixel noise, CT number uniformity and tube voltages need to be tested.

4.2.b.ii Methodology

A water phantom or a phantom made of a uniform tissue equivalent material (standard for CT quality control tests) is usually provided by the manufacturer and used for testing the following three parameters:

A. **Pixel Noise** of images (calculated as a standard deviation).

An axial image of the phantom is acquired and pixel noise is determined from the standard deviation in CT number in a region of interest placed centrally within the image. Baseline noise values should be obtained for several scan protocols that will be used clinically using the routine QC noise phantom.

For multi-slice CT scanners, the noise of the images acquired for each single axial rotation is compared.

B. CT number Uniformity (calculated in Hounsfield Units [HU]).

Beam-hardening effects can give rise to variations in the measured linear attenuation coefficients as the spectrum of the X-ray beam shifts towards higher energies with increasing depth into the object. Corrections for this effect can be applied in software. Several manufacturers use phantom transmission measurements to calibrate the system to ensure that CT numbers for objects of equivalent composition are independent of beam hardening effects.

Each pixel is assigned a numerical value (CT number), which is the average of all the attenuation values contained within the corresponding voxel. This number

is compared to the attenuation value of water and displayed on a scale of arbitrary units named Hounsfield units (HU)**.**

CT number uniformity can be assessed at the same time as measuring noise, by placing four additional regions of interest at positions near the edge of the image of the uniform phantom. Mean CT number is then noted for the four regions as well as the central one. The deviation from the central value should be calculated.

C. **Tube Voltages** (measured directly on the x-ray tube).

These measurements should be performed for all available kV values.

The uniform phantom is placed on the bed and the bed is set at the patient examination height. The patient table is positioned so that the phantom is in the scan plane and the scan is initiated. The position of the phantom is checked and it is corrected if necessary. Two measurements are performed for each tube voltage (kV step). The difference between the first and second measurement is calculated and recorded. The measurements are repeated for each tube voltage. After each measurement, a CT image of the phantom can be displayed and stored.

4.2.b.iii Acceptable performance

If the results are outside of tolerances quoted by the manufacturer, the service engineer should be called.

4.2.b.iv Frequency

The CT quality measurements should be performed daily, before patient scanning commences.

In a monthly CT constancy QC, further checks and corrections – if this is necessary- of position of the **light marker** (z position) as well as the **modulation transfer function (MTF)** that characterizes the spatial resolution of the CT system can be performed.

4.2.c <u>Two bed Test Scan/ Field of View Offset calibration (see also E.2.e)</u>

4.2.c.i Rationale

-The Two-bed Test Scan must be performed daily in order to keep the geometrical and registration accuracy of the PET/CT fusion images.

-The Field of View (FOV) Offset calibration test will be performed as an acceptance test to calibrate any possible offset of the registration.

4.2.c.ii Methodology

A two-bed test scan involving CT and PET acquisition and reconstruction should be performed, using a PET Quality Control Uniform (20 cm cylinder) compact phantom of a positron emitter (Ge-68 source) or a refillable one (positron emitter F-18).

The phantom should be positioned, as a patient would be set up. A PET/CT protocol should be performed, defining two bed positions overlapping in the centre of the phantom. The acquired images must be inspected for non-uniformity or the presence of obvious artefact.

4.2.c.iii Acceptable Performance

If there are obvious problems, it may be necessary to re-normalize the PET, or service the CT or PET. Fused PET/CT images should be checked to ensure they are well registered; If not, it may be necessary the Offset Calibration procedure to be run again.

4.2.c.iv Frequency

The two-bed test scan should be performed on a daily basis. FOV Offset calibration procedure will be performed during installation. It should then be performed once a month as part of regular maintenance, or if the need arises following inspection of the daily two-bed test scan results. The calibration should also be repeated any time the PET gantry is moved for service access.

4.2.d Efficiency Calibrating Factor (ECF)

4.2.d.i Rationale

In a PET/CT system an overall calibration is necessary to avoid artifacts.

4.2.d.ii Methodology

A uniform, 20 cm cylinder phantom of a Ge-68/Ga-68 is positioned in the PET camera gantry. A tomogram is acquired and the center of the phantom is defined. Then a CT transmission and a PET emission scan of the phantom are acquired. To perform the calibration, the uniform phantom must be steady and the table be moved towards the center of the field of view, horizontally and vertically.

4.2.d.iii Acceptable performance

The crystals efficiency calculation should always proceed of any Efficiency Calibration Factor performance.

4.2.d.iv Frequency

Monthly

E Attenuation Correction in SPECT and PET

The attenuation of emitted radiation results from a loss of photons due to absorption and scatter in the surrounding medium. The effect of this on an image is an underestimation of the amount of radioactivity in deep-seated objects compared with those containing equivalent activity which are situated more superficially. The degree of attenuation is dependent on the energy of the photons, the linear attenuation coefficients of the materials through which the photons traverse and the thickness of materials. For technetium-99m (γ ray energy 140 keV), the attenuation factor can be 4 to 5-fold when imaging the head and up to 10-fold for abdominal imaging. For PET imaging the effects of attenuation can be even more pronounced, despite the higher energy of the annihilation photons (511 keV) since both photons must traverse the medium without interaction to be detected. Where a uniform linear attenuation coefficient can be assumed, software corrections can be applied to the data to approximate the effects of attenuation, without the need for direct measurements. However, where the attenuation varies significantly throughout the area being imaged, corrections can only be applied with knowledge of the relevant attenuation map.

1 Summary of Attenuation Correction Methods

All methods rely on the use of a transmission image, created by the transmission of photons (γ or X rays) from an external source through the object of interest. This measurement is then compared with an equivalent measurement taken without the object in place (the so-called 'blank scan'). A comparison of the two enables a determination of the attenuation effects at each point in the object. This is used to generate the attenuation map.

Transmission and emission images may be acquired simultaneously or sequentially. Simultaneous acquisition ensures perfect registration of the images but requires the use of different photon energies for transmission and emission, and/or the use of physical collimation and electronic windowing to separate the transmission and the emission data.

Gamma ray transmission sources vary according to manufacturer and may be supplied as sheet sources, fixed line sources, scanning line sources, single or multiple point sources of varying activity, according to the transmission

measurement technique employed. Transmission sources for PET cameras can be of rod or point configuration and use positron sources.

X ray attenuation correction methods utilise a CT scan, which provides a measurement of the distribution of linear attenuation coefficients in the area of interest. For a dual modality SPECT/CT or PET/CT system the X ray source may either be mounted on the same gantry as the emission imaging system or, less commonly, on a second adjacent gantry. Transmission and emission images are acquired sequentially, with the object remaining stationary on the imaging couch as it moves between the two imaging positions. It is possible to use a CT image of the object acquired at a different time (i.e. in a different imaging session) in order to correct for attenuation in an emission image. This requires the use of sophisticated image registration techniques to align the transmission and emission images and can be subject to error in patient studies due to patient positioning and physiological variations arising from the temporal separation of the data acquisition. It is not recommended for routine clinical use. Increasingly, dual-modality systems combining functional and anatomical data (SPECT/CT or PET/CT) are being employed to satisfy requirements for both attenuation correction of the emission image and image fusion. The sections below deal specifically with quality control issues associated with attenuation correction. However, the CT scanner component itself must be subject to a QC programme to ensure the tube output is within required limits and that image quality meets the required specification (see section D 4.2).

2 Required Testing and Calibration

2.a Blank scan

2.a.i Rationale

The blank scan is a reference transmission scan which is used to represent the unattenuated photon flux from the transmission source in the calculation of the attenuation map. It is important that it is similar in all respects to the transmission image taken through the object of interest, apart from the variations introduced due to the attenuation of the object itself. Any additional variance, for example, due to temporal drift, transmission source uniformity etc, will result in inaccurate attenuation coefficients.

The blank scan indicates any fault in the processes contributing to the transmission image, for example, a variation in the scanning speed of a scanning line source may result in bands of different intensities in the blank scan. Similarly, incorrect positioning of a source or incorrect source exposure can manifest itself as an error in the blank scan. The blank scan is a useful check of the overall integrity of the detector system and in PET systems it can provide the dual function of a system check (normalization) [see section D.3.c] and a basis for attenuation correction.

2.a.ii Methodology

The blank scan must be acquired under identical conditions to the object transmission scan. Refer to manufacturers' instructions for specific procedures.

2.a.iii Acceptable performance

Where possible, each new blank scan should be assessed visually to identify any unexpected non-uniformity in the transmission image. It should be compared with a reference blank scan; quantitative comparisons are often performed using manufacturer's software, with limits of acceptability being system-specific.

All unexpected findings should be investigated prior to further use of the system for attenuation correction measurements.

2.a.iv Frequency

The frequency of acquisition of a blank scan depends upon the manufacturer and the attenuation correction system used. Systems using Tc-99m as a refillable transmission source will require a blank scan to be acquired every day. A daily blank scan is also recommended as best practice for those systems which use X ray CT, in order to minimise the effects of voltage fluctuations in the supply to the X ray tube. Systems which use longer lived radioactive sources will require less frequent acquisitions of the blank scan and manufacturers' recommendations should be followed. A new blank scan should always be acquired after any adjustment to the source(s), including mechanical adjustment to source motion mechanism, and after source replacement.

2.b Transmission scan counts

2.b.i Rationale

Low transmission source strength will result in low photon flux and consequent noisy transmission images. This will result in errors in attenuation correction.

2.b.ii Methodology

As per manufacturers' instructions. It should be noted that in patient studies, noisy transmission images can arise even with adequate source strength, if the patient is very large. X-ray CT systems may allow some compensation for this through adjustment of the CT parameters. In such cases it is important to ensure that an appropriate blank scan is acquired with equivalent CT settings.

2.b.iii Acceptable performance

Most manufacturers specify lower limits of acceptability for transmission scan counts, below which radioactive sources should be replaced. An assessment of the transmission scan counts should be made from the blank scan and compared with the minimum reference value supplied by the manufacturer.

2.b.iv Frequency

This should be performed in conjunction with the blank scan measurement.

2.c Rotational effects

2.c.i Rationale

Where the acquisition of transmission images is associated with the mechanical rotation of one or more moving parts it is important to ensure that there is no angular variation in the transmission data when no object is in place. This is particularly relevant when systems use only a single blank scan acquired at a fixed angle as the reference image.

2.c.ii Methodology

Where relevant for the system configuration, blank scans should be acquired at multiple angles and data compared with the reference image.

2.c.iii Acceptable performance

General accepted limits do not exist. However, the acceptable variation between sequential blank scans in the reference position is suggested as a practical useable value.

2.c.iv Frequency.

Following acceptance testing, periodic checks of the angular dependence of reference transmission images are recommended, if there is a suspicion of mechanical instability and following adjustments to gantry rotation mechanisms.

2.d Registration (see also D.4.2.c)

2.d.i Rationale

For accurate attenuation correction, the transmission and emission images must be accurately aligned. When the image sets are acquired sequentially, it is possible for the two data sets to be out of alignment; for example, if the object needs to be translated between the emission and transmission imaging planes by an automated bed/pallet movement.

2.d.ii Methodology

Accuracy of pallet positions in the y and z directions can be determined from a visual assessment of acquired image overlays or from the measured coordinates of fixed structures in corresponding transmission and emission images of a phantom acquired at maximum resolution.

2.d.iii Acceptable performance

It is important to note also that patient movement between the emission and transmission images can affect registration accuracy. In addition, in clinical imaging where fast breath-hold CT is undertaken, there can be some mismatch between the transmission and emission images, which results from the difference in time taken to acquire the two data sets, with emission data being effectively averaged over a number of respiratory cycles. It is important that users are aware of the potential artefacts that can result.

2.d.iv Frequency

Following acceptance testing, checks of registration accuracy should be undertaken when there is concern over mechanical pallet positioning or when adjustments have been made to robotic control mechanisms or pallet support. In clinical studies a visual assessment of image alignment should always be undertaken prior to using the attenuation corrected data for diagnostic purposes.

F SPECT, PET, PET/CT Image Quality Control

1 Image Quality

For a tomographic system image to be of optimum diagnostic quality, it is essential that the equipment is performing correctly, the acquisition parameters are carefully selected and optimised according to the particular application; the acquired images are reviewed to identify any errors or inconsistencies in the data and the reconstruction parameters are appropriate and consistently applied.

1.a Acquisition Data

The review of the acquired data should include the following:

i. Cine review of the projection images to identify:

Motion, data truncation, sources of high activity outside the area of interest which may lead to reconstruction artefacts, data inconsistencies (eg blank projections)

ii. Sinogram review, as an adjunct to cine review

iii. Review of transmission images where attenuation correction is to be applied, to identify: Data inconsistencies, data quality (particularly important when imaging large subjects where transmission images may be very noisy and result in poor quality attenuation corrected images), sources of potential attenuation artefact, misalignment of emission and transmission data.

1.b Reconstruction parameters.

Reconstruction parameters should be optimised during protocol development, clearly documented and consistently applied. It is important that these are checked following any upgrades or modifications to reconstruction software to ensure consistency. [Refer to procedure-specific guidelines for further information]

1.c Display formats

Display formats for processed data should also be consistent for a given application, to enable ease of comparison and recognition of abnormal features. The choice of display scale (eg colour/grey scale, continuous/banded, linear/logarithmic) should be appropriate for the application and display maxima and minima selected to ensure features in the objects(s) of interest are clearly

delineated. The photographic process is critical in assuring optimal image quality and, where hard copy is used, a daily check of the settings of the formatter and film processor is necessary. Films should be selected carefully for superior density response characteristics. Where images are reported from soft copy it is important that the viewing monitor is correctly calibrated and that images are viewed using the recommended ambient lighting.

1.d Contrast & background

Relative fluctuations of counts between adjacent areas in the image of an object determine contrast. Statistical variations of the count rates, decrease information density or count density and give rise to noise. A minimum number of counts are needed for a reasonable image contrast. Even with adequate spatial resolution of the scanner, lack of sufficient counts may give rise to poor contrast due to increased noise; so lesions may be missed. Image contrast to delineate a lesion depends on its size relative to system resolution and its surrounding background. The effect of lesion size depends on the background activity surrounding it (maximum scatter radiation that increases background, arises from the patient) and on whether it is a "cold" or "hot" lesion.

1.e Patient Motion

Patient motion during imaging reduces the image quality and may introduce blurring of the image. This primarily results from the overlapping of normal and abnormal areas due to movement of the organ. It is helpful if the patient is kept in a comfortable position. Breath holding (where possible) or respiratory gating may improve the thoracic images. Artefacts due to heart motion can be reduced by using a cardiac gating technique.

2 PET/CT Imaging Artefacts

An artefact is a feature or appearance that is seen on an image, which does not actually exist. It may occur in any imaging modality and is sometimes unavoidable. Recognizing the presence of artefacts is important in order to avoid confusion with pathology.

2.a Attenuation correction artefact

Attenuation correction artefacts usually arise as a result of overcorrection due to very dense objects in field, e.g., metallic implants, pacemakers, and oral and IV contrast. This can be misinterpreted as increased activity and can can be avoided by inspection of the corresponding non-attenuation corrected images.

2.a.i Oral Contrast Artefact

Normally a lower-density barium oral contrast is used in CT test for a PET/CT examination. At common concentrations, the diluted oral contrast is not dense enough to produce attenuation over-correction artefacts. However with time, the contrast can become compacted in the bowel with a significant increase in the density and foci of increased uptake can be seen. A CT density less than 400-500 Hounsfield units should not produce any artefact. Recent studies using dense oral contrast agents for GI studies (barium meal or enema) can lead to artefacts following attenuation correction.

2.b Motion artefact

Misalignment of PET and CT images can occur at the diaphragm due to respiratory motion, cardiac motion, muscle relaxation and any other movement of internal organs above and below the diaphragm. During the CT acquisition the patient is in deep inspiration and the organs around the diaphragm would be in different position from that in the PET scan. Because the PET scan is acquired over several minutes with the patient breathing normally, the structures at the diaphragm undergo slight blurring and the contour of the diaphragm is in the neutral position. Motion artefact can result in inaccurate fusion of CT and PET data. The liver dome and spleen may be seen “floating” above the diaphragm.

Methods used to avoid motion artefact include establishing consistent breathing instructions and respiratory gating techniques.

2.c CT Truncation

The CT scanner has a relatively narrow axial field of view of only 50 cm and often the arms, shoulders, and hips lie outside the visualized area. PET emission images are acquired over many minutes (25-40 min) and patients are often unable to keep their arms above their head for the duration of the scan. Artefacts are produced then as the CT attenuation correction reconstruction algorithm does not account for any attenuation of the CT x-rays by the tissues outside the field of view. Truncation artefacts appear as dark lines extending cranio-caudally along the patient.

G Documentation, Record Keeping and Action Thresholds

The aim of a QC programme for nuclear medicine instrumentation is to maintain performance at an optimal level and to identify any deterioration in equipment performance, which may be sudden or gradual, which could have a deleterious effect on clinical image quality and hence clinical outcomes. To do this, quality control procedures must be undertaken with sufficient frequency, must be accurate and reproducible and must be subject to rigorous record keeping.

It is essential that QC procedures are carried out in a consistent manner (for example, same collimator, orientation, activity, energy window width etc) and that all procedures are documented with sufficient detail to enable any trained operator to undertake QC to the same standard and without introducing any additional variance into the data. All acquisition parameters should be recorded, dated and retained with the test results to enable meaningful comparisons to be made over time. Good record keeping is paramount for the detection of gradual deterioration of performance over an extended period of time, which requires the analysis of temporal variations in measurement parameters to identify trends.

A baseline set of QC results should be recorded after installation and acceptance testing, to serve as a reference for the life of the instrument. A new baseline may need to be established if the equipment is subject to major modification or upgrade (for example, replacement of a detector). Details and dates of equipment servicing, repair and any upgrades should be documented and cross-referenced to QC results.

Quality control procedures are only effective if the results are linked to outcomes. It is necessary to determine, in advance, the appropriate course of action following a QC test. This requires action thresholds to be established. A QC result which exceeds an action threshold may trigger one of a number of subsequent actions, for example, repeating the test, undertaking of an additional more comprehensive test, arranging a service visit, taking the equipment out of clinical service until the problem is rectified. Action thresholds will be dependent on the particular equipment, the baseline equipment performance at acceptance testing and the expected normal variance of the measurement parameters. It is usual for quality control procedures to be carried out with increased frequency for the first few months following installation so that the latter can be established.

It is important that consideration is given to the impact of any equipment performance deterioration on patient outcomes. A gamma camera non-uniformity in the peripheral part of the field of view, for example, might render the equipment unusable for bone imaging and SPECT, whilst still being satisfactory for planar gastric emptying studies where only the central field of view is required. Whilst it is always necessary for equipment faults to be rectified as soon as practicable to ensure optimal performance, each circumstance will need to be considered individually. Practitioners advised to consult with an experienced nuclear medicine physicist for advice.

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IV. Acknowledgments

The members of the EANM working group on NM Instrumentation QC, acknowledge the following persons (in alphabetical order) Dr Fängewisch G.L. from University Clinic of Giessen and Marburg, Germany, Mr. Gogos C. from University of Athens, Greece, Mr. Harris A. from Sheffield Teaching Hospitals, UK, Mr. Laspas I. from University of Athens, Greece and Dr Preiß M. from University Clinic of Giessen and Marburg, Germany, for their valuable assistance in performing some QC tests, reading the guidelines drafts and providing constructive comments .

To whom it may concern

As Taskgroup and Committee Coordinator of the European Association of Nuclear Medicine (EANM), I would like to express the cordial thanks on behalf of the EANM to Prof. Lyra Georgosopoulou for having chaired the Quality Assurance Working Group.

The Quality Assurance Guidelines Working Group was established by Prof. Manfred Fischer in 2005 with the mission of preparing Quality Control Guidelines for Nuclear Medicine instrumentation.

Within the two year drafting process, Prof. Lyra Georgosopoulou closely cooperated with the other two members of the Quality Assurance Working Group, Prof. Wendy B. Tindale (UK) and Prof. Rigobert Klett (Germany).

In her role as Quality Assurance Working Group Chair, Prof. Lyra Georgosopoulou represented the EANM at the IAEA International Conference on Quality Assurance and New Techniques in Radiation Medicine (QANTRM) in Vienna, Austria, 13-15 November 2006. Her report on the EANM contribution to that conference was subsequently published on the EANM Website: https://www.eanm.org/news/news_detail.php?newsId=63&navId=291&referrer=/news/news_archive.php

In December 2007, the comprehensive guidelines draft submitted by the Quality Assurance Working Group has been accepted by the EANM Executive Committee as basic EANM document on the issue of Quality Assurance. Upon completion of its mission, the Working Group was dissolved in accordance with the EANM Instructions for Taskgroups and Committees.

Since then the EANM Physics Committee has taken the lead in the continuing discussion on Quality Control in Nuclear Medicine Instrumentation. The expertise gathered within the former EANM Quality Assurance Working Group remained safeguarded for the EANM by the kind offer of the Working Group Members to provide continuous advice.

In great appreciation of the committed work accomplished by the EANM Quality Assurance Working Group, we convey our best wishes for the future endeavours undertaken by Prof. Lyra Georgosopoulou.

With best regards.

Prof. Dr. Hatice Durak EANM Taskgroup and
Committee Coordinator

Vienna, March 18, 2009

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