COMMISSIONING AND ROUTINE TESTING OF FULL FIELD DIGITAL MAMMOGRAPHY SYSTEMS

2025

Contents

Revisions to the previous unpublished report

This report supports the quality assurance activities of the breast screening programme carried out by the Screening Quality Assurance Services and incorporates some important updates such as changes to remedial dose levels and National Diagnostic Reference Levels (NDRLs). This report updates version 4 (never officially published) to incorporate urgent updates of tests.

The revision to the previous guidance include the following

Removals:

- most of the introductory material on technology and digital imaging
- Computed Radiography from the text: technology and testing
- references to technology and testing of CR and CRT
- details of x-ray models
- geometric distortion
- testing of printers
- description of manual reading of CDMAM
- TOR(MAS)/TOR(MAX) test
- noise separation (until further evidence)
- DQE tests

Revisions:

- Updated remedial dose levels under AEC
- Update of testing of reporting monitors
- Adapted the dark noise test
- Addition of calliper test
- Update of AEC repeatability
- The frequency of some tests is reduced: image retention, x-ray field to breast support edge distance
- Recommend to aim for achievable level for CDMAM results
- Change in National DRL to 2.5 mGy
- Recommend use of MTF in preference to SWCTF

1 INTRODUCTION

The use of digital imaging in mammographic imaging is now well established. Indeed, the technology has been extended to cover new modalities, e.g. digital breast tomosynthesis (DBT) and contrast enhanced imaging. There is separate NHS Breast Screening Programmes (NHSBSP) guidance on "Routine quality control tests for breast tomosynthesis". Contrast enhanced imaging is supported for use in breast screening assessment, where available. Whilst there is no NHSBSP specific testing guidance, there are publications that a protocol can be based upon. See "A protocol for quality control testing for contrast-enhanced dual energy mammography systems" and "Technical evaluation of TiCEM contrast enhanced mammography on the Siemens Revelation system"

This document recommends suitable test protocols for commissioning and routine performance testing for full field digital mammography systems. It should be used in conjunction with the current edition of "Institute of Physics and Engineering in Medicine (IPEM) Report 89", which fully describes the testing of the mammography x-ray unit. Note that some tests may be different or may have to be adapted, such as those for the automatic exposure control (AEC) system.

The limiting values (remedial levels and suspension levels) given in this document are based on the current experience of testing full field digital systems. Further evidence and updated guidance from other professional bodies will mean that these tolerance may need to be revised.

The remedial level is a level of performance at which some form of action needs to be initiated. The suspension level is a level of performance at which it is recommended that the equipment should be removed from clinical use until the performance is corrected (a fuller discussion of remedial and suspension levels is given in IPEM Report 91 and European Guidelines for Quality Assurance in Breast Cancer Screening and Diagnosis When deciding on action, the effects on clinical outcomes should be taken into consideration.

Terms in *italics* are explained in the Glossary.

1.1 Target audience

This guidance is aimed at providers of medical physics services for the NHSBSP. This sets out the minimum standards for commissioning and routine performance of full field digital mammography systems.

1.2 Detector and environmental temperature control

A digital detector can be sensitive to environmental changes (in terms of both temperature range and rate of change of temperature) and the recommended range of operating temperature can be comparatively narrow (see supplier's product data or the National Co-ordinating Centre for the Physics of Mammography (NCCPM) report 2002). Certain types of detector may be irreparably damaged if subjected to a temperature that is either too high or too low. Thus, an air conditioning/heating system capable of maintaining the ambient room temperatures at the required levels and at all times is an important consideration in digital installations.

Certain designs of detector may need an integrated cooling system, primarily in order to reduce *dark current* noise or to maintain the detector at a certain temperature. The temperature control system may take the form of a heat exchange mechanism with a circulating cooling fluid, provision of a fan or fans in the detector enclosure or some form of electronic temperature control.

1.3 Detector dose indicator and breast dose indication

It is important that digital imaging devices provide a dose index to give an indication of the exposure received by the detector. Without such an index it is possible that doses may drift from the optimum. The wide dynamic range of a digital imaging system will allow images produced by a wide range of detector exposures to be displayed with a similar greyscale appearance. Consequently, such drifts may not be readily detected. The manufacturer of the system should state the relationship of dose index indication to detector entrance exposure (along with the calibration conditions). Digital systems provide a direct indication of breast dose based on the exposure parameters. The method of calculation should be specified. For specific information, users need to refer to manufacturer's specifications.

1.4 Quality assurance (QA) procedures

This is a quality control (QC) protocol that sits within the QA framework of the NHSBSP. In particular, there is "Guidance for medical physics services".

In addition to the tests and measurements described in this document, mammography systems may be provided with a supplier's built-in quality control (QC) facility. This may be automatic on switching the system on and will provide a pass/fail indication. In addition, the user may be required to carry out certain calibration procedures prior to use on a routine basis (daily, weekly) such as *flat-fielding* or imaging of a test object supplied with the system. There is also "Guidance on routine user QC testing for full field digital mammography systems" carried out by breast screening services.

2 TESTING METHODOLOGY

This section explains the methodology behind the testing protocols outlined in section 3 of this guidance.

In assessing the performance of digital imaging systems, it seems natural to separately assess characteristics related to the performance of the detector and the display system. We have adopted this approach for many of the tests. However, some parameters require that the overall performance of the system is assessed. One such parameter is image quality since this depends on the performance of the image generation stage, the selection of radiographic factors by the AEC system, the performance of the detector, the image processing and the image display.

Detector tests should use '*for processing'* image data which have minimal or no display processing applied. Tests of display devices largely involve the use of synthetic images with well defined content (display test patterns).

To fully evaluate the performance of digital mammography systems in accordance with the recommendations of this protocol, it is necessary to have access to a means of undertaking quantitative measurements. This may be achieved either by having the relevant tools available on a review workstation or a means of exporting images in a *DICOM* format for remote analysis, for example, to a removable hard-drive.

2.1 The reference plane

The detector *pixel size* is defined at a reference plane parallel to the detector. In full field digital mammography, it could be the detector plane or an arbitrary plane above the breast support platform. The manufacturer may or may not specify the reference plane. The reference plane may be different for magnification imaging.

Distance measuring tools use the *pixel size* to calculate distances on the image. They are readily available with most digital systems and convenient to use. However, in order to make use of these tools, the accuracy of distance measurements in the reference plane(s) must be established on all the devices that measurement might be made on. When no reference plane is defined, the plane in which measurements are accurate must be determined.

2.2 Detector uniformity and artefacts

Non-uniformities may arise within the detector system due to spatial variations in the sensitivity of the *x-ray converter* and readout device. Integrated digital detectors can correct for these inherent non-uniformities by a process of *flat-fielding*. This relies on the non-uniformities being spatially consistent between images. The flat-fielding procedure also compensates for non-

uniformities in the x-ray beam due to the anode heel effect and x-ray beam divergence. Flat-field correction maps are obtained using a standard beam attenuator for a range of exposure conditions (e.g. different target/filter combinations and focal spot sizes). Image uniformity is assured for the calibration conditions. However, deviations from these, e.g. by adopting different spectra or beam attenuator thicknesses, will result in non-uniformities as the distribution of x-ray flux emerging from the attenuator may not be completely compensated for by the correction map. Therefore, this report suggests a standard method for testing all systems rather than using each manufacturer's method. The method sets a baseline measurement for future tests.

The pixelated readout arrays in flat panel integrated detectors will usually have some defective or "dead" *pixels* which are unresponsive to the signal generated in the *x-ray converter*. *Flat-fielding* will not compensate for these and their presence will cause signal dropout in the image. These artefacts can be compensated for by firstly identifying the defective *pixels* and then interpolating new *pixel values* at that location using surrounding *pixel values*. To view the defective *pixel* map it is necessary to have access to the *raw image data*. Manufacturers should be able to provide a specification as to what level of defective *pixels* is acceptable for the detector.

2.3 Detector response

The exposure range over which the detector response is linear may be specified by the manufacturer. At acceptance it is necessary to confirm this aspect of the detector's response against the performance specification, if available. In any case, it is useful to establish that the range is greater than the dynamic range in the x-ray signal emerging from the breast. The testing methodology described in this document uses 45 mm thick polymethylmethacrylate (PMMA) blocks or 2 mm thick aluminium sheet. The attenuator is placed proximal to the x-ray tube to reduce the amount of scatter reaching the detector; placing the attenuator close to the detector may be more realistic but would produce inconsistent results as the level of scatter varies rapidly over short distances from the attenuator. Measurements at commissioning are best carried out with the grid removed, so that a measurement of the air kerma incident on the detector can be made. For routine measurements the grid may be left in and a *grid transmission factor* applied to the air kerma readings.

For a linear x-ray detector whose performance is x-ray quantum limited the relationship between exposure to the detector and image *pixel* variance (square of standard deviation) should be a linear function. Most detector systems may exhibit quantum limited performance over only a limited range of exposures. This may be identified by deviations from the above-mentioned linear relationship. This may occur due to the presence of electronic noise or structure noise in the images. This is discussed further in "An alternative method for noise analysis using pixel variance as part of quality control procedures on digital mammography systems".

2.4 Detector resolution

The resolution can be characterised for the purposes of quality control by a measure known as the *limiting spatial resolution*. This is the highest frequency bar and space grouping that can be resolved on an image of a high contrast resolution test grating. As such, it represents an upper limit to the resolving capacity of the imaging system and represents the point at which the modulation transfer function (MTF) of the system falls to a low value (usually < 5%). In fact, for normal quality control, with the test grating positioned approximately 40 mm above the breast

platform, it is the resolution not just of the detector that is determined, but of the entire imaging chain.

The resolution of a digital imaging system detector will mainly depend on three factors.

- detector *pixel size*
- *pixel* aperture
- inherent unsharpness of the *x-ray converter* material

The resolution limit of the digital detector can be characterised as being either *pixel* limited or *x-ray converter* limited. The theoretical limiting resolution of a digital detector is given by the *Nyquist frequency*. This is determined by the *pixel pitch*, which is the sampling interval. A *pixel* limited system is one in which the resolving capacity of the system is limited by the sampling interval of the readout/digitisation stage. In this case, the detector/conversion stage has resolution capabilities that exceed that of the *Nyquist frequency*; however, due to the limited sampling frequency, these cannot be properly represented in the digital image. In this case the MTF of the detector/conversion stage extends beyond the *Nyquist frequency*. This results in *aliasing* of the higher frequency components of the signal to frequencies below the *Nyquist frequency*. In such cases, the signal is said to be under-sampled.

Measurement of the spatial resolution of such systems with a bar pattern grating should always result in a *limiting spatial resolution* which corresponds to the system *Nyquist frequency* (depending on the modulation of the signal, *aliasing* should be visible in groupings which have frequencies above the *Nyquist frequency*). Pure confirmation of the *Nyquist frequency* in this way provides limited information as it does not confirm the modulation present at the *Nyquist frequency*. Note that if the measurement is performed at 45° to the *pixel* matrix axes, the effective *pixel* pitch is smaller than the *pixel pitch* by a factor of $\sqrt{2}$ and the *Nyquist frequency* will therefore be greater by this factor.

In the previous reports, the square wave contrast transfer factor (SWCTF) was used as analogous to MTF. In this protocol MTF is the main recommended method, with SWCTF as an alternative.

2.5 Automatic exposure control

The control of exposure for a digital imaging system is important to retain the optimal image quality and breast dose for a wide range of compressed breast thicknesses and densities. The operation of AECs have become more sophisticated since the publication of the earlier versions of the report. Different manufacturers have different methods for selecting the optimal exposures, often with a pre-pulse exposure and the system will base the exposure on the densest region.

2.6 Display systems

In the digital imaging environment, it is essential to obtain consistent display of the medical image. To this end the American College of Radiologists (ACR) and National Electrical Manufacturers Association (NEMA) developed the Grayscale Standard Display Function (*DICOM* 3.14). This *DICOM* standard ensures that a medical image displayed on *DICOM* calibrated imaging devices will have a consistent greyscale appearance regardless of the specification of the device, as long as the viewing conditions are adequate. It is recognised that primary display systems used for diagnosis should be *DICOM* calibrated and matched in performance. Furthermore, it may be

considered desirable for secondary display systems used for manipulating the acquired image to be *DICOM* calibrated as well.

Primary display systems are considered to be those on which diagnostic decisions are made, they are generally found in reporting rooms and have a high specification for image display. Secondary display systems are those where images can be viewed either to confirm positioning or to review an image in conjunction with a radiological report. They are generally of lower specification and can include the display integrated with the x-ray equipment. The distinction between these two classes of display may be clear cut in some areas but there some areas where the distinction between diagnostic and review may be blurred, where 'medical management' decisions are made on monitors not regarded as diagnostic displays. It is suggested that in these areas a risk based review is made regarding the types of decision being made from the image and the criticality of the monitor's performance to that decision. This can then guide an appropriate level of QC.

Following the installation of a full field digital mammography system, it is therefore necessary to:

- ensure that the environmental conditions are suitable for the viewing of medical images
- evaluate the performance of all the display devices against the manufacturer's specification
- confirm that primary devices are *DICOM* calibrated and matched in terms of minimum and maximum luminance.

Routine quality control is essential to monitor the performance of the display devices. Evidence suggests that the display devices can be the weak link in the imaging chain, and hence it is imperative not to overlook them. This is discussed in "The impact of technical and environmental conditions on the quality assessment in mammography" and "How does the display luminance level affect detectability of breast microcalcifications and spiculated lesions in digital breast tomosynthesis images".

Since the previous draft of this report comprehensive acceptance and routine tests have been developed for display systems in "Display Quality Assurance: The report of the American Association of Physics in Medicine (AAPM)" In reviewing the present document it was decided that some of the more important recommendations of this report be adopted within the NHSBSP and updated assessment criteria are detailed later in this report. Of specific note is a change in the requirements of the ambient light levels. There is some evidence that if the ambient environment is too low it can cause eye fatigue and also cause the user's visual system to enter into mesopic vision in which low-contrast objects can no longer be distinguished. We have adopted a pragmatic approach to monitor testing. We have selected tests from the TG270 document that we believe will easily and reliably demonstrate monitor performance without excessive time or equipment resources. Environmental light levels, luminance ratio, *DICOM* calibration, luminance uniformity and monitor resolution and distortion can be assessed using the TG18 test patterns and a light meter. Overall imaging performance can be evaluated visually using the TG18-MM image, which contains some microcalcifications, or a clinical reference image selected by the users showing similar subtle detail.

2.7 Image quality - detail detection

Threshold contrast tests are a common means of assessing image quality for noise limited imaging systems. Test objects have been designed which provide details (usually circular) covering a suitable range of diameters, each with varying thicknesses of contrasting material. The visibility of a signal within an image depends on the contrast presented by that signal and the

level of background noise. The level of contrast presented depends on the radiological path length of the detail and the contrast of the display system. For smaller sized details, the contrast will also depend on the amount of unsharpness in the imaging system. The level of noise will usually depend on the x-ray quantum statistics and is related to the level of x-ray exposure to the detector and the efficiency of the detector. A limitation to being able to detect the image detail is related to the CNR. The approach taken in this protocol is to set a minimum standard for details which should be visible, defined in terms of object thickness. The use of harder x-ray spectra reduces the contrast of the target. However, the target will still be visible if the required CNR is maintained by having relatively low noise. Target values of contrast of the gold discs in the CDMAM phantom using standard beam conditions for anode/filter combinations in use, alongside a description of the methodology for their calculation, can be found in

The x-ray attenuation coefficient of materials is x-ray energy dependent; hence, the contrast between different materials and/or different thickness of material depends on the x-ray spectra used. Simulations have indicated that the relative change in contrast with energy of gold and aluminium details (materials which are commonly used in test objects and are used in this protocol) are similar (within 5 to 10%) to that of glandular tissue/calcifications over the range of x-ray spectra which may be used clinically (K C Young & B Johnson, personal communication).

The standards expected for the threshold thickness of contrasting material at different detail sizes have been derived from the European protocol for the quality control of the physical and technical aspects of mammography screening.[6] They have been designed to ensure that digital systems have a detail detection performance that is at least as good as the majority of film-screen systems. Since these tolerances were set the technology has improved as discussed in "Historical trends in image quality and mean glandular dose in digital mammography", and the EUREF guidance recommends that systems operate as far as possible at a standard equal to or better than the achievable level. In the European protocol, the image quality measurements and the limiting values apply to unprocessed images. Most systems apply some additional image processing to clinical images before display. As these processing algorithms are specifically designed for clinical images rather than contrast-detail test objects, it was thought that these should not be used.

The images of the CDMAM phantom should be automatically read rather than scored visually. Artinis supply software called cdcom to analyse individual images which is available on the euref website. NCCPM supply a program (CDMAM analysis) for processing a stack of images with cdcom and applying the method as demonstrated in "Evaluation of software for reading images of the CDMAM test object to assess digital mammography system" to calculate the threshold gold thickness for the different diameters. There are other options available for calculating the threshold gold thickness.

The EUREF guidelines set out a method to equate the measured CNR with the threshold gold thickness of the CDMAM phantom. This is then extended to set out the estimated acceptable CNR for different thicknesses of PMMA. The methodology is set out in APPENDIX 6.

2.8 Mean Glandular Dose

The methods of measuring dose are the same as those described in IPEM Report 89. Where blocks of PMMA are used, the dose calculated is the MGD to an equivalent breast as described in "Additional factors for the estimation of mean glandular breast dose using the UK mammography dosimetry protocol" by Dance et al Where measurements are made from exposures to real breasts, the composition is assumed to vary with thickness in the manner found to be typical by the paper by Dance et al. The limiting values for MGD for a 53 mm thick standard

breast model (measured using 45 mm of PMMA) and other thicknesses were derived from the European protocol. Periodic audits of clinical exposures should be carried out with reference to local and national diagnostic reference levels (DRLs) as outlined in "Guidance for the implementation of the IR(ME)R Regulations 2017".

It should be noted that a new breast dose model has been published by the Joint AAPM Task Group 282/EFOMP on Breast dosimetry for standard and contrast-enhanced mammography and breast tomosynthesis. This is an international collaboration intended to standardise breast dose calculations. It has been shown that average breast glandularity is significantly lower than the 50% assumed in the Dance model as discussed in "The myth of the 50-50 breast". They have also used breast computed tomography (CT) to determine a more realistic distribution of glandular tissue within the breast as well as a more realistic breast shape. This is discussed in "Patient-derived heterogeneous breast phantoms for advanced dosimetry in mammography and tomosynthesis".

The approach is then quite similar to the Boone model published in "Glandular Breast Dose for Monoenergetic and high energy X-ray beams: Monte Carlo Assessment" where Monte Carlo simulations are performed with mono-energetic x-ray beams. The results for each energy can be combined with weightings reflecting the proportion of each energy in the incident clinical beam. In this way dose estimates can be made for novel beam spectra with no need for additional Monte Carlo simulations. At the time of this report, the AAPM TG232 approach has not been accepted for use in the NHSBSP.

2.9 Quantitative measurements

The use of quantitative measurements in routine QC has been recommended for many years as discussed in "Quality control measurements for digital x-ray detectors". In this protocol we recommend the traditional limiting resolution and pseudo MTF method of square wave transfer function is replaced by the Modulation Transfer Function. This is a method that produces consistent results but with more information about the system. The test is described in section 3.2.4, with further details in APPENDIX 5.

The use of Noise Power Spectra has not been included in this protocol, however, it is likely to be included in future editions. The use of NPS provides a sensitive test for examining noise and the images will already be acquired from the detector response function (section 3.2.2). More information can be found in APPENDIX 5.

3 TEST PROTOCOLS

The tests outlined in this report are those which specifically address the performance of the digital imaging components of a digital mammography system or those tests whose performance or results are affected by the fact that the image is acquired in a digital format. The recommended tests, their frequency and limiting values are summarised in APPENDIX 1. The performance of the following listed tests should be undertaken following the protocols outlined in the latest edition of IPEM Report 89:

- x-ray tube leakage
- tube voltage accuracy
- radiation output

- focal spot dimensions
- half value layer
- guard timer
- compression force.

When performing x-ray tube and generator tests, any integrated detector should be protected from direct x-ray exposure (e.g. use a lead sheet to cover the whole of the detector area).

When analysing images numerically, a standard sized Region of Interest (ROI) should be used. This should be small enough to avoid errors caused by non-uniformity (for example, see Impact of heel effect and ROI size on the determinations of contrast-to-noise ratio for digital mammography systems) and will be typically 5 mm x 5 mm. When performing tests users should be aware that the view selected (laterality), e.g. LCC, RCC, will affect the orientation of the presented image. It is preferable to select a consistent view.

3.1 Beam alignment

The alignment tests required for a full field digital mammography system are as follows:

- alignment of the light field to the x-ray field
- alignment of the x-ray field to the imaged field/detector
- size of the imaged field
- separation between the chest wall edge of the visible field and the chest wall edge of the breast support platform
- Electronic caliper calibration

For all tests ensure that the appropriate collimation is selected. In the case of dual track x-ray tubes, the tests need to be repeated for each of the different target materials (Mo, Rh or W) at commissioning, and routinely only if they are used clinically.

3.1.1 Alignment of the x-ray field to the light field and to the visible image field/detector

Test protocol

Test procedure given in IPEM 89 should be followed. If the unit has a laterally shifting paddle, the alignment should be checked in all positions. Use Gafchromic film or fluorescent screens as available to acquire images.

Evaluation

Evaluation should be as described in IPEM 89. However, users must be aware that the image dimensions may be affected by electronic shuttering on some systems.

Alignment of light field to x-ray field

Remedial level: misalignment >5 mm along any edge.

Alignment of x-ray field to imaged field/detector

Remedial level: > 5 mm or < 0 mm overlap of image by x-ray field on any side.

Suspension level: > 10 mm overlap or > 2 mm unexposed border along chest wall edge with respect to the image

Frequency: All targets, field sizes and paddle positions to be tested at commissioning and a sample of common clinical settings every six months.

3.1.2 Size of imaged field

Test protocol

Use images acquired in **Error! Reference source not found.**.

Evaluation

A direct indication of the imaged field size is given by the test object's scaled markings and compared to the specified dimensions. Alternatively, the electronic measuring tool can be used to measure the visible field size. A geometric correction may be needed to transform the measurements to the reference plane defined by the manufacturer or the detector plane.

Remedial level: Ratio of measured to specified dimension < 0.95

Frequency: commissioning only.

3.1.3 Separation between the image edge and the chest wall edge of the breast support platform

Test protocol

Test procedure given in IPEM 89 should be followed. Acquire an image of the test object marking the front edge of the breast support platform and view it on the display monitor.

Evaluation

The image of the test object will give a direct indication of the separation between the image and the chest wall edge of the breast support platform. Alternatively, the electronic measuring tool can be used to determine the offset. A geometric correction may be needed to transform the measurements to the reference plane defined by the manufacturer or the detector plane.

Remedial level: > 5 mm between edge of the image and front edge of the breast platform.

Frequency: Commissioning only

3.1.4 Electronic calliper calibration and reference plane

This protocol can be used to establish the accuracy of distance measurements in the reference plane.

Test protocol

For this test it will be necessary to verify with the manufacturer the position of the reference plane and its distance from the focal spot (d_1). If this is not known, the test can alternatively be used to determine the plane at which the calliper is accurate. Position an object of known dimensions (k) on the breast support platform. Acquire an image (use low values of kV and mAs). Measure the distance between the focus and the breast support platform (d_2). View the image on the display monitor. Repeat for all magnification settings.

This test should be undertaken on any display system where measurements may be made that affect clinical decision making. This can include the acquisition workstation, PACS or mini-PACS. The measurement just needs to be undertaken on one reporting system per PACS.

Measure the dimension (m) of the test object on the image using the electronic measuring tool.

Evaluation
The percentage error (a) of the measuring tool is calculated as follows:

$$a = \left(\left(\frac{md_2}{kd_1} \right) - 1 \right) \times 100 \quad (1)$$

Where m is the measured dimension of the test object using the electronic callipers, k is true dimension of the test object, d_1 is the distance from the focal spot to the manufacturer's reference plane, and d_2 is the distance from the focal spot to the breast support platform.

Remedial level: error >2%.

Frequency: commissioning and after software changes that might change the measurement tool.

3.2 Detector performance

The following measurements aim to evaluate certain performance characteristics of the imaging detector. The dose measurement for tests such as detector response should be performed at a *standard position* in the x-ray field. The recommended position is on the midline of the detector at 40 mm from the chest wall edge.

3.2.1 Artefacts and uniformity

It is important to have a reproducible setup each time the test is performed.

Test protocol

Before starting, ensure the breast support and PMMA is clean and free from specks of dust.

Place a 40 or 45 mm thick PMMA phantom on the breast support table (if it is large enough to cover the detector) or at the tube port and expose under AEC or at a typical mAs value. The compression paddle (covering full area) and grid should be in place as for clinical use. The PMMA phantom, paddle and breast support should be clear of dust and dirt. Repeat for the different target/filter combinations used clinically.

Alternatively, follow the manufacturer's protocol if available.

Acquire '*for processing' image data*.

Evaluation of artefacts

View the images on the reporting workstation using 1:1 image pixel to display pixel mapping and a high contrast presentation. This may require readjustment of display window level and width. Set a narrow window width to highlight any subtle differences in signal. Record the window width and level settings for use on subsequent occasions.

Inspect the image for artefacts such as “dead” pixels (white or black pixels), black-and-white pixel pairs, structured pixel clusters or lines, and images or ghost images of foreign objects (e.g. specks of dust), which may appear blurred and indistinct if the object is close to the tube exit port. To rule out artefacts from the test object then it can be rotated and re-imaged. If artefacts are noted, rotate or pan images; if artefact does not move with image, then it is on the monitor rather than the test object or detector. Image based artefacts will move as the image is moved with respect to the display system. Display system artefacts will keep a fixed position and orientation relative to the monitor. Record the details of the artefacts observed.

If there are significant detector artefacts, the flat fielding procedure should be carried out.

Record the number of “dead *pixels”* and their position and formations (e.g. lines, clusters).

Remedial level:

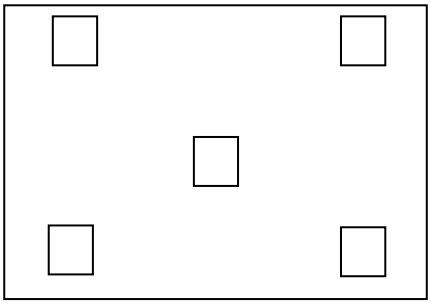
- Any dead *pixel* dropout – a recalibration of the detector is required
- other artefacts that may affect clinical image quality visible.

Frequency: commissioning and every six months.

Evaluation of uniformity

Uniformity can be evaluated in many ways. The suggested methods below are quick to carry out and provide uniformity measures for constancy purposes. Uniformity measures can also be obtained using the same exposure conditions and analysis techniques as the manufacturer; only the latter uniformity measures can be compared directly with the manufacturer’s specification.

Measure the mean pixel value for an ROI at a position in the centre and at each of the four corners of the *‘for processing*’ image as shown in *Figure 1*. The location of the ROIs can affect the results, so their positions should be recorded so that the same setup may be used on subsequent routine visits. Linearise the mean Pixel Values (as determined in section 3.2.2) before calculating the uniformity.



Chest wall edge

Figure 1. Nominal ROI locations for measuring uniformity

Calculate the percentage deviation of the corner means from the central ROI mean value using linearised *pixel* values:

$$\text{Unifomity} = \frac{\max\left|\text{Centre ROI mean} - \text{Corner ROI mean}\right|}{\text{Centre ROI mean}} \times 100 \qquad (2)$$

Remedial level: 5 percentage points above baseline uniformity.

Frequency: commissioning and every six months.

3.2.2 Detector response function

The following test should be carried out where possible with the anti-scatter grid removed to enable accurate estimation of the air kerma at the detector input plane. Removal of the grid on some systems may result in the detector being vulnerable to mechanical damage. Care is necessary to prevent items falling directly on the detector. It is recommended in such cases that a *grid transmission factor* is measured at the commissioning stage and this factor is then used in subsequent detector response measurements (see APPENDIX 3).

Test protocol

The protocol is best carried out in three stages: 1) Decide on acquisition parameters 2) Measure air kerma per mAs; 3) Acquire images across mAs range and plot detector response .

1) Choose a standard uniform attenuator (e.g. 2 mm thick Al, no paddle or 45 mm thick PMMA with or without paddle). It is important to consistently use the same beam load as at acceptance. If using aluminium as an attenuator care should be taken to ensure that it has been manufactured in such a way as to not add significant non-uniformities into the image. A study: "Measurement of the detective quantum efficiency in digital detectors consistent with the IEC 62220-1 standard: practical considerations regarding the choice of filter material", has shown that the use of some types of ultra-pure aluminium can influence QC measurements. At commissioning, select a typical beam quality (kV/target/filter) applicable to 45 mm PMMA (e.g. as selected by AEC), and use the same beam quality every time the test is repeated on the unit.

2) A measurement of x-ray tube output is made with the standard uniform attenuator placed at the x-ray tube port. This part of the test can be done at the end of the tube and generator tests. The detector should be protected from direct exposure by covering it with a sheet of highly attenuating material (e.g. lead or steel). Place the dosemeter at the *standard position* on the breast support table and remove the compression paddle. Use full field collimation (e.g. 18 cm x 24 cm or 24 cm x 30 cm). Set three mAs values (e.g. 5, 50 and 160 mAs) and record the chamber readings. The readings should be adjusted if necessary, using the inverse square law and *grid transmission factor* to give the air kerma at the detector entrance plane. No corrections are made for attenuation in the breast support and detector cover.

   Plot entrance air kerma versus mAs and apply a linear fit to the data. The coefficients of this fit can be used to calculate detector entrance air kerma for any mAs setting. For the second stage, remove the dosemeter but leave the attenuator at the tube exit port.

3) Acquire images for calculating detector response function.

Zero dose image: set a low kV and mAs and acquire a '*for processing*' image without any exposure to the detector by covering the detector with a sufficiently thick metal plate, e.g. lead or stainless steel.

Remove the protective sheet from the detector. Using the linear equation from stage one, calculate the mAs values needed for a range of detector air kerma values (e.g. 12.5, 25, 50, 100, 200, 400 and 800 µGy). Remove the grid, set the calculated mAs values (closest mAs station on the system) and acquire uniformly exposed '*for processing'* images, over the detector's air kerma range.

Evaluation

Examine the zero dose image with a narrow window width and look for artefacts.

Measure the mean *pixel value* and standard deviation of the mean using a standard size ROI placed at the *standard position* on each digital image.

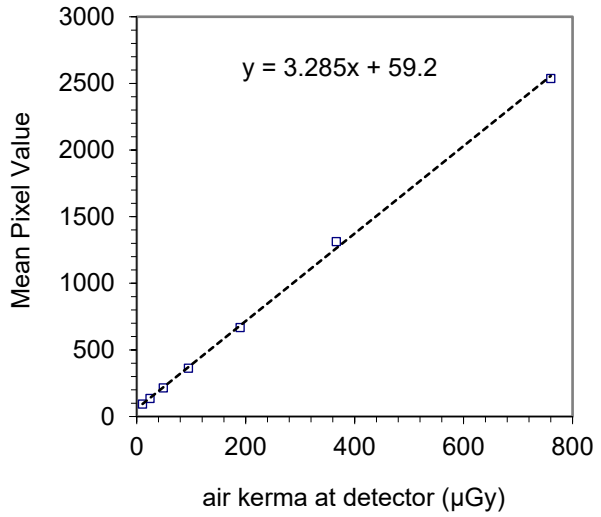


Figure 2. Pixel value versus detector entrance air kerma

To obtain the detector response, plot mean *pixel value* against detector entrance air kerma as shown in Figure 2. Fit a trend line of the form $y=ax+b$ or $y=a\log(x)+b$ or $y=ax^b+c$, as appropriate for the system tested, and record the *a, b (*and *c)* constants. The detector response is used to quantify detector gain but also can be used to linearise and normalise images as well as estimate detector entrance air kerma for a given *pixel value*. At commissioning choose a clinically representative target *pixel value* (PV_{clin}) (applicable to 45 mm PMMA as selected by AEC). Use the detector response to determine the detector entrance air kerma required to produce the target *pixel value*. This air kerma level is the "detector reference air kerma" (AK_{ref}).

Remedial level:

- detector reference air kerma > 20% change from commissioning value
- Artefacts seen in zero dose image.

Frequency: commissioning and every six months.

3.2.3 Noise Analysis

Evaluation

Use the images acquired in section 3.2.2 to analyse the noise. Correct the measured PV_{clin} and σ_{clin} to PV_{lin} and σ_{lin}, use the processes set out in APPENDIX 2. Plot standard deviation against detector entrance air kerma using log-log axes. (For this graph omit the point corresponding to zero air kerma.) Fit a power trend line of the form $y=dx^e$ to points in the proximity of PV_{lin} and record *d* and *e* (for example as in Figure 3). For a quantum limited detector, the expected relationship is $y=dx^{0.5}$. The presence of certain noise sources in the system other than quantum noise (e.g. fixed pattern noise, electronic preamplifier noise) will cause the response to deviate from a straight line at low and high air kerma values. Use the fitted line to determine the pixel standard deviation at the detector reference air kerma (σ_{ref}). Use this to calculate the *signal to noise ratio* (SNR) at the detector reference air kerma:

$$SNR_{ref} = {(PV_{lin} - b)}/{\sigma_{ref}} \quad (3)$$

These measurements at commissioning will serve as the baseline for subsequent noise measurements. For routine tests, compare each standard deviation to the baseline value at the corresponding detector entrance air kerma, if necessary using the fitted function to make corrections for differences in air kerma at survey visits.

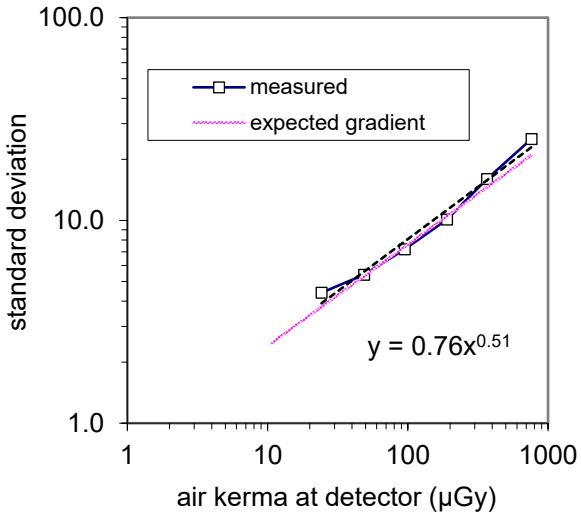


Figure 3. Standard deviation versus detector entrance air kerma.

Remedial level:

- Standard deviation (linearised) at any entrance air kerma > 10% change from baseline at same air kerma
- SNR_{ref} change > 10% compared to baseline

Frequency: commissioning and every six months.

3.2.4 Modulation Transfer Function

The MTF describes the spatial frequency response of a linear, spatially invariant imaging system. IEC 62220-1-2 suggests the use of the slanted edge method; this is a robust technique that is also suitable for routine QC. A stainless steel sheet of thickness about 0.8 mm, with straight edges, can be used. The following describes the use of a steel sheet with two of the orthogonal edge being used, if there is only one edge then separate images need to be acquired for the horizontal and vertical directions.

The following test is a quantitative measurement of the detector's resolution properties and as such offers reproducible, objective estimates and is sensitive to changes in detector performance over time.

Test protocol

Set the same beam quality and standard attenuator as used to acquire the detector response images (section 3.2.2), and a mAs setting to produce an entrance air kerma at the detector that is approximately 3.2 times the detector reference air kerma (AK_{ref}) value as set out in 3.2.2 (there must be no saturation in the edge image). Place the test object on the breast support table, with a vertical edge along the midline of the image receptor with an approximate angle of 1° to 3° with respect to pixel matrix, and the horizontal edge closest to the chest wall about 40 mm from the chest wall edge. This positioning allows the MTF to be evaluated in orthogonal directions simultaneously in the proximity of where the detector entrance air kerma has been measured. This is acceptable for routine QC measurements. Acquire three images, repositioning the test object between them.

In a full evaluation, the centre of the edge is positioned 40 mm from the chest wall edge along the midline, with the edge orthogonal (1° to 3° angle) to the direction of interest and an image is acquired. This is repeated for the four directions and an MTF is calculated for each acquisition (left-right direction (low to high signal change for the edge spread function (ESF) and high to low signal change for the ESF); similarly for the chest wall-nipple direction).

Evaluation

Import the image into the chosen analysis software (APPENDIX 5), ensuring that the image pixel pitch is correctly read from the DICOM header information. If not, correct it manually. Linearize the image *pixel value* data before calculating the MTF. Extract a sufficiently large region containing the edge such that glare (low frequency signal spread) within the detector is characterized; the actual ROI dimension will depend on the characteristic distance of the glare, however an ROI of at least 50 mm x 50 mm should be used. Note the conditioning applied when obtaining the MTF result (smoothing, windowing, extrapolation of the line spread function (LSF) tails *etc*.).

Record the spatial frequencies at which the MTF reaches 50% and 10% (left-right and chest wall-nipple directions).

Remedial level: measured MTF_{50} & MTF_{10} > ± 10% change in spatial frequency from commissioning values.

Frequency: commissioning and every six months

3.2.5 Detector resolution

The following tests (3.2.5.1 and 3.2.5.2) can be omitted if the detector resolution is tested by the MTF method (3.2.4).

3.2.5.1 Square wave contrast transfer factor

The following tests require the availability of a resolution test grating which contains groups of line pair patterns. Each pattern should have at least 4.5 cycles. The available frequencies of the patterns should range from approximately 1 cycle per mm up to a frequency which exceeds the *Nyquist frequency* of the detector under examination. As an illustration, a detector with a 50 μm pixel size will have a *Nyquist frequency* of 10 cycles per mm while a detector with a 100 μm pixel size will have a *Nyquist frequency* of 5 cycles per mm.

Test protocol

Place the resolution test grating as close as possible to the detector, generally on top of the breast platform. The test pattern bars should be orientated at a small angle (<10°) to each axis of the pixel matrix. Acquire 'for processing' images. Obtain the images with manual exposure factors of 26 kV (a low kV setting maintains a high subject contrast between bars and spaces in the test pattern) and approximately 15 mAs (a suitable mAs should be chosen to ensure that the signals arising from the bars and spaces in the test pattern are within the dynamic range of the imaging system). The same set of radiographic parameters should be used in subsequent resolution tests.

Evaluation

First, establish the normalizing factor. To do this, use an ROI to measure the mean *pixel value* in a region corresponding to the attenuating level of the test piece, M_B (i.e. corresponding to a bar). This is done by placing the ROI on the lead border of the resolution test grating. Next, measure the *pixel value* relating to the lowest attenuating region in the resolution grating, M_S (i.e. that corresponding to a space). This is done by placing the ROI on the background region of the image (away from the test grating). Next, locate the line pair group closest to 1 lp/mm. Measure the standard deviation for this group ($M(f)$) using an ROI that just covers the bars and spaces for this group – the ROI must not include the background region between the different line pair groups.

Figure 4 shows an example of the location of the ROIs and for three different frequencies of f_1, f_2 and f_3.

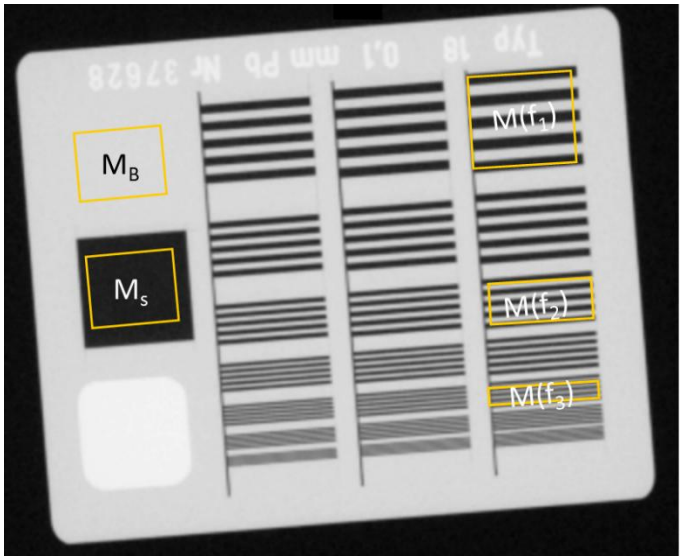


Figure 4. Example image for the SWCTF calculation for three frequencies

The object amplitude, M_0, is given by the modulus of the difference between M_B and M_S:

$$M_0 = \left| M_S - M_B \right| \quad (4)$$

The transfer factor at a given frequency is given by:

$$\mathrm{SWCTF}(f) = \frac{M(f)}{M_0} \quad (5)$$

Repeat with the grouping most closely corresponding to 4 lp/mm and at 80% of the *Nyquist frequency* of the detector.

Remedial level: measured SWCTF(*f*) > 10% change from commissioning values.

Frequency: commissioning and every six months.

3.2.5.2 Limiting spatial resolution

For systems where the resolution is converter limited it may be useful to also determine the *limiting spatial resolution* as a constancy check.

Test protocol

Place the resolution test grating as close as possible to the detector, generally on top of the breast platform. The test pattern bars should be orientated at 45° to the principal axes. Obtain the images with manual exposure factors at a low tube potential (e.g. 26 kV) and approximately 15 mAs.

Evaluation

Use appropriate display magnification, windowing and viewing distance. Evaluate the number of groups where the bars and spaces are seen. The correct number of bars and spaces in a group should be resolved in a direction perpendicular to the bar direction.

Remedial level:

- commissioning: the *limiting spatial resolution* fails to meet the manufacturer's specification (where given) or is < 70% of the *Nyquist frequency* of the detector
- routine: detector *limiting spatial resolution* < 75% of the commissioning value.

Frequency: commissioning and every six months.

3.2.6 Spatial discontinuity and resolution homogeneity

Test protocol

Place an extremely fine radio-opaque mesh (matched to pixel pitch) on the breast support table. Expose manually using low exposure parameters (e.g. 28kV, 10mAs). Acquire 'for processing' images. An alternative method can be undertaken using a *variance image* of a large flat field image, as discussed in "Early experience in the use of quantitative image quality measurements for the quality assurance of full field digital mammography x-ray systems". This has the advantage of not requiring extra images or equipment.

Evaluation

Inspect the image for discontinuities such as line artefacts due to data interpolation that can be attributed only to the detection process (e.g. it moves if the image is panned). Examine the image for regions of blurring.

Remedial level: any evidence of discontinuities or regions of blurring.

Frequency: commissioning and every six months.

3.2.7 Image retention

Test protocol

Place a 45 mm thick PMMA phantom on the breast support table and acquire two images using manual exposure factors similar to those used under clinical conditions (e.g. as described in section 3.3.2 for 45 mm thick PMMA). For the first image, the PMMA is positioned in such a way that one half of the detector is covered and the other half is not. For the second image, place a 0.1 mm thickness of Al sheet on top of the PMMA (exactly centred). This time the PMMA covers the whole of the detector. The time between both images should be approximately one minute.

Acquire for processing images. Linearise the data as described in APPENDIX 2.

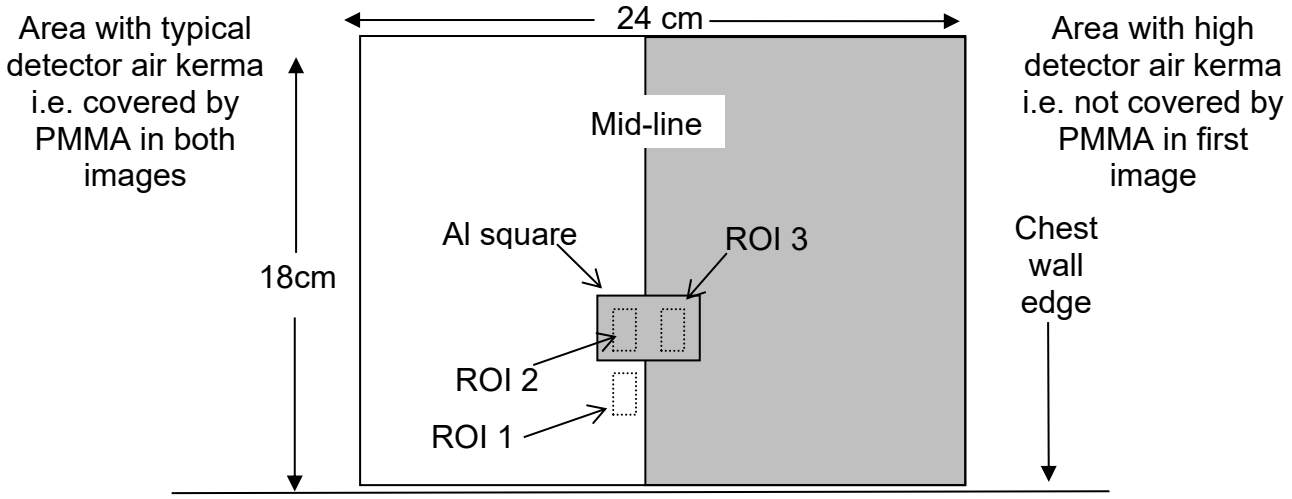


Figure 5. ROI locations for image retention measurement.

Evaluation

Measure the mean pixel value in the ROI on the locations shown in Figure 5 and calculate the image retention factor.

$$\text{Image retention factor} = \frac{(\text{ROI 3 mean} - \text{ROI 2 mean})}{(\text{ROI 1 mean} - \text{ROI 2 mean})} \quad (6)$$

Remedial level: image retention factor > 0.3.

Frequency: commissioning.

3.3 Automatic exposure control

In general, full field digital systems have an automatic exposure control (AEC) in order to select the appropriate tube voltage and target/filter combination as well as to control the duration of each exposure. The choices made by the AEC depend on the thickness and composition of the compressed breast and may involve a pre-exposure pulse to measure the transmission through the breast. The decision-making process can be quite complex but on sophisticated systems the detection of a local dense area will result in an increase in the overall exposure. The relatively simple tests in this section using uniform blocks of PMMA provide a guide to the performance of the AEC and a check on whether any systematic changes have occurred since the previous tests.

In this test, one paddle size and type may be used. In reality, different paddle sizes and types (e.g. flex, fixed) are used clinically, these may affect the outcomes of this test. Typically, the 18 cm x 24 cm paddle is used for the test, but a range of paddle sizes can be used for the different thicknesses e.g. small paddle for 20 mm thick PMMA.

Since the tests described here employ uniform blocks the exposures to real breasts may be greater than expected due to the heterogeneous nature of breasts. This effect can be investigated by comparing a dose survey for real breasts to the dose estimations for standard breasts using blocks of PMMA.

The following tests require a PMMA block, which may be composed of several PMMA plates covering a total thickness range from 20 to 70 mm. The area of the PMMA block should be at least 100 cm² or large enough to cover the whole of the detector's dominant area. Place the PMMA block on the tabletop so that the front edge is slightly overlapping the chest-wall edge of the detector (e.g. by 5 mm) and ensure that the block is centred left-to-right in the image field.

'*For processing*' images should be acquired (note that using the '*for presentation*' will invalidate any quantitative measurements on the images). It is recommended to linearise the data as described in APPENDIX 2.

If moveable the AEC detector should be positioned at the chest wall on the mid-line if possible. All tables and modes e.g. magnification that are used clinically should be tested.

3.3.1 AEC repeatability

This is a test to ensure the exposures are consistent.

Evaluation

Make at least 4 exposure of the 45 mm thick PMMA under AEC. Record the mAs for each exposure. The AEC repeatability can also be undertaken in conjunction with the imaging of the CDMAM phantom, if acquired under AEC.

Remedial action

Remedial: Maximum deviation of mAs from mean >5%

Suspension maximum deviation of mAs from mean >10%

Frequency: Every 6 months

3.3.2 AEC performance - automatic mode

Test protocol

Select one of the automatic modes. Place 20 mm of PMMA in the beam. Place a 0.2 mm thickness of aluminium sheet (≥99.9% purity) of dimension 10 x 10 mm² or 20 x 20 mm² under or on top of the PMMA, ideally this should be in between the two 10mm thick blocks, in line with EUREF guidelines[6] and to provide consistency for the national key performance indicators (KPI) database. The aluminium should be placed as shown in Figure 6. If the detector's dominant region is adjustable then select a region that excludes the aluminium. On some systems this is not possible, in which case the influence of the presence of the aluminium on the AEC should be explored and appropriate corrections made. Compress to a standard force (e.g. 50 to 100 N) required to achieve the specified breast equivalent thickness (with spacers in place e.g. 53 mm indicated for 45 mm thick PMMA, Table 1). Expose and record the selected exposure parameters (e.g. filter, target, kV, delivered mAs, displayed CBT and displayed mean glandular dose). Repeat adding additional thicknesses of PMMA (20 to 70 mm), keeping the aluminium in the same position. The use of 80 mm thick PMMA is optional, but it should be noted that the average compressed breast thickness has increased since this test was introduced, as discussed in "Radiation doses in the UK breast screening programmes 2016-2019". It should be noted that since PMMA is generally denser than breast tissue any automatic selection of kV, target or filter may be slightly different from real breasts. This can be corrected by adding appropriate spacers to the PMMA to make up a total thickness equal to the equivalent breast (Table 1).

Table 1. Equivalent breast thickness to PMMA thickness*

PMMA thickness (mm)	Equivalent breast thickness (mm)	Glandularity of equivalent breast
20	21	97
30	32	67
40	45	41
45	53	29
50	60	20
60	75	9
70	90	4
80	103	3

*From "Additional factors for the estimation of mean glandular dose using the United Kingdom, European and IAEA breast dosimetry protocols". *Physics in Medicine and Biology.* 2009;54(14):4361-4372.

These exposures can also be used to measure the MGD as described in section 2.8.

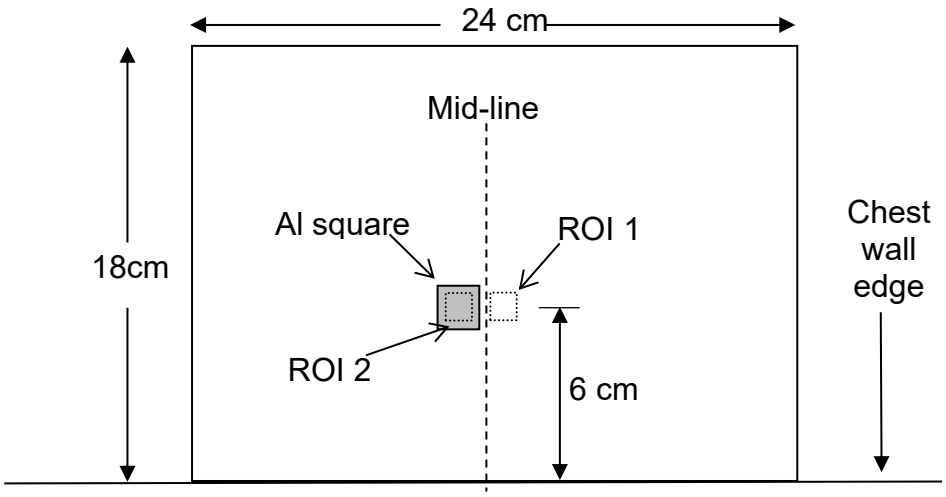


Figure 6. ROI locations for CNR measurement.

Evaluation

Measure the average pixel value and standard deviation in ROI 1 (m_1 and σ_1 respectively) on the '*for processing'* image and in the area of the aluminium square in ROI 2 (m_2 and σ_2 respectively) as shown in Figure 6. Before ROI measurements are used to calculate CNR the data should be linearised. The procedure for linearising the data is described in APPENDIX 2. Calculate CNR as

$$CNR = \frac{|m_1 - m_2|}{\sqrt{\frac{\sigma_1^2 + \sigma_2^2}{2}}} \quad (7)$$

Remedial level

Measured CNR < 90% of CNR measured at baseline

Frequency

Commissioning and every six months. Repeat for fine focus at commissioning and annually if this mode is used clinically.

3.3.3 AEC variation with density control setting

This test is only applicable to units whose density control setting is adjustable.

Test protocol

Set target and filter to those selected by the AEC corresponding to 45 mm PMMA in the beam. Vary the density control setting by ± 50% from the centre mAs value.

Record the mAs.

Evaluation

The mammography unit's density adjustment should result in a constant change (e.g. 10-15%) in mAs per step.

Remedial level: AEC density control step outside manufacturer's specification.

Frequency: commissioning.

3.3.4 AEC variation with AEC region within detector

This test is only applicable to units where the position of the AEC dominant region is adjustable.

Test protocol

Set the AEC on standard density setting and set clinically representative exposure factors with 45 mm PMMA in the beam. Vary the location of the AEC region within the detector and make exposures.

Record the delivered mAs.

Evaluation

Calculate the maximum variation in mAs from the chest wall position mAs.

Remedial level

Variation in mAs > 10%

Frequency: Commissioning

3.4 Image presentation

3.4.1 Monitors

Verify with the manufacturer that the monitors are correctly calibrated to conform to the *DICOM* Greyscale Standard Display Function (GSDF). This should be done before the soft copy display tests.

In order to test the performance of the viewing monitors, the TG18-QC (Figure 7), TG18-LN12 and TG18-UN (or UNL) test patterns must be available in the image archive. The test patterns are available as 1k and 2k resolution versions; the correct version, matching the monitor's resolution, must be used for the monitor to be evaluated. The patterns must be displayed at full resolution (one display *pixel* for each *pixel* in the digital image). Use the archive query/retrieve function to load the appropriate test pattern onto the local hard disk. Also load TG18-MM (Figure 8) onto the local hard disk.

A suitable photometer with a narrow acceptance angle (< 5°) should be used for the luminance measurements. The photometer must have a valid calibration certificate to ensure accuracy of measurements, especially at low luminance values. Do not use the photometer that might be provided with some workstations unless it has a valid calibration certificate.

Contact photometers or telescopic photometers may be used, however contact photometers are not suitable for making direct measurements of ambient luminance. See "Display quality assurance: The report of the American Association of Physics in Medicine (AAPM)" Where contact dosimeters are used, ambient luminance (L_{amb}, cd/m^2) can be assessed by making measurements of ambient illuminance (E, lux) and multiplying by the Diffuse Reflection Coefficient (R_d) appropriate for the display being tested. If this is not available, an R_d value of 0.005 cd/m^2/lux

may be assumed. See "Ambient illumination revisited: A new adaptation-based approach for optimizing medical imaging reading environments"

$$L_{amb} = E \times R_d \quad (8)$$

All monitors should be tested at commissioning. Tests on primary monitors (reporting workstations) should be carried out at the specified routine testing frequency. It may not be possible or necessary to carry out tests on secondary monitors at this frequency.

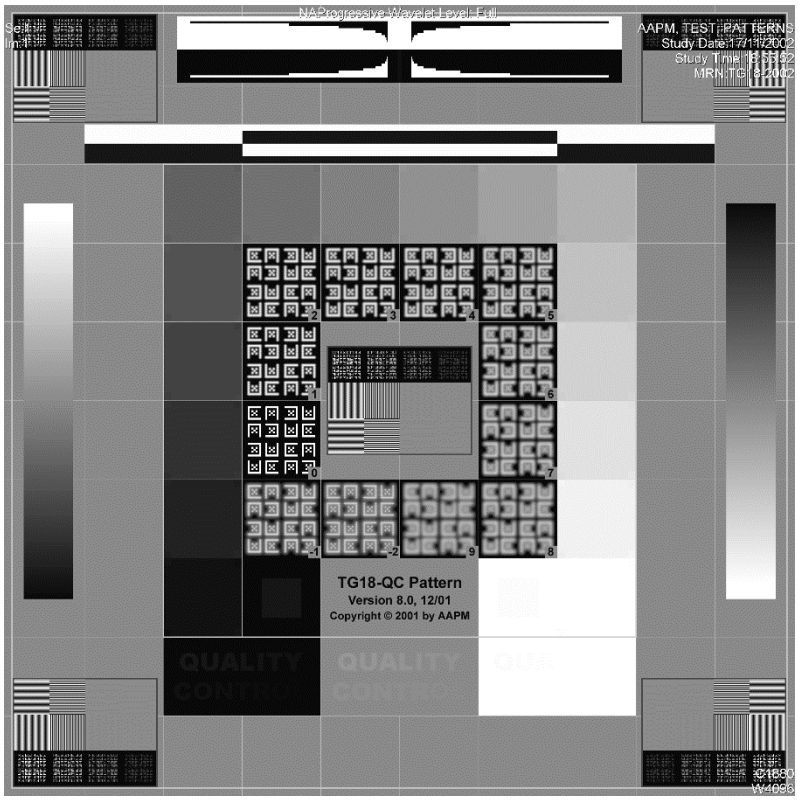


Figure 7. TG18-QC test pattern.

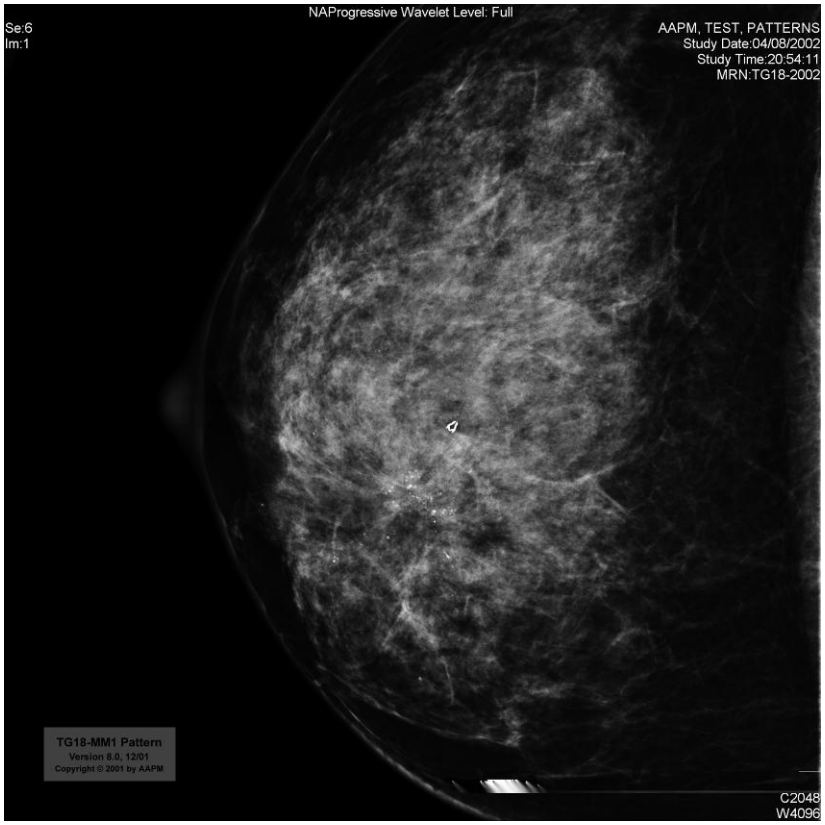


Figure 8. TG18-MM anatomical image.

3.4.1.1 Viewing conditions

Test protocol

Adjust the ambient light level to that used clinically. Switch the display off and measure the illuminance (lux) at the centre of the display with the appropriate detector facing outwards. Then measure the ambient luminance L_{amb} (cd/m^2) at about 30 cm away from the face of the monitor at the centre of the display with the appropriate detector facing the monitors (inwards). If using a contact photometer, determine L_{amb} as described in 3.4.1. Record the readings. Display the TG18-MM images or a pair of mammograms from the local database and examine the images for any reflections, e.g. of room lights, windows, self.

Remedial level:

- $L_{amb} > 0.25 * L_{min}$, where L_{min} is the luminance for the minimum pixel value
- Illuminance greater than 75 lux or less than 25 lux
- any disturbing reflections visible

Frequency: commissioning and annually

3.4.1.2 Luminance response

Test protocol

Display test pattern (TG18-LN12 or TG18-QC) using the default window settings. Measure the luminance of all the greyscale steps by placing the lightmeter in contact with the monitor in the centre of the square. If not using TG18-LN12 and if possible, the image should be panned to bring each greyscale step to the centre of the image to ensure that non-uniformity of luminance across the display device does not affect the measurements. If the light meter probe is larger than the

greyscale squares, the image can be zoomed. Inspect the small contrast steps (5% in 0% level, 95% in 100% level).

Evaluation

Add the ambient luminance determined in section 3.4.1.1 to all the above measurements.

Calculate the luminance ratio: 100% greyscale to 0% greyscale.

Using the protocol spreadsheet (see APPENDIX 4), select the appropriate test pattern and enter the luminance values to obtain the contrast response and compare against the *DICOM* standard.

It may not be practicable to carry out these tests on secondary workstations. It is however recommended that the maximum luminance of acquisition workstation displays is tested at commissioning and annually due to its use in displaying mammograms to the operator deciding whether a repeat exposure is necessary. If a suitable test pattern is unavailable for this test, display any image and window it such that the screen is displayed at its maximum brightness.

Remedial level:

The small contrast steps not visible

Primary monitor	
minimum luminance (L_{min})	<1.2 cd/m^2
maximum luminance (L_{max})	< 350 cd/m^2 (recommended value of 420) ± 5% between paired monitors
luminance ratio	< 250 or >450 (recommended value of 350.)
	contrast response luminance outside DICOM standard ± 10%
Secondary monitor	
minimum luminance (L_{min})	< 0.8 cd/m^2
maximum luminance (L_{max})	< 200 cd/m^2 (recommended value of 250.)
	luminance ratio < 250 or >450 (recommended value of 350)
	contrast response luminance outside *DICOM* standard ± 20% (if DICOM calibrated)

Frequency

- primary monitor: commissioning and annually
- Maximum luminance of acquisition workstation: commissioning and annually
- secondary monitor: commissioning

3.4.1.3 Luminance uniformity

Test protocol

Display the test pattern (TG18-UN or TG18-UNL) using the default window settings. If these are unavailable, display the TG18-LN12 or TG18-QC test pattern. Select the 100% greyscale step, zoom and pan until it covers the whole of the display on the monitor. Adjust the window level setting until the whole image is at maximum luminance. Measure the luminance at the centre of the image and at the four corners. Add L_{amb} to each measurement.

Evaluation

Calculate the percentage difference in luminance (L_{diff}) between the areas of maximum (L_{max}) and minimum (L_{min}) uniformity by using the following equation.

$$L_{diff} = 200 \times \frac{L_{max} - L_{min}}{L_{max} + L_{min}} \quad (9)$$

Remedial level: maximum percentage difference in luminance > 15%

Suspension level: maximum percentage difference in luminance > 30%

Frequency: commissioning and annually

3.4.1.4 Monitor resolution

Test protocol

Display the TG18-QC test pattern using the default window settings. Inspect the resolution gratings (and the *Cx* pattern on TG18-QC) using an optical magnifying glass if necessary. All line groups should be resolved. Inspect all text in the image. It should be sharp and clear.

Remedial level: any loss in resolution.

Frequency: commissioning and annually.

3.4.1.5 Display artefacts

Test protocol

Display the TG18-QC test pattern using the default window settings. Inspect the black-to-white and white-to-black step transitions for smearing and overshoot artefacts. Also inspect the image for flicker and dead pixels.

Remedial level: any disturbing artefacts visible

Frequency: commissioning and annually

3.4.2 Fidelity of transferred images

When images are sent from the acquisition workstation to DICOM 3.14 compliant display devices such as reporting (primary) workstations and the image archive, look-up tables can be applied to pixel values at the acquisition workstation as part of the DICOM send protocol, or at the display device as part of the display configuration. It is imperative that the appearance of the image (i.e. the pixel value to grey scale) is maintained irrespective of the display device. It is possible to assure the correct configuration of all the devices by checking the systems configurations with the x-ray manufacturer and the vendors of the PACS and display devices. If this is not possible, problems can be identified by DICOM sending a reference image from each acquisition workstation to every DICOM 3.14 display device, and between the latter.

Test protocol

The test protocol below describes how to assure the correct configuration of all display devices by sending a reference image from each acquisition workstation to every DICOM 3.14 display device, and between the latter.

Test protocol

From the mammography acquisition workstation, DICOM send an image of the TOR(MAM) test object (e.g. the image acquired in section 3.5.2 above) to every DICOM 3.14 compliant reporting (primary) workstation that the mammography unit can send to. For every one of these display devices, visually inspect the images and score the test object.

Remedial level: the image quality must be the same on all displays.

Frequency: commissioning of mammography system and any critical changes to the PACS or reporting workstations.

Note this test is optional in the routine testing of FFDM and may be carried out as part of the "Guidance on routine user QC testing for full field digital mammography systems".

3.5 Image quality

Image quality measurements must be made to establish a baseline on commissioning new equipment and whenever there are major changes in the system.

Testing of image quality is also part of the routine quality control procedures by the operators of x-ray sets. The image should be compared with any quantitative information and with previous images and data. Any deterioration in image quality will necessitate further investigation.

3.5.1 Threshold contrast – CDMAM

The procedure described here for measuring threshold contrast uses the test object CDMAM (version 3.4) available from Artinis. This has the advantage that it is the same procedure set out in "Evaluation of software for reading images of the CDMAM test object to assess digital mammography systems" and was used to develop the standards in the European protocol. The development of alternative and possibly simpler test objects that fulfil a similar task is to be encouraged. However, allowances would need to be made for differences in the design and method of scoring which could affect the measured threshold contrasts.

Test protocol

Detail visibility should be determined using the CDMAM test object. This is a contrast-detail phantom with circular details with diameters in the range 0.1 to 2 mm. The test object should be used with 20 mm thick plates of PMMA above and below. This has a physical thickness of approximately 45 mm. However, the total attenuation of this combination is approximately equivalent to a 50 mm thickness of PMMA, which in turn is equivalent to a typical breast thickness of 60 mm. Fully automatic AEC should be used but verify that the same beam quality is used as that for 50 mm PMMA, for example by using a radio-transparent spacer. If not, set the beam quality manually and use the AEC or manual mAs. Repeat the exposures until a minimum of 8 'for processing' images have been recorded. Consult the supplier for the appropriate algorithms for obtaining unprocessed images of test objects. It is desirable to vary the acquisition conditions slightly by moving the test object by a few millimetres between exposures.

Automatic image scoring software ('CDMAM Analysis' provided by NCCPM) should be used to score images.

Evaluation

The threshold gold thicknesses should be averaged for all the images assessed and plotted against the detail diameter and the data fitted with a curve of the form:

$$T_c = a + bx^{-1} + cx^{-2} + dx^{-3} \quad (10)$$

where T_c is threshold gold thickness (μm), x is detail diameter (mm) and a, b, c and d are coefficients adjusted to achieve a least squares fit. Note that it is important to ensure that all the data points are equally weighted. This can be done by fitting to a log-log plot or by selecting a relative weighting i.e. one that minimises the relative distances from the data points rather than absolute values.

The threshold gold thicknesses determined at each diameter using the fitted curve should be compared with the limiting values given below. The detail detection standards defined in Table 2 are designed to ensure that digital mammography systems perform at least as well as film-screen systems. They have been derived from measurements on film-screen and digital mammography systems using the CDMAM contrast detail phantom version 3.4. However, it is intended that they are sufficiently flexible as to allow testing by other designs and makes of test object. The values quoted form a smooth curve and may be interpolated for other detail diameters.

The above procedure requires the threshold gold thickness to be verified for at least four detail diameters covering a range 0.1 to 2 mm. However, precise measurements are best made by determining the threshold gold thickness for a larger number of detail diameters (e.g. all the detail diameters in the CDMAM test object) and fitting a smooth curve as described above. The fitted curve should then be used to determine threshold gold thickness at the specific detail diameters given in the table below. This procedure helps to reduce the effect of image noise and random observer errors, noticeable when just a few details are assessed.

Where a system appears to fail, the raw output of CDCOM should be inspected in case the change is due to a single image. The raw CDMAM images should also be inspected to check for artefacts that could be affecting the results.

Remedial level: Table 2 shows minimum acceptable and achievable levels published in the European protocol. The minimum acceptable level is considered to be the remedial level. However a system should not continue to be used if it cannot be adjusted to meet this level.

EUREF guidance states that we should be aiming for the achievable level. Users should be aware of the performance of their CDMAM phantoms when using the criteria.

It is important to record the dose used for these measurements. The equivalent MGD can be calculated using the output from the mAs given for each exposure with the methodology set out in section 3.6.1.

Frequency: commissioning and every 6 months.

Table 2. Minimum acceptable and achievable levels of detail detection.

	Threshold gold thickness (µm)	
Diameter of detail (mm)	Minimum acceptable value	Achievable value*
2	0.069	0.038
1	0.091	0.056
0.5	0.150	0.103
0.25	0.352	0.244
0.1	1.68	1.10

**The achievable value or better should be expected for a modern DR system*

3.5.2 TOR(MAM)

The TOR(MAM) test object should be used to check the imaging chain for the clinical 'for presentation' images.

Test protocol

The TORMAM test object should be placed on top of 30 mm of PMMA and imaged using the exposure factors typical of those used clinically. Where available, this should be done under AEC. The image processing normally applied to clinical images should be used for this test.

Evaluation

The image should be read using the normal reporting display media and the details scored and recorded. Adjust magnification (1:1 recommended) and window width and level to optimise the appearance of the image.

Remedial level: visibility of details is significantly inferior to baseline

Frequency*: commissioning and every six months.*

3.6 Mean Glandular Dose

3.6.1 Doses to typical breasts

Test protocol

The doses to a range of typical breasts can be assessed using blocks of PMMA as breast substitutes. This method relies on the equivalence in attenuation between different thicknesses of PMMA and typical breasts, as listed in Table 3.

The doses should be determined using the usual clinically selected exposure factors including any automatic selection of kV and target/filter combination. This should be done using AEC where available. It should be noted that since PMMA is generally denser than breast tissue any automatic selection of kV, target or filter may be slightly different from the settings chosen when using real breasts. This can be corrected by adding appropriate spacers to the PMMA to make up a total thickness equal to the equivalent breast. A standard compression force should be applied (e.g. 50-100 N).

Measurements should be made using PMMA blocks with thicknesses of 20 to 70 mm on commissioning. For routine testing, the dose to the standard 53 mm breast should be measured using a 45 mm thickness of PMMA.

Evaluation

The mean glandular dose (*D*) to a breast of thickness *T*, equivalent to PMMA of thickness *P*, is calculated by applying the following formula:

$D = K g c s$

where *K* is the incident air kerma (without backscatter) calculated at the upper surface of the PMMA. The factor *g* corresponds to a breast with a glandularity of 50% and is derived from the values calculated in "Additional factors for the estimation of mean glandular breast dose using the UK mammography dosimetry protocol". by Dance et al. The factor *c* corrects for the difference in composition of typical breasts from 50% glandularity. Note that factors *c* and *g* applied are those for the corresponding thickness of typical breast rather than the thickness of PMMA block used. Where necessary, interpolation may be made for different values of HVL. The factor *s* corrects for any difference due to the choice of x-ray spectrum. Software to calculate the MGD and tables of the *c*, *g* and *s* values can be found on the NCCPM website (medphys.royalsurrey.nhs.uk/nccpm/?s=mgd).

If the system provides estimates of mean glandular dose, compare the values recorded in section 3.3.2 with those calculated above.

Remedial level:

- the remedial levels for doses at different breast thickness are given in Table 3.
- displayed values of MGD > 30% different from calculated values
- change in MGD to standard breast from commissioning value > 25%.

Table 3. Equivalent breast thickness to PMMA thickness and Dose remedial levels.

Thickness of PMMA (mm)	*Thickness of equivalent breast (mm)*	*Glandularity of equivalent breast*	*Remedial level for mean glandular dose to equivalent breasts (mGy)*
20	*21*	*97*	*> 1.2*
30	*32*	*67*	*> 1.5*
40	*45*	*41*	*> 1.8*
45	*53*	*29*	*> 2.0*
50	*60*	*20*	*> 2.5*
60	*75*	*9*	*> 3.5*
70	*90*	*4*	*> 5.0*
80	*103*	*3*	*N/A*

Frequency:

- MGD to the standard breast (45 mm PMMA): commissioning and every 6 months
- MGD at other breast thickness: commissioning and when the AEC software is changed.

3.6.2 Clinical breast doses

Test protocol

The "Guidance for the implementation of the IR(ME)R Regulations 2017" recommend that the MGDs for a selection of 200 women attending for routine screening or assessment on each mammographic system using the procedures described in IPEM 89. This should be undertaken at least every 3 years. Software for making such dose calculations has been published by the NHSBSP Breast Dose – NCCPM. These data should be used to determine that the appropriate DRL for mammography is not being exceeded. The dose audit measure for mammography is the average MGD for mediolateral mammograms for breasts with a compressed thickness of 55 ± 5 mm. A minimum of 40 women should be included in the dose sample.

Evaluation

The current national DRL for this dose audit measure is 2.5 mGy for compressed breast thicknesses between 50 and 60 mm as specified in "National Diagnostic Reference Levels (NDRLs) from 8 July 2025". Corrective action should be taken where exceeding the DRL cannot be clinically justified. Local DRL values may be established. The data collected can also be used to establish how dose varies with breast thickness, and whether the doses are consistent with the doses determined using the standard breast model.

Remedial level:

- dose audit measure > 2.5 mGy
- dose audit measure significantly > local DRL

Frequency: commissioning, and at least every 3 years

Acknowledgements

The authors (Alistair Mackenzie, Oliver Morrish, Claire Skinner and John Loveland) would like to thank the support and help from our physics and breast screening colleagues in preparing this updated guidance.

APPENDIX 1: SUMMARY OF TESTS

Where given, the limiting values are remedial levels unless otherwise indicated. All tests are performed at commissioning. Optional tests are not shown.

Section	Subsection	Test	Frequency	Limiting value
3.1 Beam alignment	3.1.1	X-ray/light field	Six monthly	Misalignment > 5 mm
	3.1.1	X-ray/image field	Commissioning (all settings) Six monthly (common settings)	Remedial: > 5 mm or < 0 mm overlap Suspension: > 10 mm overlap or > 2 mm unexposed at chest wall edge
	3.1.2	Size of imaged field	Commissioning	Ratio of measured to specified dimension < 0.95
	3.1.3	Chest wall – image separation	Commissioning;	> 5 mm between edge image and edge breast platform
	3.1.4	Electronic calliper calibration and reference plane	Commissioning, upgrades	Error >2%
3.2 Detector Tests	3.2.1	Artefacts	Six monthly	Pixel dropout, Any artefacts that may affect clinical image quality
	3.2.1	Uniformity	Six monthly	5% points above baseline
	3.2.2	Detector response	Six monthly	Reference air kerma > 20% change; artefacts in zero dose image
	3.2.3	Noise analysis	Six monthly	Change to baseline: standard deviation > 10%SNR > 10%
	3.2.4	Modulation transfer function	Six monthly	MTF_{50}, MTF_{10}>±10% baseline
	3.2.5.1	SWCTF	Six monthly (if MTF not performed)	10% change to baseline in SWCTF(*f*)
	3.2.5.2	Limiting spatial resolution	Six monthly (if MTF not performed)	limiting resolution < 75% of baseline
	3.2.6	Spatial discontinuity and resolution homogeneity	Six monthly	Discontinuities or blurring
3.3 Automatic exposure control	3.2.7	Image retention	Commissioning	Image retention factor > 0.3
	3.3.2	Variation with absorber thickness – automatic mode	Six monthly Annually for fine focus	Measured CNR < 90% of CNR measured at baseline
	3.3.3	Variation with density control	Commissioning	AEC density control step outside manufacturer's specification
	3.3.4	Variation with AEC detector position	Commissioning	Variation in mAs > 10%
3.4.1 Image display - monitors	3.4.1.1	Viewing conditions	Annually	L_{amb} > 0.25 x L_{min}' Illuminance > 75 lux or <25 lux Any disturbing reflections

	3.4.1.2	Luminance response	Primary monitor - annually Secondary monitor - commissioning only except max. Luminance which is annually.	Minimum < 1.2 cd/m^2 Maximum < 350 cd/m^2 luminance ratio < 250 and >450; luminance variation from *DICOM* standard ± 10% Difference between paired monitors, maximum luminance >5% Minimum < 0.8 cd/m^2 Maximum < 200 cd/m^2; luminance ratio < 250 and >450; luminance variation from *DICOM* standard > 20%
	3.4.1.3	Luminance uniformity	Annually	Remedial: Maximum variation > 15% Suspension: Maximum variation > 30%
	3.4.1.4	Resolution	Annually	Any loss in resolution
	3.4.1.5	Artefacts	Annually	Any artefacts
	3.4.2	Fidelity of transferred images	Any critical changes to the PACS or display	Image quality same on all displays
3.5 Image quality	3.5.1	Detail detection	Six monthly	See Table 2
	3.5.2	TOR(MAM)	Six monthly	Significant changes from baseline
3.6 Dose	3.6.1	Dose vs thickness	Commissioning; AEC software updates	See Table 3; displayed values >30% different from calculated values
	3.6.1	Dose to the standard breast	Six monthly	> 2.0 mGy; > 25% change from commissioning value
	3.6.2	Clinical breast doses	One to three yearly	Dose audit measure > 2.5 mGy; dose audit measure significantly > local DRL

APPENDIX 2: LINEARISATION OF ROI MEASUREMENTS

Before ROI measurements are used in calculations the data should be linearised with respect to the detector response. This is done using the signal transfer property (STP) which describes the relationship between the detector entrance air kerma and the pixel values in the '*for processing*' images (as measured in section 3.2.2). The linearised pixel value is given by inverting this relationship. Provided that the variation within an ROI is small, then the mean value of the linearised pixel values can be approximated by linearising the mean pixel value.

For some systems, the STP is logarithmic:

$$P = a \ln(K) + b \tag{1}$$

where P is the *pixel value* corresponding to a detector entrance air kerma of K, and a, b are fitted coefficients. This can then be inverted to give:

$$P' = K = \exp\left(\frac{P-b}{a}\right) \tag{2}$$

where P' is the linearised value of P.

In cases where the standard deviation and mean pixel value are measured directly on the 'for processing' image without linearization, these can still be corrected. Using the method described in "Validation of correction methods for the non-linear response of digital radiography systems", the linearised standard deviation, σ', can be estimated by simply dividing the *pixel value* standard deviation, σ, by the point gradient of the STP, g. By differentiating (1), g is given by:

$$g = a/K \tag{3}$$

and therefore:

$$\sigma' = \frac{\sigma}{g} = \frac{\sigma K}{a} = \frac{\sigma}{a}\exp\left(\frac{P-b}{a}\right) \tag{4}$$

The linearised SNR is given by:

$$SNR' = \frac{P'}{\sigma'} = \frac{a}{\sigma} \tag{5}$$

The linearised *contrast to noise ratio* between two regions (denoted 1 and 2) as defined in section 3.2.2 is given by:

$$CNR' = \frac{|P_1' - P_2'|}{\sqrt{\frac{\sigma_1'^2 + \sigma_2'^2}{2}}} \tag{6}$$

Using (2) and (4) this becomes:

$$CNR' = \frac{a\sqrt{2}\cdot\left|\exp\left(\frac{P_1}{a}\right) - \exp\left(\frac{P_2}{a}\right)\right|}{\sqrt{{\sigma_1}^2\left(\exp\left(\frac{P_1}{a}\right)\right)^2 + {\sigma_2}^2\left(\exp\left(\frac{P_2}{a}\right)\right)^2}} \tag{7}$$

For a system with a straight line STP:

$$P = a\,K + b \tag{8}$$

This can then be inverted to give:

$$P' = K = \frac{P - b}{a} \tag{9}$$

The point gradient of the STP, *g*, is given by:

$$g = a \tag{10}$$

and therefore:

$$\sigma' = \frac{\sigma}{g} = \frac{\sigma}{a} \tag{11}$$

The linearised signal to noise ratio is given by:

$$SNR' = \frac{P'}{\sigma'} = \frac{P - b}{\sigma} \tag{12}$$

Thus, the only correction needed is to subtract the offset from the pixel values.

The linearised *contrast to noise ratio* is given by:

$$CNR' = \frac{\left|P_1' - P_2'\right|}{\sqrt{\frac{{\sigma'_1}^2 + {\sigma'_2}^2}{2}}} = \frac{\left|P_1 - P_2\right|}{\sqrt{\frac{{\sigma_1}^2 + {\sigma_2}^2}{2}}} \tag{13}$$

Thus, the STP coefficients cancel out and the CNR can be calculated simply from the pixel values. The method can be generalised to any system by using the appropriate STP equation.

APPENDIX 3: GRID TRANSMISSION FACTOR

The measurement of detector response (section 3.2.2) requires an estimation of the air kerma at the detector input plane. Air kerma is usually measured above the breast platform and anti-scatter grid. If the grid can be removed easily, then the measured air kerma can simply be corrected using the inverse square law. However, removal of the grid on some systems may result in the detector being vulnerable to mechanical damage. Care is necessary to prevent items falling directly on the detector, and it may not be easy or advisable to remove the grid frequently. It is recommended in such cases that a *grid transmission factor* (GTF) is measured at the commissioning stage. This factor is then used to correct subsequent detector response measurements made with the grid in place. If a *grid transmission factor* is to be measured, it should be measured under the irradiation geometry used for the detector response tests.

At commissioning, remove the grid following the manufacturer's instructions and taking advice from the installation engineer if necessary. Carefully place the standard attenuator (e.g. 45 mm PMMA) in the beam, on the tube exit port. Select a typical beam quality applicable to 45 mm PMMA. Expose using an mAs value M to achieve a mid-range pixel value P in an ROI at the *standard position*. Replace the grid. With the standard attenuator still in place and using the same beam quality, make exposures at a range of mAs values and for each image measure the mean pixel value in an ROI at the *standard position*. Plot the mean pixel values against mAs and interpolate to find the mAs value M_g required to give pixel value P with the grid in place. The *grid transmission factor* for these exposure conditions is then calculated as:

$$GTF = M / M_g$$

Subsequent measurements of air kerma with the grid in place should be multiplied by GTF and corrected using the inverse square law to obtain the air kerma at the detector input plane.

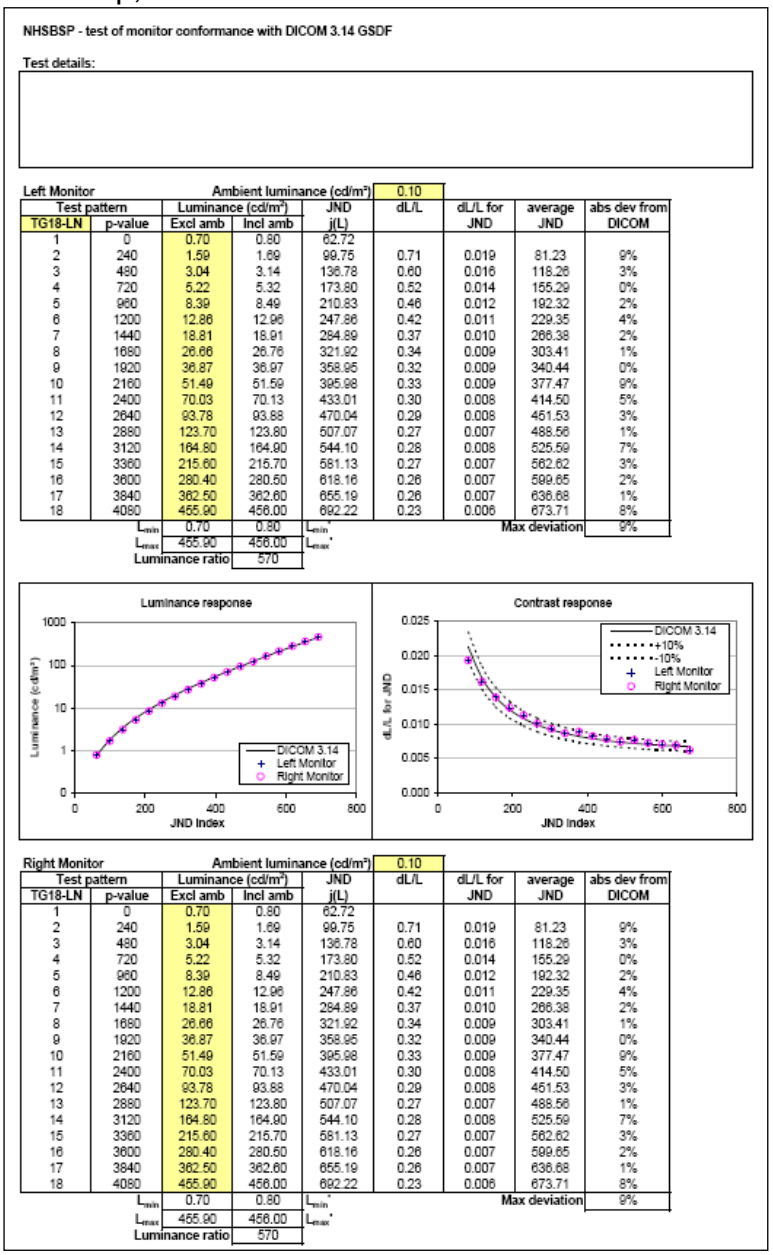
Note that GTF is not the same as the conventional "grid factor", which is an attenuation factor and is measured using a different beam quality and geometry.

APPENDIX 4: GSDF COMPLIANCE OF A DISPLAY DEVICE

Spreadsheet for the calculation of the compliance of a display device to the DICOM 3.14 Greyscale Standard Display Function (GSDF)

The spreadsheet can be downloaded from medphys.royalsurrey.nhs.uk/nccpm/?s=tools, there is an equivalent excel sheet available from EUREF (euref.org/download/monitor-check/). To test that a display device conforms to the *DICOM* 3.14 GSDF, load the TG18-LN, TG18-QC or SMPTE test pattern on the device and follow the test procedure as described in section 3.4.1.2. Fill in the measured luminance values in the table (one table per monitor) and the spreadsheet will calculate conformance to the DICOM standard. Graphs showing the luminance and contrast response will also be plotted (see example below).

AAPM TG18 test patterns can be downloaded from: http://www.aapm.org/pubs/reports/OR_03_Supplemental/
(TG18-LN in lumin-1k-dcm.zip, TG18-QC and TG18-PQC in multi-2k-dcm.zip).



NHSBSP - test of monitor conformance with DICOM 3.14 GSDF

Test details:

Left Monitor — Ambient luminance (cd/m²): 0.10

Test pattern		Luminance (cd/m²)		JND	dL/L	dL/L for	average	abs dev from
TG18-LN	p-value	Excl amb	Incl amb	j(L)		JND	JND	DICOM
1	0	0.70	0.80	62.72				
2	240	1.59	1.69	99.75	0.71	0.019	81.23	9%
3	480	3.04	3.14	136.78	0.60	0.016	118.26	3%
4	720	5.22	5.32	173.80	0.52	0.014	155.29	0%
5	960	8.39	8.49	210.83	0.46	0.012	192.32	2%
6	1200	12.86	12.96	247.86	0.42	0.011	229.35	4%
7	1440	18.81	18.91	284.89	0.37	0.010	266.38	2%
8	1680	26.66	26.76	321.92	0.34	0.009	303.41	1%
9	1920	36.87	36.97	358.95	0.32	0.009	340.44	0%
10	2160	51.49	51.59	395.98	0.33	0.009	377.47	9%
11	2400	70.03	70.13	433.01	0.30	0.008	414.50	5%
12	2640	93.78	93.88	470.04	0.29	0.008	451.53	3%
13	2880	123.70	123.80	507.07	0.27	0.007	488.56	1%
14	3120	164.80	164.90	544.10	0.28	0.008	525.59	7%
15	3360	215.60	215.70	581.13	0.27	0.007	562.62	3%
16	3600	280.40	280.50	618.16	0.26	0.007	599.65	2%
17	3840	362.50	362.60	655.19	0.26	0.007	636.68	1%
18	4080	455.90	456.00	692.22	0.23	0.006	673.71	8%
	L_{min}	0.70	0.80	L_{min}'			Max deviation	9%
	L_{max}	455.90	456.00	L_{max}'				
	Luminance ratio		570					

Right Monitor — Ambient luminance (cd/m²): 0.10

Test pattern		Luminance (cd/m²)		JND	dL/L	dL/L for	average	abs dev from
TG18-LN	p-value	Excl amb	Incl amb	j(L)		JND	JND	DICOM
1	0	0.70	0.80	62.72				
2	240	1.59	1.69	99.75	0.71	0.019	81.23	9%
3	480	3.04	3.14	136.78	0.60	0.016	118.26	3%
4	720	5.22	5.32	173.80	0.52	0.014	155.29	0%
5	960	8.39	8.49	210.83	0.46	0.012	192.32	2%
6	1200	12.86	12.96	247.86	0.42	0.011	229.35	4%
7	1440	18.81	18.91	284.89	0.37	0.010	266.38	2%
8	1680	26.66	26.76	321.92	0.34	0.009	303.41	1%
9	1920	36.87	36.97	358.95	0.32	0.009	340.44	0%
10	2160	51.49	51.59	395.98	0.33	0.009	377.47	9%
11	2400	70.03	70.13	433.01	0.30	0.008	414.50	5%
12	2640	93.78	93.88	470.04	0.29	0.008	451.53	3%
13	2880	123.70	123.80	507.07	0.27	0.007	488.56	1%
14	3120	164.80	164.90	544.10	0.28	0.008	525.59	7%
15	3360	215.60	215.70	581.13	0.27	0.007	562.62	3%
16	3600	280.40	280.50	618.16	0.26	0.007	599.65	2%
17	3840	362.50	362.60	655.19	0.26	0.007	636.68	1%
18	4080	455.90	456.00	692.22	0.23	0.006	673.71	8%
	L_{min}	0.70	0.80	L_{min}'			Max deviation	9%
	L_{max}	455.90	456.00	L_{max}'				
	Luminance ratio		570					

APPENDIX 5: OBJECTIVE IMAGE QUALITY MEASUREMENTS

5.1 Introduction

Quantitative measurements of the presampled modulation transfer function (MTF), the noise power spectrum (NPS) and the detective quantum efficiency (DQE) offer reproducible, objective estimates of x-ray detector noise and resolution properties and are sensitive to changes in detector performance over time. The IEC 62220-1-2 standard describes the measurement of these parameters. However, this document is intended for use by the manufacturers, who can remove the x-ray detector from the system and perform a separate bench test. This is not possible for detectors in clinical use and hence a pragmatic approach, suitable for routine QC conditions, is presented here. Measurement geometry is likely to vary between systems, resulting in a loss of generality. Caution must therefore be exercised when using these metrics to compare across systems. Measured with care, however, these parameters offer significant insight into the performance of an individual detector, they can isolate performance changes over time and are useful when troubleshooting the entire imaging chain. Definitions of the equations used to calculate these parameters are given in the recommended literature. A firm grasp of the theory underlying MTF, NPS and DQE is required before performing these measurements. Given that many QC physicists will not have the time to develop the required software, existing validated/verified is recommended for carrying out the calculations.

The IEC document prescribes standard measurements to be performed at 'the detector surface' (defined as 'the accessible area which is closest to the image receptor plane'); for routine QC measurements there will be additional non-removable parts (e.g. breast support table, anti-scatter grid and/or detector covers) in the x-ray beam during the measurement. A consistent geometry should be employed for a given system/model. When comparing quantitative measurements from different physics centres, the data acquisition conditions must be stated explicitly for the sake of transparency. These include the geometry (position of anti-scatter grid and breast support table, use of collimation/field area), beam energy and detector air kerma, along with the data conditioning parameters used in the calculation of the MTF and NPS (ROI dimensions, sectioning etc.).

5.2 Image processing packages

IQWorks and DRIQ are recommended for carrying out the calculations of MTF, NPS and DQE.

IQWorks is written in C#. This is an open source program downloadable from the IQWorks website at http://iqworks.org. Users will need the .NET framework installed. Input images may be DICOM or various other formats. Output can be .csv, pdf, MS Word, html or database.

DRIQ is a suite of plugins for ImageJ written in Java. Users will need to install ImageJ which is open source and available at https://imagej.net/ij/. They will also either require a Java engine or will need an ImageJ version bundled with Java. DRIQ is free to NHS staff and is available on request from https://www.physicssoftware.co.uk/. The software comes with a user manual available from the same site which includes installation instructions. Input images may be any image format handled by ImageJ or associated plugins. DRIQ extends several of ImageJ's classes to enable, for example, opening a stack containing images of different sizes and from different directories. It provides a DICOM browser optimised for speed and easy sorting of images from any directory structure. Output is in the form of ImageJ windows and tables which can be saved using the standard ImageJ menu. It is possible to programmatically access and call DRIQ functions though documentation for the API is not yet available for download.

5.3 Image type

The first step is to identify/select 'For Processing' images on the system.

5.4 Detector response function

This is described in section 3.2.2. Measure the mean *pixel value* at the standard position using the standard ROI size. Plot the mean *pixel value* against detector entrance air kerma, fit the appropriate curve for the system type (linear, logarithmic or power) and record the fit parameters *a* and *b*. This function is used to linearize the *pixel value* data on a pixel-wise basis in the edge and system response images before calculating MTF and NPS. This must be done for all systems, even systems that produce linear '*for processing*' images. Following this step, the linearized images will have unity gain and zero offset (the mean PV in this image should be equal to the air kerma used to acquire the image).

5.5 Noise Power Spectrum

The NPS describes the variance of an image intensity (image *pixel value*), binned into its frequency components. It is calculated from ROIs taken from a region of a uniformly exposed image. Each ROI undergoes a 2D Fourier transformation to yield an estimate of the noise power spectrum. The individual noise power spectra are then summed and divided by the number of ROIs to obtain a best estimate of the 2D NPS (the ensemble 2D NPS). It is usual to report the 1D NPS; this is sectioned from the ensemble 2D NPS. This can be a radial average for systems with an isotropic 2D NPS, while for detectors with a non-isotropic 2D NPS, the spectra sectioned from the 0° and 90° axes should be recorded separately. The axes (0° and 90° spatial frequency bins) and should not be included in the 1D NPS estimate. The ensemble 1D NPS is then normalized to give the Normalized Noise Power Spectrum (NNPS) by dividing by the mean *pixel value* of the linearized detector response image used to calculate the spectral estimate *i.e.* divided by the detector entrance air kerma used to acquire the detector response image.

IEC 62220-1-2 defines an area of 50 mm x 50 mm for the NPS estimation, divided into ROIs of 256 x 256 which overlap each other by 128 pixels. This strictly limits the physical region from which the NPS is calculated, reducing the effects of non-stationarity and large area non-uniformity on the NPS. However, several images are required to increase the number of noise power spectra in the ensemble and hence reduce uncertainty on the average noise power spectrum.

Evaluation

As the NPS is a function of detector entrance air kerma, it is essential to specify an air kerma level at which it is evaluated. A suitable detector entrance air kerma is 100 µGy. Select the system response image acquired closest to 100 µGy and use the same image throughout the life of the detector. Import the image into the chosen analysis software, ensuring that the image pixel pitch is correctly read from the DICOM header information. If not, correct it manually. Linearize the image using the detector response curve.

For QC purposes, a region 100 mm x 100 mm acquired from the image centre can be used and 128 x 128 pixel ROIs taken from this area. It is recommended that a 2D polynomial is fitted to and subtracted from the 100 mm x 100 mm area before extraction of the 128 x 128 pixel ROIs.

Record the NPS at 0.5 mm^{-1} and 2.0 mm^{-1} (either a radial average or the 0° and 90° axis values separately).

Limiting value: Expect < ± 15% change in NPS at 0.5 mm^{-1} and 2.0 mm^{-1} from previous QC visit value and from baseline.

Frequency: Every six months

Comments

The IEC standard specifies the use of collimation of 100 x 100 mm when acquiring the detector response images in order to control the quantity of the scattered radiation in the image. A higher quantity of scattered radiation effectively leads to a higher detector air kerma per image and hence an increased noise power spectrum. While essential for laboratory detector measurements, the value of collimation in a QC setting is limited and should be considered optional (the collimation is heavy and the same collimator dimension must be used between QC visits).

The anti-scatter grid can influence the measured NPS in a number of ways. First, the grid can introduce structured noise, predominantly of low spatial frequencies, which is often seen along the 0° and 90° NPS axes. Structured noise is multiplicative in nature and increases relative to other noise sources as detector air kerma is increased. The spatially periodic nature of the grid can also introduce spikes, indicating increased noise power at distinct spatial frequencies; this may indicate a grid motion problem. For example, a linear grid with 30 lines cm^{-1} will generate spikes at 3.1 mm^{-1} (and associated harmonics) in the NPS; these will be seen on the axis (0° or 90°) that is parallel to the direction of grid movement in the image.

The presence of the anti-scatter grid in the X-ray beam during detector calibration presents a further complication. Some systems may have flat field corrections explicitly for the case of grid in and grid out of the x-ray beam, while others may have a single flat field correction which presumes that the grid is present. For this latter system type, an ‘imprint’ of the flat field correction will be applied to flood images that have been acquired with the grid removed, leading to a potential increase the structured noise present in the image and hence in the NPS.

APPENDIX 6: CNR ANALYSIS

To apply the standards in the European protocol the limiting value for CNR (using 50 mm PMMA) can be determined using equations 1 and 2. These equations determine the CNR values necessary to achieve the minimum and achievable threshold gold thickness (*Tg*) in the image quality measurements for the 0.1 mm detail size at this thickness.

$$CNR_{mininum} = CNR_{measured} \frac{Tg_{measured}}{Tg_{minimum}} \quad (1)$$

$$CNR_{mininum} = CNR_{measured} \frac{Tg_{measured}}{Tg_{achievable}} \quad (2)$$

[Note that strictly the threshold contrast rather than threshold gold thicknesses should be used in equations 1 and 2. However calculating these contrasts adds an extra layer of complication and it is estimated that the maximum error in estimating the target CNRs is 4%. Thus for routine QC this simplified method is proposed. It is assumed in equations 1 and 2 that the exposure factors for the CDMAM and the 50 mm of PMMA are the same. If for some reason they are not, then a small correction can be applied.]

Equations 1 and 2 calculate the target CNR for a 50 mm thickness of PMMA and the European protocol adjusts this target for other thicknesses of PMMA according to Table 3. The $CNR_{minimum}$ at other thickness of PMMA is calculated according to equation 3 and Table 4. It is not necessary to adjust $CNR_{achievable}$ for PMMA thickness.

$$CNR_{mininum} = z\ CNR_{measured} \frac{Tg_{measured}}{Tg_{minimum}} \quad (3)$$

Table 4. Factors to calculate $CNR_{minimum}$ at different thickness of PMMA

Thickness of PMMA (mm)	z-factor
20	1.15
30	1.10
40	1.05
45	1.03
50	1.00
60	0.95
70	0.90

To evaluate the AEC performance the measured CNR should be plotted against the thickness of PMMA and compared to the target CNR values as shown in Figure 9. In this case the system failed to exceed the minimum acceptable CNR at the 60 and 70mm thickness of PMMA.

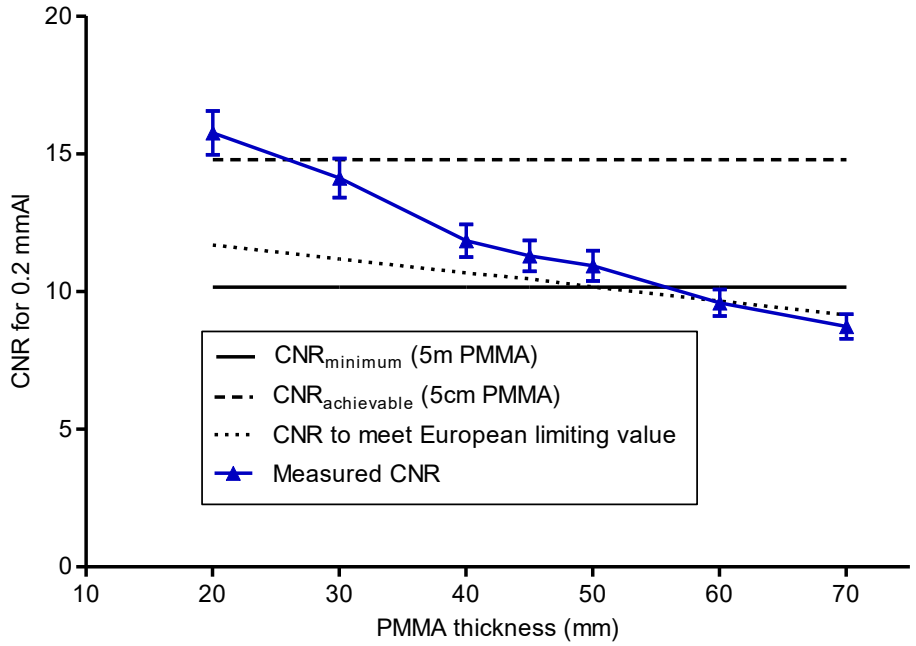


Figure 9. Example of an evaluation of measured CNR against target CNR

Note that in general a change to a higher CNR will be associated with better image quality if all other factors are unchanged. However, if such a change is associated with a change in image sharpness the opposite may be true and an investigation of sharpness and image quality should be undertaken.

Suspension level

Measured CNR < $CNR_{minimum}$ at any thickness

GLOSSARY

Aliasing

A phenomenon which arises in sampling when the frequencies present in the signal to be sampled are higher than can be represented by the sampling process (i.e. higher than the *Nyquist frequency*). Such frequencies will be undersampled and would be erroneously represented as lower frequencies (aliases) in the sampled signal. If the original frequency to be sampled is f, and this is greater than the *Nyquist frequency*, f_N, then this would be aliased in the sampled signal to a frequency ($2f_N$-f).

Contrast to noise ratio (CNR)

Difference in mean pixel value between contrasting detail and background divided by pixel standard deviation in background ROI.

Dark current

Even in the absence of light, electrons will be generated in electronics of the detector.

DICOM

Digital Imaging and Communications in Medicine standard is a set of protocols that enables a piece of medical equipment or software produced by one manufacturer to communicate with software or equipment produced by another. *DICOM* v3.0 is the third version of the standard.

Flat-fielding

An image correction procedure carried out to remove the effects of non-uniformities in the image acquisition process. These include non-uniformity in the x-ray field due to effects such as the anode heel effect etc. In addition, non-uniformities and spatial variations in sensitivity of the image detector are compensated for. Such corrections are usually applied to images from integrated detectors where non-uniformities and spatial sensitivity variations are spatially consistent between images.

Grid transmission factor

The fraction of radiation transmitted by the anti-scatter grid. This is usually determined for well-defined irradiation conditions and geometry.

Limiting spatial resolution

The highest spatial frequency that can be resolved from the image of a high contrast bar pattern test piece. In an analogue imaging system, the limiting spatial resolution is usually determined by the modulation transfer function (MTF) of the imaging system and is defined as the point where the MTF value has fallen to some low modulation figure (usually in the range 3-5%). For a digital imaging system, the limiting spatial resolution may also be affected by the sampling interval (*pixel size*) used when digitising the analogue image data. Sampling theory imposes a maximum spatial frequency which can be represented in a sampled image, and this is determined by the sampling interval and called the *Nyquist frequency*. If the MTF value at the *Nyquist frequency* is still

	significant, then the system will be undersampled and the limiting spatial frequency will be limited by the sampling process, i.e. equal to the highest sampled frequency.
Nyquist frequency	In a sampled system, the highest frequency component that can be represented by the sampled data. The Nyquist frequency is given by $1/2\Delta x$, where Δx is the distance between samples.
Pixel	An abbreviation for “picture element”, a pixel is the smallest discrete element which makes up a digital image. It has a spatial dimension and is assigned a discrete intensity value.
Pixel pitch	The distance between sampling points in a detector. Pixel pitch is distinct from the active area of the detector, which is the size of the light sensitive element. .
Pixel value	A digital value which represents the greyscale level assigned to a *pixel*.
For processing image data	Image data with corrections applied for pixel defects, flat fielding, etc., but with no display processing applied.
Processed image data	Image data that have been processed for display, usually with the application of contrast enhancement and spatial filtering. Often referred to as ‘For Presentation’ image
Raw image data	Image data obtained directly following digitisation. Normally, no corrections due to non uniformity or artefacts will have been applied to these data.
Region of interest (ROI)	A graphically defined region of *pixels*. Software tools usually allow statistics such as the pixel mean and standard deviation to be calculated within the region.
Signal to noise ratio (SNR)	Mean *pixel value* in ROI divided by pixel standard deviation in ROI.
Standard position	A standard position on the breast platform or image where detector response measurements are made. This is defined as a position 40 mm from the chest wall edge and on the midline.
Variance image	*The variance is measured in a small ROI which is moved over the whole image. Viewing the image can identify areas where the variance is different. Example area are regions or lines of defective pixels or the anode heel effect*
X-ray converter	A material layer that absorbs incident x-rays and converts x-ray energy to secondary carriers. Examples are a phosphor layer, in which x-ray energy is converted to light photons, or a photoconductor layer, in which x-ray energy is converted to charge.