COMMISSIONING AND ROUTINE TESTING OF FULL FIELD DIGITAL MAMMOGRAPHY SYSTEMS

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PREFACE

The original version of this report, dated June 2006, was prepared by the Digital Working Party of the NHSBSP QA Coordinating Group for Physics. The members were Mr A Workman (Chair), Ms I Castellano, Ms E Kulama, Mr C P Lawinski, Dr N Marshall and Dr K C Young. Input into the initial stages of drafting from Mr D Goodman should also be acknowledged.

Version 2 of this report, dated September 2006, was prepared by the same authors. It incorporated changes and corrections to the following: equation relating *S*-value and air kerma in section 3.2.1.2, details of detector response evaluation in section 3.2.5, and CNR equation in section 3.3.2.

This document is version 3. It was prepared by the current members of the Digital Working Party: Ms E Kulama (Chair), Ms A Burch, Dr I Castellano, Mr C P Lawinski, Dr N Marshall and Professor K C Young. It includes a large number of changes arising from further experience of using the protocol, in particular a survey of its effectiveness carried out in December 2007 by Dr J M Oduko and members of the Working Party*. The main changes are as follows:

- introduction (section 1) updated to reflect developments in mammography equipment
- clarification about the need to linearise pixel data prior to quantitative analysis (throughout section 3) and the method of doing this (Appendix 3)
- reduction in the recommended size of region of interest (section 3)
- amendment of the method and remedial level for the dark noise test (section 3.2.1.1)
- detector response test method clarified (section 3.2.5)
- measurement and use of a grid transmission factor clarified (section 3.2.5 and Appendix 7)
- SWCTF measurement clarified (section 3.2.6.1)
- image retention measurement clarified (section 3.2.8)
- AEC test method clarified and AEC tests using fine focus added (section 3.3)
- check on any indicator of MGD added (sections 3.3 and 3.6)
- frequency for testing LCD primary monitors reduced to annually (section 3.4)
- measurement of ambient luminance added (section 3.4.1.1)
- separate remedial levels for the luminance response of CRT and LCD primary monitors added (section 3.4.1.2)
- tips on how to obtain raw (unprocessed) pixel values from various systems added (Appendices 2 and 8)
- alternative monitor test patterns referred to (sections 3.4.1.2 and 3.4.1.3)
- spreadsheet for DICOM GSDF compliance amended (Appendix 4)
- *s*-factors for W/Al and W/Ag added (Appendix 6)
- tips on obtaining pre-processed image data added (Appendix 8).

**Review of Measurements on Full Field Mammography Systems*. Oduko JM, Young KC, Burch A, et al. NHS Cancer Screening Programmes 2009 (NHSBSP Equipment Report 0901).

1. INTRODUCTION

The use of digital imaging in general radiography has increased rapidly in recent years and has now extended to mammographic imaging. A number of technologies, including *computed radiography* (CR) and several types of integrated digital detector system, are in use.

Digital imaging provides a wider dynamic range and has the potential to improve contrast resolution compared with film-screen imaging. This may improve diagnostic capability and should outweigh the potential reduction in *limiting spatial resolution*. The improved contrast performance may also allow imaging at higher kV values and with higher atomic number filters, with consequent reduction in patient dose. The greater dynamic range should also reduce the number of retakes required and lead to a greater consistency of quality. A further advantage of digital imaging systems compared with film-screen imaging is the ability to manipulate and possibly enhance the displayed image. The breast dose levels required by current digital imaging systems are, in general, similar to those of a modern mammographic film-screen combination. However, developments in detector design and optimisation of beam quality may eventually result in a reduction in radiation dose. Manufacturers have also suggested that the use of integrated digital detector systems in mammography could result in a reduction in examination time of 30–50% compared with film-screen imaging.

This document introduces full field digital imaging systems for mammography and proposes suitable test protocols for commissioning and routine performance testing. It should be used in conjunction with the current edition of Institute of Physics and Engineering in Medicine (IPEM) Report 89,[1] 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 and will be revised in the light of further experience and developments in technology.

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[2]).

Terms in *italics* are explained in the Glossary.

Table 1 lists the full field digital mammography systems currently marketed in the UK. However, a number of systems already installed in the UK in recent years are no longer available or have been superseded by later versions. These are listed in Table 2.

Table 1 Full field digital imaging systems currently marketed in the UK

Imaging system/ manufacturer	Detector type	UK supplier
Agfa 35-X	CR	Agfa
Agfa 85-X	CR	Agfa
Carestream Classic	CR	Carestream
Carestream Elite	CR	Carestream
Fuji Profect CS	CR	Fuji
Fuji Profect One	CR	Fuji
GE Senographe DS	Integrated detector CsI/amorphous silicon flat panel	GE
GE Senographe Essential	Integrated detector CsI/amorphous silicon flat panel	GE
Hologic Selenia	Integrated detector Amorphous selenium flat panel	Medical Imaging Systems
IMS Giotto	Integrated detector Amorphous selenium flat panel	Southern Scientific Ltd.
Konica Regius 190	CR	Konica Minolta
Philips Mammo-Diagnost DR	Integrated detector Amorphous selenium flat panel	Philips
Philips PCR CosimaX Eleva	CR	Philips
Planmed Nuance	Integrated detector Amorphous selenium flat panel	Xograph
Planmed Nuance Excel	Integrated detector Amorphous selenium flat panel	Xograph
Sectra MicroDose	Integrated detector Scanning silicon wafer array	Sectra
Siemens Mammomat Novation	Integrated detector Amorphous selenium flat panel	Siemens
Siemens Mammomat Inspiration	Integrated detector Amorphous selenium flat panel	Siemens

Table 2 Full field digital imaging systems no longer marketed in the UK or superseded by later systems (see Table 1)

Imaging system/ manufacturer	Detector type	UK supplier
Agfa DM1000	Integrated detector Amorphous selenium flat panel	Agfa
Agfa (Siemens) Novation	Integrated detector Amorphous selenium flat panel	Agfa
Fischer SenoScan	Integrated detector Scanning linear CsI/CCD array	Fischer
GE Senographe 2000D	Integrated detector CsI/amorphous silicon flat panel	GE
Kodak (Carestream) DirectView CR 850	CR	Kodak
Konica Regius 170	CR	Konica Minolta

1.1 Digital imaging in mammography

Most full field digital mammography systems are in principle similar to a traditional mammography x-ray unit, with the film-screen imaging system replaced by a digital detector. Depending on the design and manufacturer, the system may provide a single field size, eg 18 × 24 cm, or two field sizes, eg 18 × 24 cm and 24 × 30 cm. Magnification techniques may include the use of a magnification platform or, alternatively, electronic magnification of the image.

Owing to the wide dynamic range of digital imaging, the delivered exposure does not need to be as precise as for film-screen imaging. However, provision of some form of automatic exposure factor selection is considered essential to ensure the optimisation of radiation dose and image quality.

The x-ray unit will generally have an associated acquisition workstation, which provides a limited range of image processing facilities plus image storage and archive. Hard copy can be produced using an appropriate printing device.

1.2 Detector technologies

1.2.1 Computed radiography

Computed radiography (CR) systems, using laser stimulated photostimulable phosphor technology, are widely used for general radiography. CR for mammography was initially based on such systems and used a high resolution imaging plate contained in a mammography cassette. The standard plate reader used specific algorithms for breast imaging. Nominal *pixel size* was typically 100 µm. Dedicated mammography CR systems are now available with improved imaging capability, with a 50 µm *pixel size*. In the systems made by one manufacturer (Fuji), the phosphor layer is coated on a transparent backing plate, allowing stimulated light to be collected from both sides of the phosphor simultaneously. Table 3 summarises the properties of some CR systems.

An advantage of CR is that the imaging plate (IP) is contained in a cassette that can be used in a conventional bucky assembly or cassette holder. The standard AEC device on the x-ray unit is used with CR. However, this will have to be set up specifically for the CR system (the calibration is unlikely to be the same as for film-screen imaging). In terms of use of the equipment, CR image acquisition is a two stage process involving first exposure of the CR plate, followed by the physical transfer of the cassette to the plate reader where the plate is scanned.

Table 3 Properties of computed radiography (CR) systems

Imaging system	Phosphor	Detector dose indicator	Nominal plate size(s) (cm × cm)	Nominal pixel size (µm)
Agfa 35-X and 85-X	BaSrFBr:Eu	lgM	18 × 24, 24 × 30	50
Carestream Classic, Elite and DirectView CR 850	BaFBr:Eu	EI (exposure index)	18 × 24, 24 × 30	50
Fuji Profect CS* and Profect One*	BaFBrI:Eu	*S* (sensitivity) value	18 × 24, 24 × 30	50
Konica Regius 170 and 190	BaFBr:Eu or CsBr†	*S*-value	18 × 24, 24 × 30	43.75
Philips PCR CosimaX Eleva‡	BaFBrI:Eu	*S*-value	18 × 24, 24 × 30	50

*Dual sided readout.
†Columnar phosphor structure used with Regius 190.
‡Based on Fuji Profect.

1.2.2 Integrated digital detector systems

With integrated digital detector systems the detector and readout mechanism are built into the breast platform of the x-ray unit. Image acquisition is a single stage process in which the image is automatically transferred to the acquisition workstation following exposure. There are several different designs of full field integrated digital detectors. The properties of some systems are summarised in Table 4.

The majority of integrated detector systems are provided with some form of automatic exposure factor selection or AEC. For automatic exposure factor selection, suggested exposure parameters may be displayed on the generator control console prior to exposure depending on, for example, the compressed breast thickness and composition. Certain earlier systems may have only manual exposure control with tabulated kV and mAs values for different compressed breast thicknesses. For AEC the detector signal may be monitored during exposure to provide the required mAs. The kV and added filtration may be automatically set prior to exposure depending on, for example, the compressed breast thickness. Alternatively, the relevant exposure parameters may be automatically derived from a short pre-exposure pulse prior to the main exposure.

The detectors in integrated digital detector systems can be divided into indirect conversion and direct conversion designs. Indirect conversion detectors employ a phosphor layer as the *x-ray converter*. This converts the absorbed x-ray energy into a light signal, which is in turn converted into an electronic signal. With direct conversion detectors, the *x-ray converter* is usually a photoconductor material which converts absorbed x-ray energy to electron–hole pairs. The charge produced is collected and read out as an electronic signal.

Table 4 Properties of integrated digital detector systems

Imaging system	Detector principle	AEC	Detector dose indicator (DDI)	Breast dose indication	Nominal imaging area (cm × cm)	Nominal pixel size (μm)
Agfa DM1000	a-Se flat panel	Yes	Exposure index	Yes	24 × 29	70
Agfa (Siemens) Novation	a-Se flat panel	Yes	–	Yes	24 × 29	70
Fischer Senoscan	Scanning CsI/ linear CCD array	No	ADU	Yes	22 × 29 (11 × 15 in high resolution mode)	54 (27 in high resolution mode)
GE Senographe 2000D	CsI flat panel	Yes	–	Yes	19 × 23	100
GE Senographe DS	CsI/a-Si flat panel	Yes	–	Yes	19 × 23	100
GE Senographe Essential	CsI/a-Si flat panel	Yes	–	Yes	24 × 31	100
Hologic Selenia	a-Se flat panel	Yes	Exposure index	Yes	24 × 29	70
IMS Giotto	a-Se flat panel	Yes	–	Yes	24 × 30	85
Philips Mammo-Diagnost DR	a-Se flat panel	Yes	–	Yes	24 × 30	85
Planmed Nuance	a-Se flat panel	Yes	–	Yes	17 × 24	85
Planmed Nuance Excel	a-Se flat panel	Yes	–	Yes	24 × 31	85
Sectra MicroDose	Scanning silicon wafer	Yes	–	Yes	24 × 26	50
Siemens Mammomat Inspiration	a-Se flat panel	Yes	–	Yes	24 × 30	85
Siemens Mammomat Novation	a-Se flat panel	Yes	–	Yes	23 × 29	70

1.2.3 Indirect conversion detectors

1.2.3.1 CCD devices

Detectors for small field systems use *charge coupled device* (CCD) camera technology. The system uses a single CCD, which is either coupled directly to the image phosphor or uses a fibreoptic taper or mirror–lens system. The imaging area is limited by the size of the CCD and the optical demagnification and varies between 5 × 5 cm and 8 × 5 cm. For full field imaging, either a tiled array of CCDs or a scanning principle can be used. Nominal *pixel size* depends on the configuration of the system and choice of CCD and is typically in the range 25–50 μm. One manufacturer has produced a full field system using a tiled array system with 12 CCDs in a 3 × 4 format; however, this design is no longer in production.

One manufacturer (Fischer) has designed a slot scanning system that uses a moving slot collimator. The detector comprises a linear array of CCD devices, coupled to a phosphor layer, whose movement is synchronised with the scanning x-ray beam. This technique provides a high degree of scatter rejection without the loss of primary radiation that is associated with an antiscatter grid. The exposure time for a typical scan is approximately 5–7 s. This system can operate in either a standard resolution or high resolution mode (nominal *pixel size* 50 μm or 25 μm respectively). It is no longer available in the UK.

1.2.3.2 Flat panel detectors

In these systems a phosphor layer (caesium iodide, CsI) is coupled directly to an amorphous silicon pixelated panel that incorporates an array of photodiode/thin film transistors (TFTs). The phosphor converts absorbed x-ray energy to light. The photodiodes on each *pixel* convert the light to electronic charge, which is then read out by switching the transistor array. Because of their shape and structure, devices incorporating amorphous silicon readout arrays are often termed flat panel detectors. Current full field digital imaging systems for mammography using phosphor based flat panel detectors typically provide a nominal *pixel size* of 100 μm.

1.2.4 Direct conversion detectors

1.2.4.1 Flat panel detectors

These systems use a layer of amorphous selenium as the *x-ray converter* coupled to an amorphous silicon readout array similar to that described above. However, in this case, the panel *pixels* do not incorporate a photodiode. The x-ray photons are converted directly to electronic charge in a selenium layer that is then read out by the TFT array. A high electric field is maintained across the selenium layer, which ensures that the charge is collected without a high degree of lateral spread. This improves the sharpness of the images compared with phosphor based systems in which light spreads laterally. Nominal pixel size is typically 70 μm or 85 μm.

As the shape and structure of amorphous selenium devices are similar to those of amorphous silicon devices, they are also termed flat panel detectors.

1.2.4.2 Multislit scanning

One manufacturer (Sectra) has developed a multislit scanning system in which a fan beam of x-rays is coupled to a linear detector comprising a series of thin silicon wafers. The system has a nominal *pixel size* of 50 μm and is highly sensitive, being able to detect the signal from individual x-ray photons. The manufacturer claims up to an 80% reduction in the breast dose compared with film-screen imaging. Another unit in development uses a similar approach employing a gaseous avalanche detector and again is designed to detect individual photons.

1.2.5 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 CEP report 08022, *Buyers' Guide: Digital Mammography*[3]). 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 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.

1.2.6 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.[4,5] 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). Certain digital systems provide a direct indication of breast dose based on the exposure parameters. The method of calculation should be specified.

1.3 The workstation and image processing

1.3.1 Acquisition (secondary) workstation

The acquisition workstation will be located at the plate reader for CR systems and adjacent to the x-ray unit for integrated detector systems. This provides input/review of patient/examination information, a limited range of image processing facilities and, for integrated detector systems, the operating system for image acquisition. The exact facilities provided vary from manufacturer to manufacturer.

Normally a single monitor will be provided and is designed for image review, but not for soft copy image reporting. On integrated digital detector systems, the image will appear a few seconds after exposure.

The image data from the detector will automatically be processed by application of a suitable algorithm prior to display on the monitor. A range of algorithms may be provided and the system may allow the user to select the most appropriate processing to optimise the image depending on the examination and projection. It is important to note that typical clinical algorithms may not be the most suitable for images of certain test objects.

1.3.2 Reporting (primary) workstation

A reporting workstation will typically be provided with two high specification, high resolution monitors (5 megapixels) mounted adjacent to each other in portrait orientation. The monitors should be matched in performance and must be calibrated to relevant *DICOM* standards. The use of lower specification monitors may compromise clinical image quality.

The workstation carries a comprehensive range of image processing tools to allow image manipulation, plus

image handling, storage and communication facilities. The user may be able to reprocess the image data using additional processing algorithms, and the system may store individual user protocols.

1.3.3 Image processing tools

Current systems are provided with a selection of the following image processing tools. The range and exact functions provided vary from manufacturer to manufacturer.

- Invert – inverts the image greyscale (black to white/white to black), which may assist visualisation of certain tissues.
- Window width/level – alters the contrast/brightness of the image to optimise visualisation of the data displayed.
- Zoom – magnifies the complete image. Various degrees of magnification may be provided.
- Magnify and roam – magnifies a selected area of the image (magnifying glass effect). The magnified area can be moved about the image field.
- Edge enhancement – can improve edge differentiation. However, this may also have the effect of amplifying noise and reducing the visibility of low contrast objects. A range of levels of edge enhancement is normally provided.
- *Region of interest* (ROI) – allows a specific region of the image to be selected. Numerical data referring to the ROI may be provided.
- Measurement – provides measurement of distance in real space.
- Statistics – provides numerical data such as pixel count, histograms.

1.3.4 Advanced features

CAD (computer aided detection) is likely to become a standard option with both CR and integrated digital detector systems. Other advanced features in development are telemammography (the provision for rapid transmission of images) and various forms of three-dimensional reconstruction (tomosynthesis).

1.3.5 Image storage and archive

Digital images comprise a large amount of data (8–50 MB depending on the technology used, the resolution mode and size of the image field), which need to be stored, both in the short and in the long term. Short term storage is generally provided at the acquisition workstation on the computer hard disk. For long term storage, a separate archive facility is required and may take several forms. Digital archiving should provide compact, secure storage with the potential for simple transfer of images from centre to centre if required:

- recordable CD (CD-R)
- digital video disk (DVD)
- digital archive tape (DAT) drive
- magneto-optical disk (MOD).

Data compression can be used to improve the efficiency of storage, but this is usually only capable of compression ratios of the order of 2:1 without loss of information. To achieve an efficient workflow, links to a picture archiving and communication system (PACS) must be considered.

All current systems claim a level of *DICOM* compatibility, which should be fully explained in the manufacturer's *DICOM* conformance statement and is an important consideration if the unit is to be networked.

1.3.6 Hard copy device

Soft copy imaging is the only way to achieve the true potential and true value of digital imaging (for example, saving on the cost of film and reducing storage requirements). However, images can be printed on to film using a dedicated hard copy device such as a high resolution laser printer. The printer must be set up to give a true representation of the soft copy image and be calibrated to relevant *DICOM* standards.

Hard copy images may be required in a screening programme to allow comparison of images during the initial transition from film-screen imaging to digital imaging. Alternatively, existing images could be converted to digital format using a film digitiser, although this process is not generally considered cost-effective.

Detailed comparative specifications of laser printers used for mammography are given in KCARE/MHRA report MHRA 03095.[6]

1.4 Quality assurance procedures

In addition to the tests and measurements described in this document, integrated digital detector systems may be provided with an integral quality assurance (QA) facility. This may be automatic on switching the system on and will provide a pass/fail indication. Alternatively, 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.

2. TESTING METHODOLOGY

This section explains the methodology behind the testing protocols outlined in section 3 of this report. It gives some background information about where and why differences may arise in both the testing methodology and the expected performance parameters for digital mammography systems compared with film-screen based mammography systems. It is assumed that the reader is already fully familiar with the testing procedures for film-screen based mammography systems.

Digital mammography differs from film-screen mammography in a number of fundamental ways. The fact that the image acquisition system captures a digital image means that the display of the image data is an independent process from the acquisition. As the image is not directly captured onto film, different types of *x-ray converter* and readout system can be used to acquire the image information. These may have more favourable properties in terms of quantum efficiency, sharpness and energy response, linearity and dynamic range. The wide dynamic range of the image acquisition system along with the ability to post-process the image for display means that there is no longer a trade-off between acquisition latitude and display contrast, as with film-screen systems. The use of digital detectors potentially allows a wider range of x-ray spectra to be used, different means of scatter rejection such as slot scans and, in some cases, the removal of the scatter rejection grid.

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 *pre-processed 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).

In order 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 by exporting images in a *DICOM* format for remote analysis. The facility to import *DICOM* test images should also be available.

2.1 The reference plane

The detector *pixel size* is defined at a reference plane parallel to the detector. In the case of CR, this will be the detector plane itself, but 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.

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 must be established. 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 (eg different target/filter combinations and focal spot sizes). Image uniformity is assured for the calibration conditions. However, deviations from these, eg 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. It is therefore important to evaluate image uniformity using the calibration conditions specified by the manufacturer if the results are to be compared against performance specifications. For this reason this document offers the option of following the manufacturer's protocol.

CR systems may also have internal compensation to reduce image non-uniformity due to spatial variations in the efficiency of light collection. With such systems no compensation for x-ray beam non-uniformity takes place and uniformity level similar to film-screen systems should be expected.

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 polymethylmethacrylate (PMMA). The PMMA is placed proximal to the x-ray tube to reduce the amount of scatter reaching the detector; placing the PMMA close to the detector would possibly 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 direct measurement of the air kerma incident on the detector can be made. For routine measurements the grid can be left in (thus protecting the detector) 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. For CR systems, in which a logarithmic relationship exists between *pixel value* and detector exposure, this relationship is demonstrated by plotting *pixel* variance against log(exposure). 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[7] or structure noise[8] in the images.

2.4 Detector resolution

The resolution of a conventional film-screen based mammography system is usually 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%). For film-screen systems, in which the general shapes of the MTFs are similar, this provides a reasonably acceptable comparison and

constancy measure for the purpose of system quality control. In fact, for normal quality control, with the test grating positioned approximately 4 cm above the breast platform, it is the resolution not just of the film-screen system that is determined, but of the entire imaging chain. For many systems the limiting resolution in this case is determined primarily by the geometrical blur due to the finite size of the focal spot of the x-ray tube.

The resolution of a digital imaging system detector will 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 size*, 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, owing 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 undersampled.

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 that the *pixel size* by a factor of √2 and the *Nyquist frequency* will therefore be greater by this factor.

An *x-ray converter* limited digital detector is one in which the limiting resolution is set by the inherent unsharpness of the *x-ray converter*. The limiting resolution is lower than that determined by the *Nyquist frequency* based on the *pixel* sampling interval. In this case it is easier to determine the limiting resolution of the detector and use this as a measure of resolution for quality control purposes.

Examples are shown in Table 5 for both *pixel* limited and *x-ray converter* limited detectors.

The normal protocol for assessing system resolution with film-screen systems is to place the resolution test grating at a certain height above the breast platform. The results are thus influenced by both the unsharpness of the detector and the geometrical unsharpness arising from the focal spot size. Most of the detectors for full field digital mammography have limiting resolutions that are substantially lower than those for film-screen systems. Therefore, it is unlikely that the resolution limit for the system for full field digital mammography

Table 5 Examples of pixel limited and converter limited detectors

Pixel limited	Converter limited
GE 2000D	Fuji 5000 MA (20 pixels/mm sampling)
LoRad Selenia	Fischer Senoscan (25 µm pixel size)
Fuji 5000	
Fuji 5000 MA (10 pixels/mm sampling)	
Fischer Senoscan (50 µm pixel size)	
Sectra MicroDose	

systems, obtained under normal measurement conditions, will be set by the focal spot size. Measurements of system resolution for digital mammography may be relatively insensitive to changes in geometrical unsharpness.

In this document, an alternative approach to monitoring resolution is outlined based on the measurement of the modulation of certain square wave frequencies. This concept is analogous to MTF, in which the modulation of sinusoidal signals is measured over a range of spatial frequencies. To make a distinction from MTF, this measure is termed the square wave contrast transfer factor (SWCTF). This can be measured fairly simply at a number of relevant and different bar pattern frequencies. This protocol advocates resolution measurements only on the detector.

2.5 Automatic exposure control

The limited latitude of film-screen mammography makes the use of AEC mandatory. The goal of the AEC in this case is to produce films of suitable density independent of the thickness and composition of the breast being imaged and the beam quality being used.

Since digital imaging systems have a wide acquisition dynamic range and display brightness is independent of exposure, the requirement to control exposure to the detector is less stringent. Nonetheless, the control of exposure for a digital imaging system remains important. However, the goal in controlling exposure for the digital imaging system is not the same as that for a film-screen system. For a film-screen system, the maintenance of optical density with variation in imaging parameters is linked to the maintenance of the amount of energy absorbed in the radiographic screen. A similar strategy could be used for determining the performance of AEC devices for digital imaging systems. The image *pixel value* is usually directly related to the amount of x-ray energy absorbed in the *x-ray converter*. Thus, maintaining this value would be analogous to maintaining optical density for film-screen systems. There is a need to ensure that the performance of the AEC system meets the specification of the manufacturer. However, there is also a need to ensure the optimisation of the performance of the system with respect to patient dose and image quality.

2.6 Optimisation

Most modern mammography units adjust exposure factors and beam quality automatically based on the thickness or transmission of the breast. For a constant detector dose, increasing beam quality reduces dose to thicker breasts but also reduces subject contrast. With digital imaging systems, the display contrast can be enhanced and the loss of subject contrast can to some extent be compensated for. However, the limitation in image quality is the loss of *contrast to noise ratio* (CNR). One strategy for an AEC system for a digital imaging system is to maintain CNR whilst also keeping breast dose and exposure time at an acceptable level.

2.7 Display systems

In the digital imaging environment it is essential to obtain consistent display of soft and hard copies 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.

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 soft and hard copy 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. In the case of cathode ray tube (CRT) monitors, luminance and resolution are expected to deteriorate with time as the phosphor ages and the cathode becomes depleted. Anecdotal evidence suggests that the display devices are the weak link in the imaging chain, and hence it is imperative not to overlook them.

Comprehensive acceptance and routine tests have been developed for display systems by Task Group 18 of the American Association of Physicists in Medicine.[9] They form the basis of the tests described by the American College of Radiology Imaging Network (ACRIN) full field digital mammography screening trial, and in the European protocol for mammography screening.[5] They make use of a set of test patterns designed by the task group for this purpose, which can be easily identified by the prefix 'TG18-'. They are available on the web, but it is not always straightforward to incorporate them in an image archive.

In common with the groups identified above, we have adopted a pragmatic approach to monitor testing. We have selected tests from the TG-18 document that we believe will easily and reliably demonstrate monitor performance without the need for 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-QC or SMPTE (Society of Motion Picture and Television Engineers) test pattern and a light meter. Overall imaging performance can be evaluated visually using the TG-18MM image, which contains some microcalcifications, or a clinical reference image selected by the users showing similar subtle detail.

2.8 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 depth 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 number of digital mammography systems use harder x-ray beam qualities than film-screen mammography systems. This reduces subject contrast; with the digital imaging system, however, the ability to manipulate the display contrast may allow acceptable contrast rendition of image detail. 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. (See Appendix 5 for the calculation of the contrast of the gold discs in the CDMAM phantom using standard beam conditions.) 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.

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 5–10%) to that of glandular tissue/calcifications over the range of x-ray spectra which may be used clinically (K C Young and 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.[5] They have been designed to ensure that digital systems have a detail detection performance that is at least as good as current film-screen systems. 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.

Ideally, the image quality measurements described in this protocol should be made using the soft and hard copy display devices provided with the system. In practice, this is not always possible. An acceptable alternative is to bring copies of the unprocessed test object images back to an imaging laboratory equipped with a diagnostic quality *DICOM* calibrated display. Provided sufficient magnification and windowing are used, the visibility of small details should be limited by the image content rather than by the display itself. If a display (monitor or printer) does affect the measurements it is either inappropriate or faulty.

2.9 Dose

The methods of measuring dose are the same as those described in IPEM Report 89.[1] Where blocks of PMMA are used, the dose calculated is the MGD to an equivalent breast as described by Dance et al.[10] 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 Dance et al.[10] The limiting values for MGD for a 53 mm thick standard breast model (measured using 45 mm of PMMA) are the same as defined in IPEM Report 89[1] for film-screen systems. The additional limiting values for other thicknesses were derived from the European protocol.[5] The limiting values for measurement on samples of women are the local and national diagnostic reference levels (DRLs) established as for film-screen systems (IPEM Report 88[11]).

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 not be affected and therefore should be undertaken following the protocols outlined in the latest edition of IPEM Report 89[1]:

- 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 (eg use a lead sheet to cover the whole of the detector area). Also, when such tests are being performed on x-ray systems with a loaded CR cassette in the bucky, it will be necessary, at the conclusion of the tests, to process the IP with the primary erase function of the CR unit, to ensure that all signal is removed from the image plate.

Some of the test protocols in this document are not applicable to scanning fan beam systems. This is due to the difficulty with such systems in achieving the setup required to perform the test.

The tests listed for CR systems refer specifically to the Fuji Profect CR system because this was the first one to be evaluated by the NHSBSP. These tests may be equally applicable to other manufacturers' CR systems. However, in this version of the protocol IP read modes and test limiting values are given only in terms applicable to Fuji systems. The read modes of Fuji CR systems are described in Appendix 2. This document will be updated in due course, once more information is available on the necessary setup parameters for other manufacturers' systems and more experience has been gained on the application of the protocol to these systems. Some guidance on read modes and settings for other CR systems can be found in AAPM Report 93.[12]

When analysing images numerically, a standard sized ROI should be used. This should be small enough to avoid errors caused by non-uniformity (eg heel effect[13]) and will be typically 5×5 mm. When performing tests users should be aware that the view selected (laterality), eg 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 visible image field/detector
- size of the imaged field (not CR)
- separation between the chest wall edge of the visible field and the chest wall edge of the breast support platform.

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 the 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
Place a loaded film-screen cassette or CR cassette on the breast support platform so that it overlaps the front edge. For CR based systems, a CR cassette should also be placed in the bucky. Position an alignment test object with radio-opaque scale markings to cover the full imaging area (Figure 1). Note the position of the light field boundaries on the scale markings. Acquire an image (use low values of kV and mAs). Process the film/CR cassette, and view the image on a display monitor.

Evaluation
The displacement between the x-ray field and the light field is assessed on the film/IP by using the radio-opaque markers to localise the boundaries of the x-ray field and compare them with the known position of the light field boundaries.

The alignment of the x-ray field to the visible image field/detector is assessed by using the image of the radio-opaque markers to localise the boundaries of the detector and compare them with the known position of the x-ray field boundaries. A geometric correction will be needed to transform the measurements to the detector plane. A corresponding transformation should be made to the dimensions of the x-ray field. The correction factor can be easily determined by measuring a known distance between the radio-opaque markers.

Alignment of x-ray field to light 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 or > 10 mm overlap along left or right edge with respect to the image.

Frequency: commissioning and every six months.

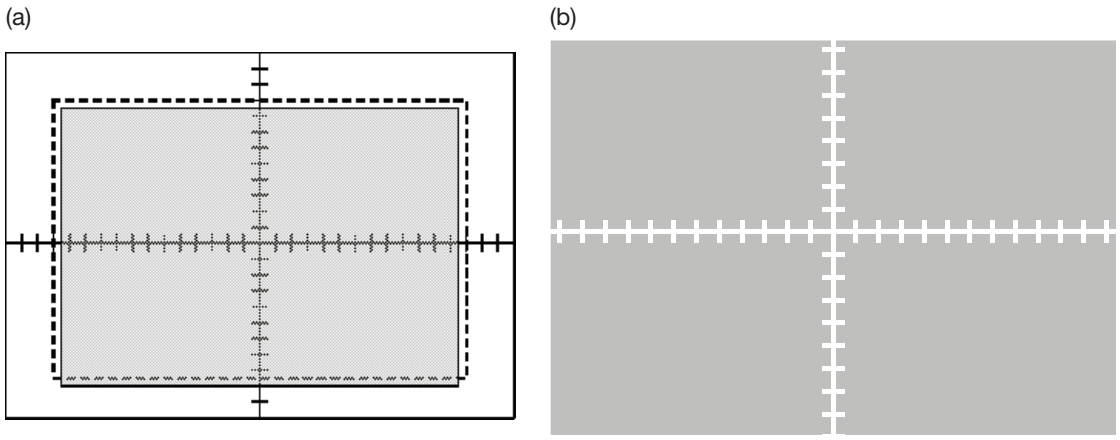


Figure 1 (a) Example of a beam alignment test object. The boundaries of the light field are shown by the dashed line; the x-ray field is shown as grey. (b) Example of a digital test object image.

3.1.2 Size of imaged field

Test protocol
Use image acquired in section 3.1.1.

Evaluation
A direct indication is given by the test object's scaled markings. 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 plane defined by the manufacturer or the detector plane.

Remedial level: ratio of measured to specified dimension < 0.95.

Frequency: commissioning.

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

Test protocol
Align the edge of the test object with scale markings with the edge of the breast support platform at the chest wall. For CR systems insert a CR cassette in the bucky. Acquire an image and view the image 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 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:
- integrated detector: commissioning
- CR system: commissioning and every six months.

3.2 Detector performance

The following measurements aim to evaluate certain performance characteristics of the imaging detector. Cassette based CR systems require a number of additional tests which are not required of integrated digital detector systems. These tests are outlined in section 3.2.1. The remaining tests in section 3.2 are applicable to both CR and integrated digital detector systems.

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 4 cm from the chest wall edge.

3.2.1 Specific CR tests

The tests in this section should be performed on CR mammography systems before they are put into service and at the specified intervals thereafter.

Image plates should be visually inspected for signs of mechanical damage. Image plate numbers used in the

tests should be recorded along with the results of the tests. For many of the tests undertaken on CR systems it may be helpful to identify a QA IP or QA cassette which can be used repeatedly in the tests.

The latent image formed in the IP by absorption of x-ray energy will decay with time from exposure. This means that, for a given fixed exposure, less light will be stimulated from the plate after greater delay times between exposure and readout. Variations in delay time will result in variations in the *detector dose indicator* (DDI) or *pixel value* depending on readout mode. Therefore, tests on CR systems which rely on a measurement of either the DDI or *pixel value* should be carried out using controlled and reproducible delay times between exposure and readout.

Image plates will build up a background signal due to environmental exposure. Thus, it is necessary to 'erase' (primary erasure where available) all image plates before commencing testing. This erasure process can be carried out in conjunction with the test described in section 3.2.1.1 below.

3.2.1.1 Dark noise

Test protocol

Erase an IP, then read using the TEST/SENSITIVITY read program. Set to FIX read mode with an *S*-value of 10 000.

Evaluation

The image should be uniform without artefacts.

Record the mean *pixel value* and the standard deviation (SD) in an ROI covering the whole image.

Remedial level: mean pixel value > 450, SD > 1.

Frequency: commissioning and annual.

3.2.1.2 CR reader sensitivity

It is necessary to verify the calibration of *detector dose indicator* number (DDI). For Fuji systems, the DDI is the sensitivity or *S*-value. This test should be performed under specified calibrated beam conditions and with a specified delay time.

Test protocol

Use the TEST/SENSITIVITY read program and set to SEMI-X 2 read mode for this test. Place an ionisation chamber on the breast support platform at the *standard position*.

Select 25 kV Mo/Mo and adjust the mAs to give a dose reading (D_R) close to the Fuji standard dose of 174 µGy.[14] Record the mAs and dose. Place the QA cassette on top of the breast table of the x-ray unit and expose at the same mAs setting. Read the IP after a fixed delay time of 10 minutes and record the DDI (S_{exp}).

Evaluation

Calculate the *S*-value (S_{cal}) for a detector entrance air kerma of 174 µGy using the following relationship:

$$S_{cal} = S_{exp} \frac{174}{D_R}$$

Remedial level: calibrated S_{cal} > 20% from 120.

Frequency: commissioning and every six months, and following CR reader calibration.

3.2.1.3 Image plate (IP) matching and artefacts

Image plates/cassettes should be matched both in their sensitivity (DDI per unit exposure) and in the mAs derived under automatic exposure.

Test protocol

Place a cassette in the bucky and 40 mm PMMA or 2 mm Al sheets in the beam on the tube exit port. Select 28 kV Mo/Mo and at a manual mAs setting (eg 30 mAs) expose the cassette and process after a consistent delay time. Use the TEST/SENSITIVITY read program and set to SEMI-X 2 read mode for this test.

Record the DDI. Repeat for each cassette/IP.

Using the above conditions, expose each cassette under automatic exposure control. Record the mAs for each exposure.

Erase each IP before proceeding to the next test.

Evaluation

View each image with a narrow display window. Plates showing scratches or marks should be cleaned and the test repeated. Check consistency of the DDI and mAs.

Remedial level:

- maximum variation in DDI > 5% of mean
- maximum variation in mAs > 5% of mean (under AEC mode)
- plates with permanent scratches and marks should be removed from service.

Frequency: new cassettes/plates at commissioning. Operators should carry out these tests every six months.

3.2.1.4 Laser function

Test protocol

Place a straight metal edge on a CR cassette in a direction almost perpendicular to the laser scan direction (ie at a small angle to the subscan direction). Expose using 28 kVp Mo/Mo (or typical beam quality used clinically if not available) and a manual mAs (30 mAs).

Repeat for different cassette formats. (Note: the direction of laser scan relative to the plate aspect may be different for different cassette formats.)

Evaluation

Use a narrow display window to view the edges of the test piece.

Remedial level: any laser jitter or dropout.

Frequency: commissioning and annual.

3.2.2 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) (eg the star test object) on the breast support platform. Acquire an image (use low values of kV and mAs). In the case of CR, process the IP using the TEST/CONTRAST program and a SEMI-X 2 read mode. 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.

Measure the known 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:

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

Remedial level: error > 2%.

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

3.2.3 Uniformity

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

Test protocol
Place a 40 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 at 28 kV Mo/Mo under AEC or at a typical mAs value (eg 50–70 mAs). If this setting is not possible, use the most appropriate clinical setting. The compression paddle and grid should be in place as for clinical use. Repeat for the different target/filter combinations used clinically.

Alternatively, follow the manufacturer's protocol if available.

For integrated systems acquire *pre-processed image data*. For CR systems, use the TEST/CONTRAST read program and set to SEMI-X 2 read mode.

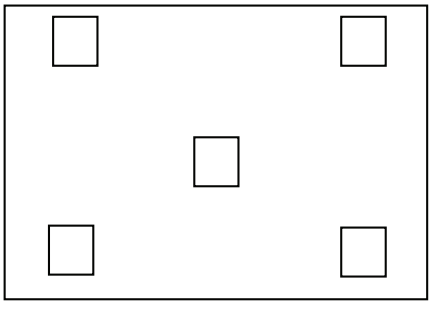


Figure 2 ROI locations for measuring uniformity for a system with an integrated detector.

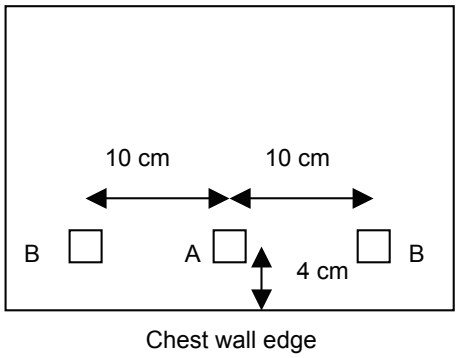


Figure 3 ROI locations for measuring uniformity on a CR image.

Evaluation

(a) Integrated detector. Measure the mean pixel value for an ROI at a position in the centre and at each of the four corners of the pre-processed image as shown in Figure 2. The exact placement of the ROIs is not important; however, the positions should be recorded so that the same setup may be used on subsequent routine visits. Subtract any offset (as determined in section 3.2.5) before calculating the uniformity.

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

$$\text{Deviation from centre mean} = \frac{(\text{Centre ROI mean} - \text{Corner ROI mean})}{\text{Centre ROI mean}} \quad 100$$

(b) CR system. Place ROIs on the CR image as indicated in Figure 3.

Measure the mean pixel value in ROIs placed at positions A and B. Linearise the measured mean pixel values as described in Appendix 3 before calculating the uniformity.

Remedial level:

- *integrated detector:* maximum deviation from centre mean > 10%
- *CR system:* difference in ROI mean value between positions A and B > 10%, difference in ROI mean between left and right > 10%.

Frequency: commissioning and every six months.

3.2.4 Artefacts

This test is applicable only to integrated detectors. For CR systems refer to section 3.2.1.3.

Test protocol

Use the image generated in the uniformity test (section 3.2.3).

Evaluation

View the images on the reporting workstation using a high contrast presentation. This may require readjustment of display window level and width. Set the window width to about 10% of the maximum pixel value (around 400 for 12-bit systems). Record the window width and level settings for use on subsequent occasions.

To separate image artefacts from monitor artefacts, rotate or pan images with significant artefacts. 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 (eg lines, clusters).

Remedial level:
- dead *pixel* dropout – see manufacturer's specification
- other artefacts that may affect clinical image quality visible.

Frequency: commissioning and every six months.

3.2.5 Detector response

The following test should be carried out where possible with the antiscatter 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 7). For CR systems, simply place the cassette on top of the breast platform rather than in the bucky.

Test protocol
The protocol is best carried out in two stages. First, a measurement of x-ray tube output is made with the standard attenuation (45 mm PMMA) placed close to the x-ray tube port. This part of the test can be done at the end of the tube and generator tests. An integrated detector should be protected from direct exposure by covering it with a sheet of highly attenuating material (eg lead or steel). Place the ionisation chamber at the *standard position* on the breast support table. Put the compression paddle in place. Use full field collimation (eg 18 × 24 cm or 24 × 30 cm). Place the standard attenuator (45 mm PMMA) in the beam close to the tube exit port. At commissioning, select a typical beam quality (kV/target/filter) applicable to 45 mm PMMA (eg as selected by AEC), and use the same quality every time the test is repeated on the unit. Set three mAs values (eg 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. 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 ionisation chamber but leave the attenuator close to the tube exit port and the compression paddle in place.

(*a) Integrated detector*. Remove the protective sheet from the detector. Acquire *pre-processed image data*, varying the mAs values between 5 and 400 to expose the detector to air kerma levels that cover its dynamic range. Record the mAs and calculate the air kerma at the detector entrance plane using the output factors from stage one.

(b) CR system. Use the TEST/CONTRAST read program and set to FIX read mode with an *S*-value of 100. Place a cassette on the breast platform (not in the bucky). Vary the mAs to give air kerma values covering a range of around 30–1000 μGy. Read the IP as soon as possible after each exposure. Record the mAs and calculate the air kerma at the IP using the output factors from stage one.

To check the autoranging function of CR units, the following protocol should be followed. Use the TEST/SENSITIVITY read program and set to a SEMI-X 2 read mode. Expose a cassette on the breast platform at the mAs values determined in the previous test and read as soon as possible after exposure. Record the DDI and the mAs for each exposure. Calculate the air kerma at the IP using the previously measured output factors.

Evaluation

(a) Integrated detector. Measure the mean *pixel value* and standard deviation of the mean in an ROI placed at the *standard position* on each pre-processed digital image.

To obtain the detector response, plot mean *pixel value* against detector entrance air kerma as shown in Figure 4. Fit a trend line of the form $y=ax+b$ and record the gradient (a) and offset (b). 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*. Choose a clinically representative target *pixel value* (PV_{clin}) (eg 900 for GE Senographe DS or 400 for a Hologic Selenia system, or as determined in section 3.3.2) and 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}).

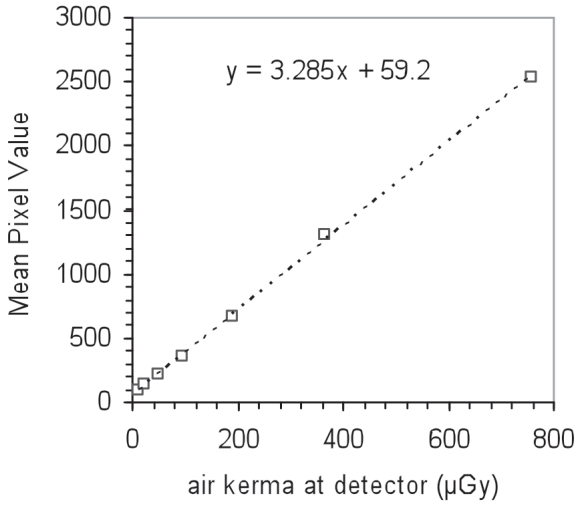


Figure 4 Pixel value versus detector entrance air kerma.

To analyse the noise, plot standard deviation against detector entrance air kerma using log–log axes. Fit a power trend line of the form $y=cx^d$ to points in the proximity of PV_{clin} and record c and d (for example as in Figure 5). For a quantum limited detector, the expected relationship is $y=cx^{0.5}$. The presence of certain noise sources in the system other than quantum noise (eg 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:

$$SNR_{ref} = (PV_{clin} - b)/\sigma_{ref}$$

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.

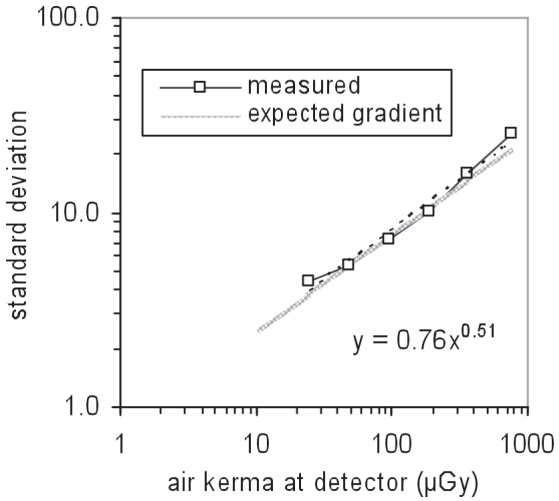


Figure 5 Standard deviation versus detector entrance air kerma.

(b) CR system. For the FIX read mode images, measure the mean pixel value and standard deviation of the mean in an ROI placed at the *standard position* on each pre-processed digital image.

To obtain the detector response plot mean *pixel value* against ln(detector entrance air kerma) – this should result in a linear relationship. Fit a trend line of the form $y = ax + b$, where x is ln(detector entrance air kerma), and record the gradient (a) and offset (b). Choose a clinically representative target PV_{clin} (eg 512 for Fuji) and 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}).

Linearise the standard deviation values (Appendix 3). Plot the linearised values (σ') against detector entrance air kerma using log–log axis and fit a power trend line as for integrated detector systems. Use the fitted line to determine the linearised pixel standard deviation at the detector reference air kerma (σ'_{ref}). Use this to calculate the SNR at the detector reference air kerma:

$$SNR_{ref} = AK_{ref}/\sigma'_{ref}$$

These measurements at commissioning will serve as the baseline for subsequent noise measurements. For routine tests, compare each linearised standard deviation with 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.

For the SEMI-X read mode data, plot DDI against detector entrance air kerma using log–log axis. This should provide a linear plot. The product of entrance air kerma and the DDI for each point should be a constant.

Remedial level:
- detector reference air kerma > 20% change from commissioning value
- standard deviation (linearised) at any entrance air kerma > 10% change from baseline at same air kerma
- SNR change > 10%.

Frequency: commissioning and every six months.

3.2.6 Detector resolution

3.2.6.1 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 that 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. For cassette based systems such as CR, this may be achieved by placing a cassette on top of the breast platform and placing the test pattern in contact with the cassette. For integrated systems, the test pattern should be placed 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. For scanning readout systems such as CR, images should be obtained with the bars orientated at a small angle to the scan and subscan directions. For integrated systems acquire pre-processed images. For CR systems, use the TEST/CONTRAST read program and a SEMI-AUTO read mode. 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 normalising factor. To do this, use an ROI to measure the mean *pixel value* in a region corresponding to the most attenuating level in the test piece, M_B (ie 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 least attenuating region in the test piece, M_S (ie 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.

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

$$M_0 = \left| M_S - M_B \right|$$

The transfer factor at a given frequency is given by:

$$\mathrm{SWCTF}\left(f\right) = \frac{M\left(f\right)}{M_0}$$

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.

3.2.6.2 Limiting spatial resolution

Additionally 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. For cassette based systems such as CR, this may be achieved by placing a cassette on top of the breast platform and placing the test pattern in contact with the cassette. For integrated detector systems, the test pattern should be placed 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 of 26 kV and approximately 15 mAs. For integrated systems, acquire pre-processed images. For CR systems, use the TEST/CONTRAST read program and a SEMI-AUTO read mode.

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:
The present NHSBSP standards for high contrast limiting resolution for film-screen systems are not applicable to digital imaging systems and it is unlikely that an absolute minimum standard for this measure will emerge for digital imaging systems.

- 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.

3.2.7 Spatial discontinuity and resolution homogeneity

Test protocol
Place an extremely fine radio-opaque mesh (eg a contact test tool for mammography) on the breast support table. For CR systems, place the cassette on top of the breast platform and the test tool on top of the cassette. Expose manually using low exposure parameters (eg Mo/Mo at 28 kV, 10 mAs). For integrated systems, acquire pre-processed images. For CR systems, use the TEST/CONTRAST read program and a SEMI-AUTO read mode.

Evaluation
Inspect the image for discontinuities such as line artefacts due to data interpolation or laser jitter that can be attributed only to the detection process (eg 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.

3.2.8 Image retention

Test protocol
Place a 45 mm PMMA phantom on the breast support table and acquire two images using manual exposure factors similar to those used under clinical conditions (eg as described in section 3.3.2 for 45 mm 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.

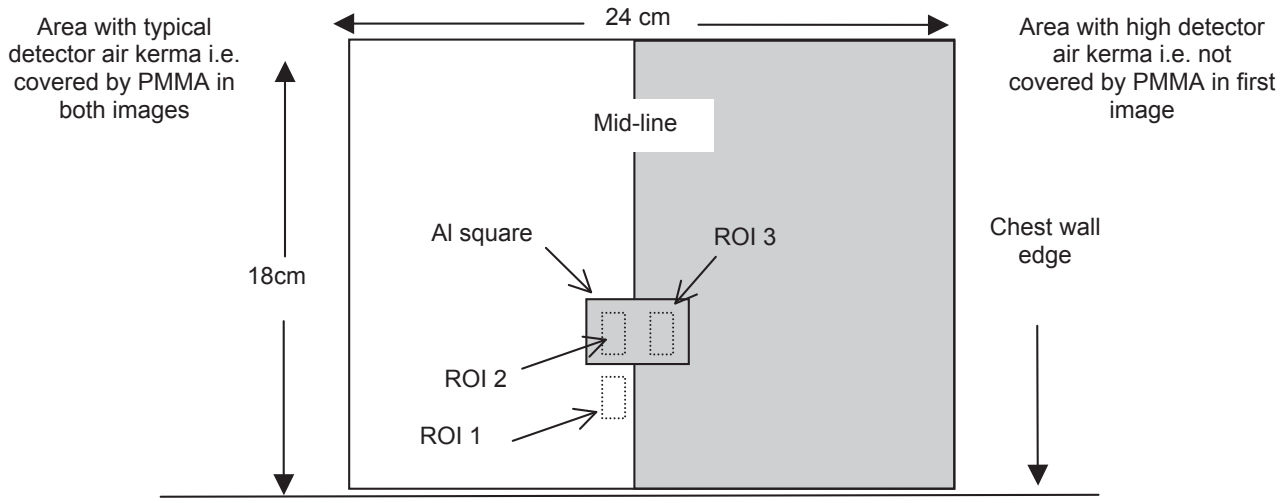


Figure 6 ROI locations for image retention measurement.

For integrated systems acquire pre-processed images. For CR systems, read the IP using the TEST/CONTRAST read program and a SEMI-AUTO read mode. Linearise the data as described in Appendix 3.

Evaluation
Measure the mean pixel value in the ROI on the locations shown in Figure 6 and calculate the image retention factor.

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

Remedial level: image retention factor > 0.3.

Frequency: commissioning and every six months.

3.2.9 Geometric distortion

This test is only applicable to scanning and CR systems, and only if distortion is suspected.

Test protocol
Use the image from section 3.2.7 or place a radio-opaque mesh (eg Leeds fluoroscopy test object M1) on the breast support table and a cassette in the bucky. Expose manually using low exposure parameters (eg Mo/Mo at 28 kV, 10 mAs) and read the IP, using the TEST/CONTRAST read program and a SEMI-AUTO read mode.

Evaluation
Inspect the image for signs of distortion.

Remedial level: any evidence of distortion.

Frequency: commissioning and if problems suspected.

3.3 Automatic exposure control

The majority of the full field digital systems have an AEC in order to control the duration of an exposure as well as select the appropriate kV and target/filter combination for a specific breast thickness. The choice of parameter to use in assessing AEC performance will depend on the manufacturer's design criteria for the AEC performance. It is necessary to be able to verify that the system meets the manufacturer's specifications (eg they may aim to maintain constant *pixel value* with change in absorber thickness). However, verification of this issue is not synonymous with ensuring that the performance of the AEC is optimised.

Use 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 table top so that the front edge is slightly overlapping the chest wall edge of the detector (eg by 5 mm) and ensure that the block is centred left to right in the image field. Use collimation that allows the full field to be imaged.

For integrated systems, acquire pre-processed images. For CR systems, use the TEST/CONTRAST read program and set to a SEMI-X 2 read mode. Read each plate after a consistent time period has elapsed after irradiation (so that consistent decay of the signal occurs on all plates). The same IP must be used throughout each test. Linearise the data as described in Appendix 3.

The AEC detector should be positioned at the chest wall on the midline if possible. All tables and modes, eg magnification, that are used clinically should be tested.

3.3.1 AEC repeatability

Test protocol
Place 45 mm thickness of PMMA on the table top and compress to a standard force, eg 50–100 N. (Note that the compression force may affect the factors selected on some systems.)

Make five exposures at the recommended clinical setting, using AEC. Record the delivered mAs for each exposure.

Evaluation
Calculate the maximum deviation from the mean for the mAs.

Remedial level: maximum deviation in mAs from mean > 5%.

Suspension level: maximum deviation in mAs from mean > 10%.

Frequency: commissioning and every six 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 × 10 mm square under or on top of the PMMA, ideally as shown in Figure 7. 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, eg 50–100 N. (Note that the compression force may affect the factors selected on some systems.) Expose and record the selected expo-

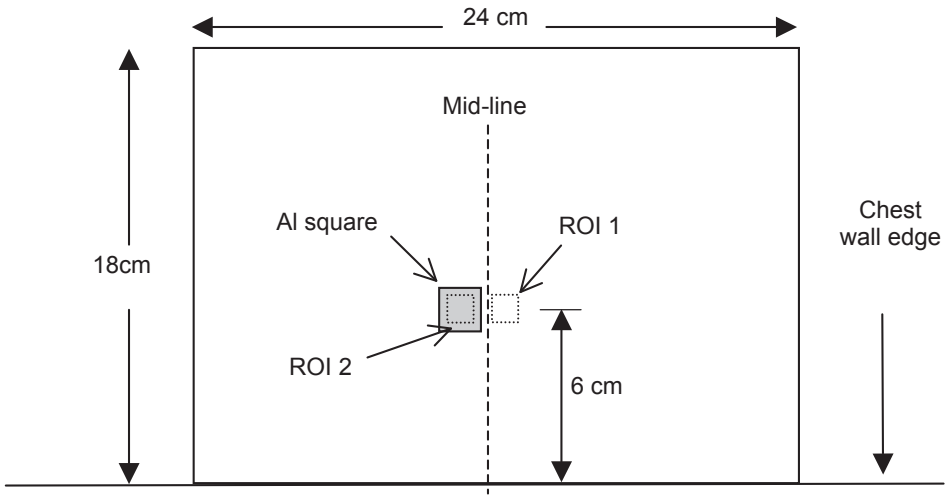


Figure 7 ROI locations for CNR measurement.

sure parameters (eg filter, target, kV and delivered mAs). If the system provides an estimate of mean glandular dose, record the value displayed. For CR systems, additionally record the DDI. Repeat with other thicknesses of PMMA (20–70 mm) and for other automatic modes, keeping the aluminium in the same position. 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 expanded polystyrene blocks to the PMMA as a spacer to make up a total thickness equal to the equivalent breast (see Table A6.1 in Appendix 6).

These exposures can also be used to measure the MGD, as described in section 3.6.1.

Evaluation
Measure the average *pixel value* and standard deviation in ROI 1 (m_1 and σ_1 respectively) on the pre-processed image and in the area of the aluminium square in ROI 2 (m_2 and σ_2 respectively) as shown in Figure 7. Before ROI measurements are used to calculate CNR, the data should be linearised with respect to the energy absorbed by the detector. This is a necessary step for data measured from CR systems. The procedure for linearising the data is described in Appendix 3. Calculate CNR as:

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

The CNR data may also be used to assess whether performance has been optimised.

The measurements obtained at commissioning should be used as baselines for subsequent routine tests.

Remedial level: change from baseline CNR for any thickness > 10%.

Frequency: commissioning and every six months. Repeat for fine focus at commissioning and six-monthly if this mode is used clinically.

3.3.3 AEC variation with density control setting

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

Test protocol
Set target and filter to Mo/Mo, kV to 28 with 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 (eg 10–15%) in mAs per step.

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

Frequency: commissioning.

3.3.4 AEC variation with position of detector

This test is applicable only to units in which 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 position of the AEC 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 8), TG18-LN12 and/or SMPTE test patterns (Figure 9) 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 10) or the clinical reference image from the archive onto the local hard disk.

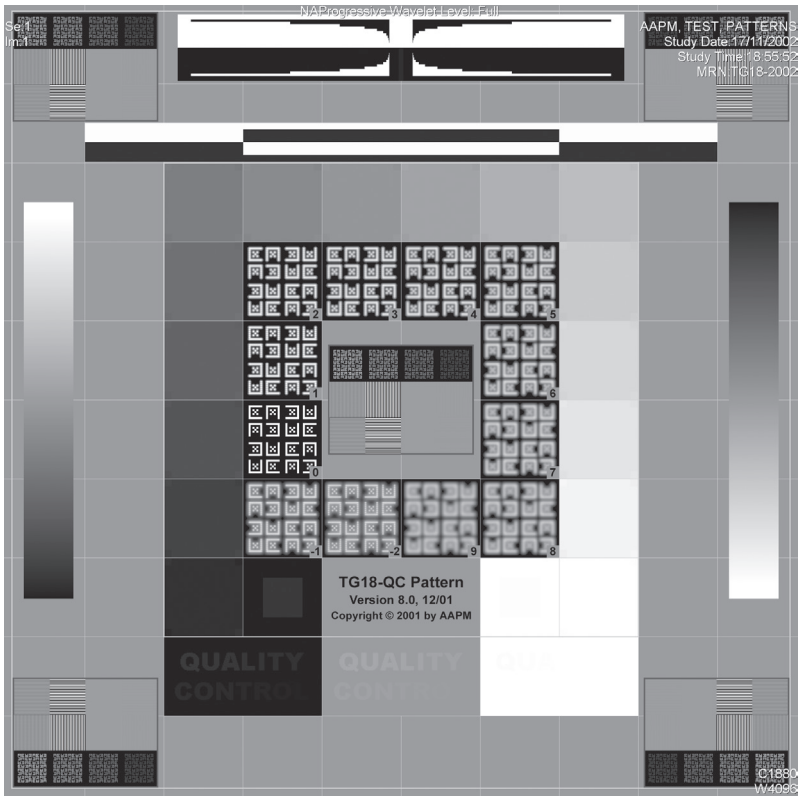


Figure 8 TG18-QC test pattern.

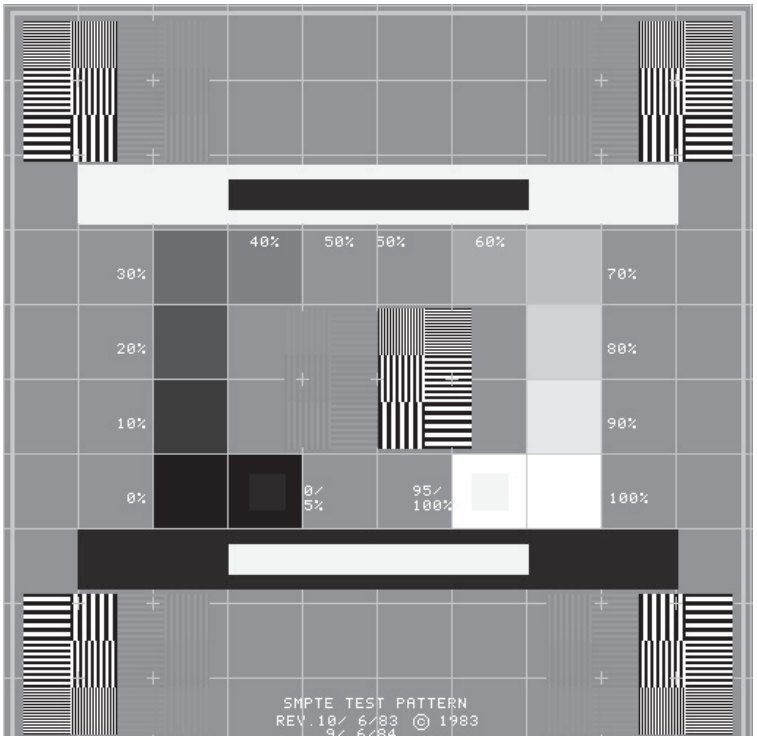


Figure 9 SMPTE test pattern.

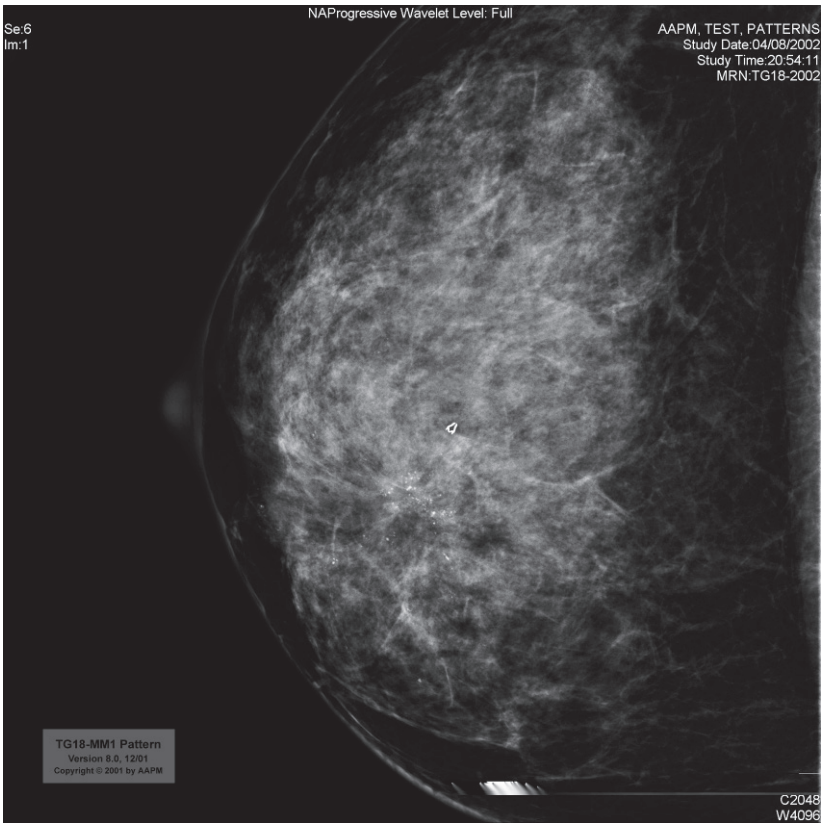


Figure 10 TG18-MM anatomical image.

A suitable photometer with a narrow acceptance angle 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.

All monitors should be tested at commissioning. Tests on primary monitors 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.

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 monitors at the centre of the display with the appropriate detector facing the monitors (inwards). Record the readings. Display the TG-18MM images or a pair of mammograms from the local database and examine the images for any reflections, eg of room lights, windows, self.

Remedial level:

- $L_{amb} > 0.4 \times L_{min}'$, where $L_{min}' = L_{min} + L_{amb}$ (see section 3.4.1.2)
- illuminance > 10 lux
- any disturbing reflections visible.

Frequency: commissioning and every six months.

3.4.1.2 Luminance response

Test protocol

Display the test pattern (TG18-LN12, TG18-QC or SMPTE) using the default window settings. Measure the luminance of all the greyscale steps. If not using TG18-LN12, 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 measured in section 3.4.1.1 to all the above measurements.

Calculate the luminance ratio: 100% greyscale/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. Note that the remedial level for primary monitors is difficult to meet in practice and allowance may need to be made for measurement accuracy.

Remedial level:

The small contrast steps not visible

CRT primary monitor

minimum luminance (L_{min})	> 1 cd/m^2
maximum luminance (L_{max})	< 240 cd/m^2
	baseline ± 10%
	± 5% between paired monitors
luminance ratio	< 250
	baseline ± 10%
contrast response	luminance outside *DICOM* standard ± 10%

LCD primary monitor

minimum luminance (L_{min})	> 1.5 cd/m^2
maximum luminance (L_{max})	< 450 cd/m^2
	baseline ± 10%
	± 5% between paired monitors
luminance ratio	< 300
	baseline ± 10%
contrast response	luminance outside *DICOM* standard ± 10%

Secondary monitor

minimum luminance (L_{min})	> 1 cd/m^2
maximum luminance (L_{max})	< 200 cd/m^2
	baseline ± 10%
luminance ratio	< 100
	baseline ± 10%
contrast response	luminance outside *DICOM* standard ± 20%

Frequency:

- primary monitor: commissioning and every six months (CRT) or annually (LCD)
- secondary monitor: commissioning.

3.4.1.3 Luminance uniformity

Test protocol

Display the test pattern (TG18-LN12, TG18-QC or SMPTE) using the default window settings. 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. (Note: the TG18-UN or TG18-UNL patterns are designed for uniformity measurements and should be used if available.) Measure the luminance at the centre of the image and at the four corners.

Evaluation

Calculate the percentage difference in luminance between the central position and each corner in turn.

Remedial level: maximum percentage difference in luminance > 30%.

Frequency: commissioning and every six months (CRT) or annually (LCD).

3.4.1.4 Monitor resolution

Test protocol

Display the test pattern (TG18-QC or SMPTE) 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 every six months (CRT) or annually (LCD).

3.4.1.5 Geometric distortion

Test protocol

Display the test pattern (TG18-QC or SMPTE) using the default window settings. Inspect the grid lines in the test pattern for any apparent distortion. Note: geometric distortion will only be present in CRT monitors.

Remedial level: any apparent distortion.

Frequency: commissioning and every six months (CRT).

3.4.1.6 Display artefacts

Test protocol

Display the test pattern (TG18-QC or SMPTE) 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 jitter.

Remedial level: any disturbing artefacts visible.

Frequency: commissioning and every six months (CRT) or annually (LCD).

3.4.1.7 Overall imaging performance

Test protocol

Display the clinical image (eg TG18-MM) and search for the known subtle details.

Remedial level: the subtle details are not visible.

Frequency: commissioning and every six months (CRT) or annually (LCD).

3.4.2 Printers

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

In order to test the performance of the hard copy devices, the TG18-PQC or the printer's own test pattern can be used together with TG18-QC or the SMPTE pattern. The TG18-PQC test pattern (Figure 11) must be available in the image archive.

3.4.2.1 Optical density response

Test protocol

Print the TG18-PQC or the printer's test pattern. Measure the optical densities on each bar. Using the protocol spreadsheet (see Appendix 4), select the appropriate test pattern and enter the measured values to obtain the contrast response and compare against the *DICOM* standard. Note that the remedial level is difficult to meet in practice and allowance may need to be made for measurement accuracy.

Remedial level: optical densities outside *DICOM* standard ± 10%.

Frequency: commissioning and every six months.

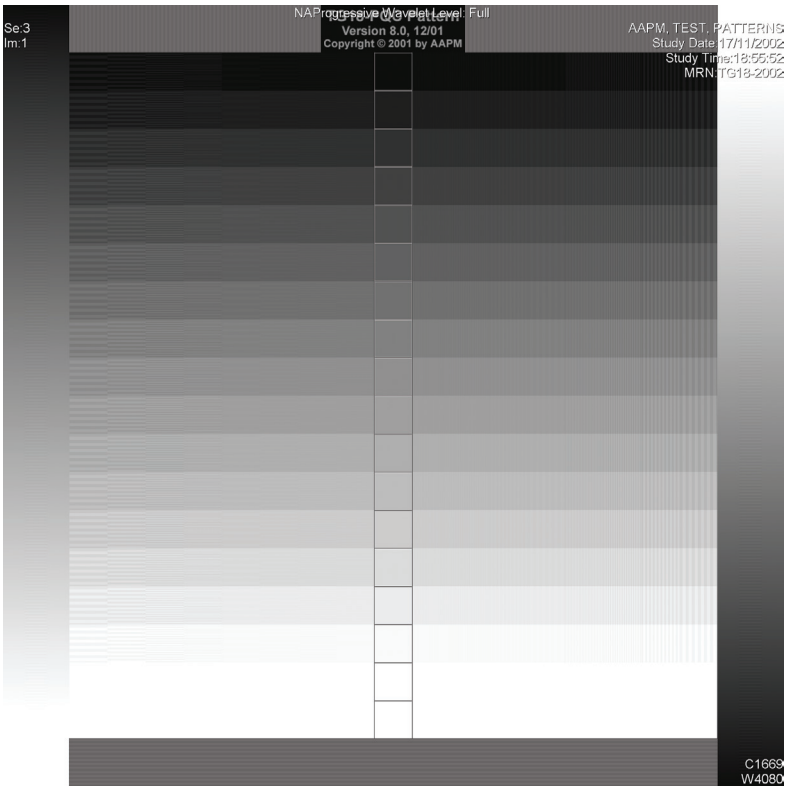


Figure 11 TG18-PQC test pattern.

3.4.2.2 Contrast visibility

Test protocol

Print the SMPTE or TG18-QC test pattern using the default window settings.

Inspect the small contrast steps (5% in 0% level, 95% in 100% level). Measure the optical densities of all the greyscale steps.

Remedial level:

- the small contrast steps are not clearly visible
- optical densities outside baseline ± 10%.

Frequency: commissioning and every six months.

3.4.2.3 Density uniformity

Test protocol

Display a uniform image (eg from uniformity tests). Print the image and measure the optical density at the centre of the image and at the four corners.

Evaluation

Calculate the percentage difference in optical density between the central position and each corner in turn.

Remedial level: maximum percentage difference in optical density > 10%.

Frequency: commissioning and every six months.

3.4.2.4 Artefacts

Test protocol

Use the film printed for the uniformity test. Inspect the image for printer artefacts such as streaking, banding, pick-off, jitter, etc.

Remedial level: any disturbing artefacts visible.

Frequency: commissioning and every six months.

3.4.2.5 Distortion

Test protocol

Use the film printed for the contrast visibility test. Examine the grid lines on the printed pattern for any signs of geometrical distortion.

Remedial level: any apparent distortion.

Frequency: commissioning and every six months.

3.4.2.6 Printer resolution

Test protocol

Use the film printed for the contrast visibility test to check the resolution of the printer. 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 on the film. It should be sharp and clear.

Remedial level: any loss in resolution.

Frequency: commissioning and every six months

3.4.2.7 Consistency between network printers

Test protocol

Load a clinical image on a reporting (primary) workstation and print it to all the printers that are used clinically. Visually compare the quality of the soft copy image with that of the printed images.

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

Frequency: commissioning.

3.5 Image quality

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 as was used to develop the standards in the European protocol.[15] 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.

3.5.1 Detail detection

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 0.1–2 mm. The test object should be used with 2 cm thick plates of PMMA above and below. This has a physical thickness of approximately 4.5 cm. However, the total attenuation of this combination is approximately equivalent to a 5 cm thickness of PMMA, which in turn is equivalent to a typical breast thickness of 6 cm. Use the exposure factors that would be selected clinically for a 6 cm thick breast. If the automatic selection depends on the compressed breast thickness, 15 mm thick spacers may be added at the corners or edges of the test object – so as not to interfere with the area covered by the gold discs. Expanded polystyrene blocks have been found to be convenient for this purpose. Repeat the exposures until four pre-processed images have been recorded. (Consult the manufacturer for the appropriate algorithms for obtaining linear unprocessed images of test objects. For CR systems, use the TEST/CONTRAST read program and a SEMI-X2 read mode.) It is desirable to vary the acquisition conditions slightly by moving the test object by a few millimetres between exposures. For CR systems, use four different cassettes.

Effective reading of CDMAM images requires some training and practice until consistent results are obtained. Experience has shown that the average threshold gold thickness for three observers reading four images has a 95% confidence interval of about 6%. However, this number of readers and images may not always be practical, and a minimum of two observers should each score two sample images of the test phantom and determine the average threshold gold thickness. Alternatively, one observer should determine the thresholds from four sample images of the test phantom and average the results. This should be repeated for all the display workstations and hard copy devices used clinically. The window width and level should be adjusted to maximise the visibility of the details on the displayed and printed images. The small detail sizes are particularly important (0.1–0.25 mm) as it is these that some digital systems may have difficulty in detecting and displaying. When appropriate, a zoom facility should be used to examine small details. It may be necessary to assess the images with and without zooming to establish what would be acceptable practice.

The images of the CDMAM test object should be scored according to the procedures described in the manufacturer's manual. This involves identifying which corners contain gold discs and applying a correction algorithm to determine the threshold gold thickness at each detail diameter. It is important that each reader continues to make his or her best guess of the location of the corner discs for at least two squares beyond the point at which he or she is confident of the true location.

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:

$$T_c = a + bx^{-1} + cx^{-2} + dx^{-3}$$

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, ie 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 6 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 Nijmegen CDMAM contrast–detail phantom version 3.4. However, it is intended that they are sufficiently flexible 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 0.1–2 mm. However, precise measurements are best made by determining the threshold gold thickness for a larger number of detail diameters (eg 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 Table 6. This procedure helps to reduce the effect of image noise and random observer errors, noticeable when just a few details are assessed.

Experience has shown that interobserver errors can be a significant problem in measuring threshold gold thicknesses. It is expected that some means of standardising this measurement will be developed, eg automatic image scoring software. Where a system appears to fail, the possibility of interobserver error should be considered, and further measurements by other experienced observers undertaken.

Remedial level: Table 6 shows acceptable and achievable levels published in the European protocol.[5] The 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.

Frequency: commissioning.

Table 6 Acceptable and achievable levels of detail detection

	Threshold gold thickness (μm)	
Diameter of detail (mm)	**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

3.5.2 Regular image quality tests

A full and accurate assessment of detail detection requires test objects such as the Nijmegen CDMAM, which may not be widely available, and may be excessively time-consuming to conduct on a frequent basis. It is therefore proposed at present to use as an acceptable alternative the existing and simpler test objects such as the TORMAX and TORMAM for more frequent tests.

Image quality measurements must be made to establish a baseline on commissioning new equipment and whenever there are major changes in the system. While the TORMAX test object may be used, it is strongly recommended to check image quality using a second and possibly more sensitive test object such as the TORMAM. These tests should then be repeated at six-monthly intervals.

Routine testing of image quality may also be a part of the routine quality control procedures by the operators of x-ray sets. This is a requirement in the NHSBSP on a weekly basis.[16] The test images may be read locally by radiography staff and/or may be submitted for review as part of regional quality control systems. The image should be compared with any quantitative information and with previous images and data. Any deterioration in image quality will necessitate further investigation.

Test protocol
The TORMAX test object should be placed on top of 4 cm of PMMA and imaged using the exposure factors typical of those used clinically. Where available this should be done under AEC. (Consult the manufacturer for the appropriate algorithms for obtaining linear pre-processed images of test objects. For CR systems, use the TEST/CONTRAST read program and a SEMI-X2 read mode.)

The TORMAM test object should be placed on top of 3 cm of PMMA and imaged using the exposure factors typical of those used clinically. Where available, this should be done under AEC. Since this test object has features similar to a breast, an algorithm suitable for a breast must be used along with the image processing normally applied to clinical images.

Evaluation
The image should be read using the normal reporting display media and the details scored and recorded. If image quality appears to have changed significantly since the baseline measurements, the detail detection procedure using the CDMAM test object described above should be repeated.

Remedial level:
- TORMAX: number of details detected fails to meet NHSBSP standards for film-screen systems or is significantly less than baseline measurement
- TORMAM: visibility of details is significantly inferior to baseline.

Frequency: every six months.

3.6 Dose

The two dose issues of concern are the risk and image quality implications of each exposure. The risk can be related to the mean glandular dose, while the dose to the receptor has image quality implications. The following procedures are described for dose measurement.

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,[1,10] as listed in Table A6.1 in Appendix 6.

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 expanded polystyrene blocks to the PMMA as a spacer to make up a total thickness equal to the equivalent breast (see Table A6.1 in Appendix 6). A standard compression force should be applied (eg 50–100 N).

Measurements should be made using PMMA blocks with thicknesses of 20–70 mm on commissioning and when optimising the system. 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:

$$D = K_P g_T c_T s_T$$

where K_P is the incident air kerma (without backscatter) calculated at the upper surface of the PMMA. The factor g_T corresponds to a breast with a glandularity of 50% and is derived from the values calculated by Dance et al.[10] The factor c_T corrects for the difference in composition of typical breasts from 50% glandularity.[10] Table A6.1 in Appendix 6 gives the equivalent breast thickness and the product of g and c for a range of PMMA thickness and half-value layer (HVL) for typical breasts in the age range 50–64. 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. Typical values of HVL for various spectra are given in Table A6.2. The factor s shown in Table A6.3 corrects for any difference due to the choice of x-ray spectrum.[10] W/Al spectra filtered by 0.5 mm aluminium (as used by Sectra mammography systems) have a softer component than many mammographic spectra and, because there is no K-edge filter, can also have a more penetrating component. As a consequence, the approximation of a single s-factor for a given target/filter combination would give rise to unacceptably large systematic errors. Instead, a table of s-factors as a function of breast thickness is provided (Table A6.4).

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 7
- displayed values of MGD> 30% different from calculated values
- change in MGD to standard breast from commissioning value > 25%.

Table 7 Dose remedial levels

Thickness of PMMA (cm)	Thickness of equivalent breast (cm)	Remedial level for mean glandular dose to equivalent breasts (mGy)
2.0	2.1	> 1.0
3.0	3.2	> 1.5
4.0	4.5	> 2.0
4.5	5.3	> 2.5
5.0	6.0	> 3.0
6.0	7.5	> 4.5
7.0	9.0	> 6.5

Frequency:
- MGD to the standard breast (4.5 cm PMMA): commissioning and every six months
- MGD at other breast thickness: commissioning and when the AEC software is changed.

3.6.2 Clinical breast doses

Test protocol
It is recommended that the MGDs for a series of about 50 breast examinations are periodically measured on each mammographic system using the procedures described in IPEM Report 89.[1] Software for making such dose calculations has been published by the NHSBSP.[17] 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 10 women should be included in the dose sample.

Evaluation
The current national DRL for this dose audit measure is 3.5 mGy.[18] Corrective action should be taken where exceeding the DRL cannot be clinically justified. Lower 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 > 3.5 mGy
- dose audit measure significantly > local DRL.

Frequency: commissioning, and at least every three years.

APPENDIX 1: SUMMARY OF TESTS

Where given, the limiting values are remedial levels unless otherwise indicated. All tests are performed at commissioning.

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	Six monthly	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; CR – six monthly	> 5 mm between edge image and edge breast platform
3.2 Detector performance	3.2.1.1	CR dark noise	Annually	Mean pixel value > 450; SD > 1
	3.2.1.2	CR reader sensitivity	Six monthly	S_{cal} > 120 ± 20%
	3.2.1.3	CR IP matching and artefacts	Commissioning	Maximum variation DDI > 5%; maximum variation mAs > 5%
	3.2.1.4	CR laser function	Annually	Any laser jitter or dropout
	3.2.2	Electronic calliper	Commissioning	Error > 2%
	3.2.3	Uniformity	Six monthly	Integrated detector: maximum deviation > 10% CR system: centre–side > 10%; left–right > 10%
	3.2.4	Artefacts	Six monthly	Any artefacts that may affect clinical image quality
	3.2.5	Detector response	Six monthly	Reference air kerma > 20% change; standard deviation > 10% change; SNR > 10% change
	3.2.6	Resolution	Six monthly	> 10% change in SWCTF(*f*); limiting resolution < 75% of baseline
	3.2.7	Spatial discontinuity and resolution homogeneity	Six monthly	Discontinuities or blurring
	3.2.8	Image retention	Six monthly	Image retention factor > 0.3
	3.2.9	Geometric distortion	As required	Any distortion
3.3 Automatic exposure control	3.3.1	Repeatability	Six monthly	Remedial: maximum deviation from mean mAs > 5% Suspension: maximum deviation from mean mAs > 10%

	3.3.2	Variation with absorber thickness – automatic mode	Six monthly	CNR change from baseline > 10%
	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	Six monthly	$L_{amb} > 0.4 \times L_{min}'$; illuminance > 10 lux; any reflections
	3.4.1.2	Luminance response	CRT primary monitor – six monthly	Minimum > 1 cd/m^2; maximum < 240 cd/m^2; luminance ratio < 250; luminance variation from *DICOM* standard > 10%
			LCD primary monitor – annually	Minimum > 1.5 cd/m^2; maximum < 450 cd/m^2; luminance ratio < 300; luminance variation from *DICOM* standard > 10%
			Secondary monitor – commissioning only	Minimum > 1 cd/m^2; maximum < 200 cd/m^2; luminance ratio < 100; luminance variation from *DICOM* standard > 20%
	3.4.1.3	Luminance uniformity	CRT – six monthly; LCD – annually	Maximum variation > 30%
	3.4.1.4	Resolution	CRT – six monthly; LCD – annually	Any loss in resolution
	3.4.1.5	Geometric distortion	CRT – six monthly	Any distortion
	3.4.1.6	Artefacts	CRT – six monthly; LCD – annually	Any artefacts
	3.4.1.7	Overall imaging performance	CRT – six monthly; LCD – annually	Subtle details not visible
3.4.2 Image display – printers	3.4.2.1	Optical density response	Six monthly	OD variation from *DICOM* standard > 10%
	3.4.2.2	Contrast visibility	Six monthly	Small contrast steps not clearly visible OD outside baseline ± 10%
	3.4.2.3	Density uniformity	Six monthly	Max variation > 10%
	3.4.2.4	Artefacts	Six monthly	Any artefacts
	3.4.2.5	Geometric distortion	Six monthly	Any distortion
	3.4.2.6	Resolution	Six monthly	Any loss in resolution
	3.4.2.7	Consistency between network printers	Commissioning	Significant differences in image quality

3.5 Image quality	3.5.1	Detail detection	Commissioning	See Table 6
	3.5.2	Regular image quality tests	Six monthly	Significant changes from baseline
3.6 Dose	3.6.1	Dose vs thickness	Commissioning	See Table 7; displayed values >30% different from calculated values
	3.6.1	Dose to the standard breast	Six monthly	> 2.5 mGy; > 25% change from commissioning value
	3.6.2	Clinical breast doses	One to three yearly	Dose audit measure > 3.5 mGy; dose audit measure significantly > local DRL

APPENDIX 2: FUJI CR READER MODES

The following briefly describes the reader modes for Fuji CR systems and gives guidance on the appropriate selection of reader modes for performance testing.

The Fuji CR system sets two main parameters in IP reading to condition the intensity data from the IP so that they are suitable for image presentation. These parameters are the *S*-value, or sensitivity value, and the latitude. The value of these parameters will depend on both the intensity and range of exposures the IP has received and the system read mode. The *S*-value basically determines the median intensity associated with the read mode and the latitude determines the range of intensities which can be read from the IP. Fuji CR systems use the following read modes:

- AUTO. This mode allows the image data on the IP to be presented automatically with suitable contrast and density. It is the read mode used for most clinical examinations; however, it is unsuitable for performance testing as it may produce unpredictable results. In AUTO mode the sensitivity and latitude of the reader are automatically adjusted.
- SEMI-AUTO. This mode automatically adjusts the sensitivity of the reader based on exposure intensity in a defined region of the IP. This is usually a central region of the IP. The SEMI-AUTO mode uses a fixed reading latitude.
- SEMI-X. This mode operates similarly to the SEMI-AUTO mode except that the region over which the intensity data are sampled can be chosen. This mode is useful for quality control testing in mammography as a region at the chest wall edge (SEMI-X 2) can be selected.
- FIX. This mode operates similarly to film in that the sensitivity and latitude are fixed.

In AUTO and SEMI-AUTO read modes the *S*-value will be inversely proportional to the incident exposure to the IP. The *S*-value has a range from 2 to 20 000.

The latitude value is $\log_{10}$ of the range of intensities which can be used, eg a latitude of 2.0 represents a 100-fold range of intensities.

Useful programs to use for testing from the TEST/QC menu are the SENSITIVITY and CONTRAST programs. They should be set up by the Fuji applications specialist at installation as follows:

- GT = A (gradation type A, ie linear relationship between final *pixel value* and original *pixel value*)
- GA = 1.0 (slope of line relating final *pixel value* to original *pixel value* = 1)
- GS = 0.00 (no shift of line relating final *pixel value* to original *pixel value*)
- FNC OFF (no 'Flexible Noise Control' – a proprietary type of spatial frequency processing)
- RE = 0.0 (no frequency enhancement).

The SENSITIVITY program should be set up with a latitude of 1.0 (ie a 10-fold range of intensities) and the CONTRAST program should be set up with a latitude of 2.0 (ie a 100-fold range of intensities).

These settings will produce images with linear look-up tables and no spatial frequency processing as required by the NHSBSP test protocols. The program settings should be checked before commencing testing. If the values are not correct, it is possible to adjust them via the user interface before each image is saved, but this is time-consuming and has the potential for error.

APPENDIX 3: 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 relationship between the detector entrance air kerma and the pixel value in pre-processed images which was measured in section 3.2.4 – the signal transfer property (STP). 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 many CR systems (including Fuji, Kodak and Konica), the STP is logarithmic:

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

where P is the pre-processed *pixel value* corresponding to a detector entrance air kerma of K, and a and 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.

Using the method described by MacKenzie,[19] 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 3.3.2 is given by:

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

Using (2) and (4) this becomes:

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

For a system with a linear STP (eg many of the integrated detector systems):

$$P = aK + b \tag{8}$$

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 pre-processed pixel values.

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

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

Thus, the STP coefficients cancel out and the CNR can be calculated simply from the pre-processed pixel values.

The method can be generalised to any system by using the appropriate STP equation.

APPENDIX 4: SPREADSHEET FOR THE CALCULATION OF THE COMPLIANCE OF A DISPLAY DEVICE TO THE DICOM 3.14 GSDF

The spreadsheet can be downloaded from the NHS Cancer Screening Programmes website. 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). A similar spreadsheet is available for testing printer *DICOM* GSDF conformance (section 3.4.2.1).

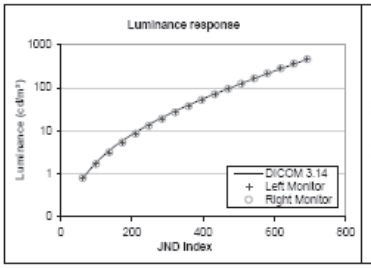
AAPM TG18 test patterns can be downloaded from: http://www.aapm.org/pubs/reports/OR_03_Supplemental/ (TG18-LN in lumin-1k-dcm.zip, TG18-QC and TG18-PQC in multi-2k-dcm.zip).

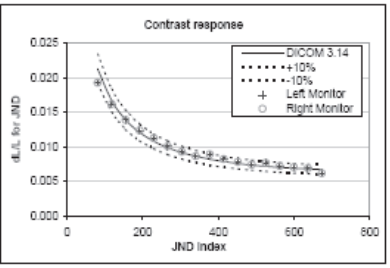
NHSBSP - test of monitor conformance with DICOM 3.14 GSDF

Test details:

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

Test pattern		Luminance (cd/m²)		JND	dL/L	dL/L for JND	average JND	abs dev from DICOM
TG18-LN	p-value	Excl amb	Incl amb	j(L)				
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 JND	average JND	abs dev from DICOM
TG18-LN	p-value	Excl amb	Incl amb	j(L)				
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: CALCULATION OF CONTRAST FOR DETAILS IN THE CDMAM TEST OBJECT

The minimum and achievable standards in section 3.5 are specified in terms of threshold gold thicknesses for a range of detail diameters. In the European protocol,[5] the standards are also specified in terms of the nominal contrast for the gold discs when a standard spectrum is used (using a tube voltage of 28 kV, a molybdenum target material and a 30 µm thick molybdenum filter). This is so that different designs of test object could be employed to achieve a similar result. The nominal contrasts for different thicknesses of gold were calculated as follows. The contrast of a detail is defined as:

$$C = \frac{|I - I_D|}{I}$$

where I is the primary transmission through the phantom material and I_D is the primary transmission through the contrasting detail and phantom. The quantity transmitted, I, is energy fluence, defined as:

$$\int_0^{kVp} N(E).E.dE$$

where E is the x-ray energy and $N(E)$ is the number of x-ray photons at the given energy E.

The nominal contrasts for the minimum and achievable standards in the European protocol were calculated using a spectrum derived from IPEM Report 78.[20] The contrast of the discs and the threshold limiting values were determined using the CDMAM phantom with a 2 cm thickness of PMMA above and 2 cm thickness below the test object. The CDMAM phantom includes an aluminium and PMMA base which is approximately equivalent to 1 cm of PMMA in terms of attenuation. The calculated contrasts for various thicknesses of gold using 28 kV Mo/Mo are shown in Table A5.1.

The contrast specified is that produced by the primary beam and the effect of scatter is not included in the calculation. The contrast of other thickness of gold can be interpolated using the following equation

$$C = aT_g - bT_g^2$$

where C is the nominal contrast (%) at 28 kVp Mo/Mo of a gold disc with a thickness of T_g in µm. For a CDMAM test object with 4 cm PMMA, a = 15.73 and b = 1.180.

Table A5.2 shows the threshold gold thickness and corresponding nominal contrasts used in the European protocol.

Table A5.1 Calculated radiation contrast for various gold thicknesses using a 28 kV Mo/Mo spectrum

Thickness of gold (µm)	Radiation contrast (%) for CDMAM with 4 cm PMMA
0.1	1.57
0.5	7.60
1.0	14.55
1.5	20.92
2.0	26.76

Table A5.2 Acceptable and achievable levels of threshold contrast

	Threshold contrast			
	Acceptable value		Achievable value	
Diameter of detail (mm)	Gold thickness (µm)	Radiation contrast using Mo/Mo 28 kV (%)	Gold thickness (µm)	Radiation contrast using Mo/Mo 28 kV (%)
5*	0.056	<0.85	0.032	<0.45
2	0.069	<1.05	0.038	<0.55
1	0.091	<1.40	0.056	<0.85
0.5	0.150	<2.35	0.103	<1.60
0.25	0.352	<5.45	0.244	<3.80
0.1	1.68	<23.0	1.10	<15.8

*This diameter size is optional and is not included in version 3.4 of the CDMAM test object.

APPENDIX 6: DATA FOR CALCULATION OF BREAST DOSE

Table A6.1 Product of *g* and *c* factors for breasts simulated with PMMA

PMMA thickness (mm)	Equivalent breast thickness (mm)	Glandularity of equivalent breast	Product of *g* and *c* factors HVL (mm aluminium)						
			0.30	0.35	0.40	0.45	0.50	0.55	0.60
20	21	97	0.336	0.377	0.415	0.450	0.482	0.513	0.539
30	32	67	0.245	0.277	0.308	0.338	0.368	0.399	0.427
40	45	41	0.191	0.217	0.241	0.268	0.296	0.322	0.351
45	53	29	0.172	0.196	0.218	0.242	0.269	0.297	0.321
50	60	20	0.157	0.179	0.198	0.221	0.245	0.269	0.296
60	75	9	0.133	0.151	0.168	0.187	0.203	0.230	0.253
70	90	4	0.112	0.127	0.142	0.157	0.173	0.194	0.215
80	103	3	0.097	0.110	0.124	0.136	0.150	0.169	0.188

Table A6.2 Typical HVL measurements for different tube voltage and target/filter combinations (data include the effect on measured HVL of attenuation by a compression plate)

	HVL (mm aluminium) for target/filter combination			
kV	Mo/Mo	Mo/Rh	Rh/Rh	W/Rh
25	0.33 ± 0.02	0.40 ± 0.02	0.38 ± 0.02	0.52 ± 0.03
28	0.36 ± 0.02	0.42 ± 0.02	0.43 ± 0.02	0.54 ± 0.03
31	0.39 ± 0.02	0.44 ± 0.02	0.48 ± 0.02	0.56 ± 0.03

Table A6.3 *s*-factors for some clinically used spectra[10,21]

Spectrum	*s*-factor
Mo/Mo	1.000
Mo/Rh	1.017
Rh/Rh	1.061
Rh/Al	1.044
W/Rh	1.042
W/Ag	1.042

Table A6.4 *s*-factors for W/Al spectra filtered by 0.5 mm aluminium. Such spectra are used by Sectra mammography systems and the values have been calculated[21] for the tube voltage range 25–40 kV, depending on breast thickness. These factors are appropriate for the typical glandularities for screened women in the NHSBSP aged 40–49 and 50–64 reported in Dance et al.[10]

PMMA thickness (mm)	Equivalent breast thickness (mm)	*s*-factor
20	21	1.075
30	32	1.104
40	45	1.134
45	53	1.149
50	60	1.160
60	75	1.181
70	90	1.198
80	103	1.208

APPENDIX 7: MEASUREMENT AND USE OF GRID TRANSMISSION FACTOR

The measurement of detector response (section 3.2.5) requires an estimation of the air kerma at the detector input plane. Air kerma can be measured only above the breast platform and antiscatter 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* 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 (eg 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 *Mg* 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*/*Mg*

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.

*For CR systems, the effect of the grid can be removed by simply placing the CR cassette on top of the breast platform rather than in the bucky.

APPENDIX 8: TIPS ON OBTAINING PRE-PROCESSED IMAGE DATA FROM VARIOUS SYSTEMS

Hologic Selenia

Image type:	'FLATFIELD'
How:	When adding additional views to the patient, simply add "FLATFIELD"

Siemens Novation DR

Image type:	'QC RAW'
How:	Register patient and click on 'Start Exam'. Change Views/Image type to 'QC RAW'. The box at the bottom right-hand side of the Syngo AWS screen contains the thumbnails of the mammography views that are to be done for the patient. Right click to add a new image. Generally add RCC to keep chest wall position consistent.

GE Senographe DS

Image type:	'RAW'
How:	On the Browser page that shows completed exams for the patients (ie not the *DICOM* schedule page), go to the Tools Menu box at the top right-hand side of the page ('Hammer and Saw' icon). Click on this and select 'Medical Application Preferences' from the drop-down menu list. Click on the 'Image Process' Box at the top right-hand side of the page. Select Disable for both FineView and PremiumView. Remember to Enable these settings at the end of the QA.

Fuji CR

See Appendix 2.

GLOSSARY

Aliasing

A phenomenon that arises in sampling when the frequencies present in the signal to be sampled are higher than can be represented by the sampling process (ie 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 $(2f_N - f)$.

Charge coupled device (CCD)

An integrated circuit that acts as an array of photodiodes converting light to electrical charge. The charge is collected in a series of storage wells, which can be read out sequentially after an exposure.

Computed radiography (CR)

Generic name for imaging technology that uses photostimulable storage phosphor plates for image acquisition.

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 the *CCD* elements due to thermal energy. This generation of charge gives rise to a dark current whose magnitude will depend on the integration time of the *CCD* and also, critically, on the temperature. Dark current can be compensated for by removing its offset signal. However, the noise or random variability associated with this signal cannot be removed. Dark current and hence dark noise may be reduced by means such as cooling the sensor device.

Detector dose indicator (DDI)

A numerical figure produced usually by CR read unit indicating the level of dose to the CR IP.

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 antiscatter 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 trans-

	fer function (MTF) of the imaging system and is defined as the point at which 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, ie 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 that makes up a digital image. It has a spatial dimension and is assigned a discrete intensity value.
Pixel size	The distance between sampling points in a detector. Pixel size is distinct from the active area of the detector, which is the size of the light sensitive element. For a *CCD* the active area may be smaller than the distance between the element centres (sampling points).
Pixel value	A digital value that represents the greyscale level assigned to a *pixel*.
Pre-processed 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.
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 4 cm from the chest wall edge and on the midline.
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.

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