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Type Testing of CT Scanners: Methods and Methodology for Assessing Imaging Performance and Dosimetry

ImPACT

MDA Evaluation Report

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ImPACT

Imaging Performance Assessment of CT Scanners

Type Testing of CT Scanners : Methods and Methodology for Assessing Imaging Performance and Dosimetry

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Please note this report is referenced as MDA/95/43 in previous Blue Cover Reports of Individual Scanner Evaluations, as well as in some other documents.

Acknowledgements

Many of the tests, and associated comments, reported on in this document are referenced where they have been developed from written documents or reports. However we wish to further acknowledge the assistance of personnel from the CT scanner manufacturers, as well as from hospital physicists in the field, all of whom have contributed in many ways in increasing our understanding of scanners, scanner technology and methods of scanner evaluation.

We would like to express our appreciation of comments made by Dr. P.C. Shrimpton (National Radiological Protection Board).

We are also grateful for the copyright permission granted from the American Association of Physicists in Medicine and Advanced Medical Publishing for the reproduction of data and a figure from the book Medical CT and Ultrasound: Current Technology and Applications AAPM 1995 Summer School June, Editors Goldman L.W. and Fowlkes J.B, (Advanced Medical Publishing) (1995). See Appendix 4.

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Introduction

■ Type Testing

The ImPACT (Imaging Performance Assessment of CT Scanners) group is funded by the Medical Devices Agency to undertake the type testing of CT Scanners. ImPACT is a group within the Medical Physics Department at St. George's Hospital, London, which is part of St. George's Healthcare Trust.

Type testing is a procedure whereby one machine, that has been identified by the manufacturer or supplier as operating within specification, is assessed and the results taken to be representative of that model. Type tests are carried out with the active participation of the manufacturer at a mutually acceptable site and on a current CT system running the latest release of software. This helps to ensure that the performance is within specification and is representative of the current capabilities of the model.

This report describes the techniques and test objects used by ImPACT for these type tests. The different sections cover techniques for the measurement of performance parameters such as resolution, noise and dose, expected variation of performance parameter with scan parameters, such as tube voltage and current, and comparison with manufacturers' data. This report also describes the methodology of the assessments, for the production of clinically relevant dose and image quality information.

This report will be updated periodically to include revised techniques and methodologies.

■ Individual Reports

The technical reports produced by ImPACT contain the results from the performance measurements, a detailed discussion of the scanner, its performance measurements, the comparison with the manufacturers data and the comparative, clinically relevant information.

■ Comparison Reports

The clinically relevant data is presented in the Comparison Reports[1,2] which tabulate and discuss results from several scanners. Comparison data is presented in individual reports in a histogram form.

■ CT Scanners - General

CT Scanners were developed for clinical diagnostic applications in the early 1970's. They provided a tremendous step forward in imaging, particularly of the brain, and whilst initially only head scanners were produced, body scanners were soon developed.

The first and second generation scanners were 'translate-rotate' systems, so called because the tube and detectors first translated across the patient to collect a set of attenuation data at a particular angle, the gantry then rotated around a few degrees. The tube and detectors were translated across the patient again to obtain another set of attenuation data, followed by another rotation, and this continued until a number of sets of data were collected at different angles.

With the introduction of whole body scanners came the evolution to the third and fourth generation scanners. All modern scanners are either third or fourth generation. In third generation scanners, or 'rotate-rotate' scanners, the tube and the detectors both rotate around the patient, with a fan beam covering the whole (or occasionally only part) of the patient. Fourth generation scanners, or 'rotate only', have a ring of stationary detectors and only the tube moves around the patient. (The tube is generally on the inside the detector ring, however there is a design, no longer marketed, where the tube is outside the detector ring for the 'rotate-nutate' version).

This labelling by generation can be a bit misleading, since the fourth generation appeared after the third, but now just represents a different approach. Currently almost all scanners are third generation with only one manufacturer producing fourth generation scanners.

The very early 'translate-rotate' scanners used two banks of detectors to acquire two slices simultaneously. Currently, with one exception, all scanners are constructed with a single bank of detectors. However, this may change in the next few years since manufacturers are beginning to publish work-in-progress on multi-bank systems.

As opposed to the newer slip ring scanners, the older scanners transmit the power to the tube and the detectors through cables, which means that the direction of the tube rotation has to be reversed for each scan. The tube is first accelerated in one direction for the first scan, then decelerated to a halt, reversed in its direction of rotation, and then accelerated ready for the scan in the other direction. This reversal of direction continues with each subsequent scan.

■ CT Scanners - Spiral CT

The largest technological advance in CT since its beginning in the early 1970's has been the introduction of spiral scanning.

In conventional CT the images are acquired from successive incremental slice irradiations. In spiral CT, the tube (and generator on some systems, and detectors on third generation scanners) rotate continuously around the patient as the couch is moved through the gantry aperture at a constant speed. A helical or spiral set of data is thus acquired, and an image at a particular plane is reconstructed from the interpolation of attenuation profiles from either side of the required image plane. In this context the terms spiral, helical, volume and continuous acquisition scanning are synonymous.

One advantage of spiral CT over conventional CT is that after scanning the reconstruction position and spacing of reconstructed images can be chosen retrospectively. Other advantages include the ability to scan a whole region in, for example 30 seconds, as opposed to 30 individual scans each of 1 second with an interval allowing for couch movement and resetting of the tube. This has obvious advantages in the ability to undertake an examination in a single breath hold, leading to the reduction of misregistration and movement artefacts in body imaging.

The technological breakthrough for spiral CT was provided by the introduction of slip rings to the gantry structure. The first slip ring scanners became available in 1987. They were primarily designed for fast dynamic scanning, to obtain a minimum scan to scan time, and were not initially designed for helical or spiral scanning. Slip rings enable electrical power to be transferred to the generator, or tube, and other rotating parts, and enable the data signal to be transferred back from the detectors to the computer system without the use of cables. As a result, the tube (and generator on some systems, and detectors on third generation scanners) can rotate continuously around the patient.

Slip rings are used on both third and fourth generation scanners. They are described as either 'high voltage' or 'low voltage'. The low voltage slip ring systems have the generator fixed onto the rotating gantry, whereas the high voltage slip ring systems do not.

Following the interpolation of projections either side of the required image plane, an image from a spiral data acquisition is reconstructed in the same way as in conventional CT, using the same algorithms and reconstruction filters.

For a performance test, a spiral scanner can in many ways be treated as a conventional scanner. However, the acquisition of images from a spiral data set requires some special considerations which are dealt with in this report.

To perform spiral scanning a scanner needs not only to have slip ring technology, but also to be supplied with the necessary software to deal with the spiral data set. The scanner also needs to be equipped with an X-ray tube which has a high enough heat capacity to operate continuously at clinically acceptable mA values whilst also irradiating for durations of at least 30 seconds.

For these reasons originally only top of the range scanners could perform spiral scanning as standard. The mid-range or bottom of the range scanners, whilst having slip rings, were not necessarily capable of spiral scanning unless the necessary optional software had been bought, or the tube upgraded. However it is now the exception to find a scanner that does not have spiral capability as standard, although the higher range scanners still tend to be equipped with the higher power tubes and greater flexibility in dealing with the spiral data set.

Currently all but two bottom of the range scanners on the market in the UK perform spiral scanning as standard. Of the 360 scanners installed and operational in the UK in 1996 about one third were spiral scanners (figure 1). This fraction is gradually increasing since only spiral scanners are now purchased.

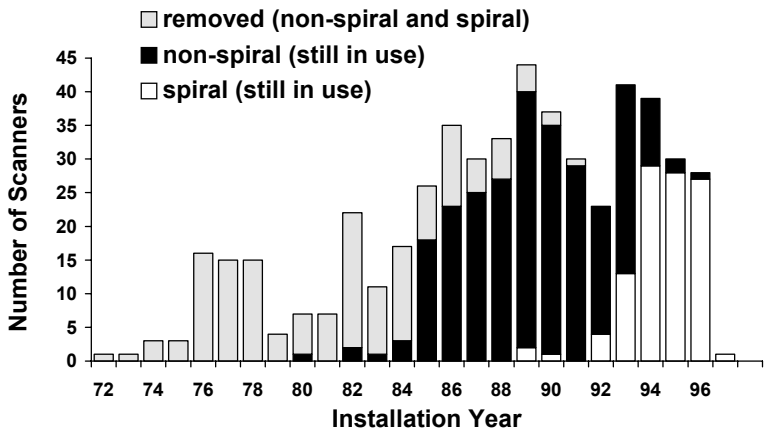


Figure 1. Status of scanners in the UK according to installation date.

Overview

■ Clinical Applications

In order to standardise an evaluation procedure, as well as for the resultant data to be clinically relevant, the technical evaluation of the scanners is based around certain clinical applications. These are termed in this report as follows (but with a more specific clinical application given in brackets):

'Standard Head' (Posterior Fossa)
'Low Noise Head' (Standard Brain)
'High Resolution Head' (Inner Ear)
'Standard Body' (Abdomen)
'Spine (low contrast)'
'Spine (bony)'

■ Measurements

Using ImPACT phantoms, the following imaging and dosimetry performance parameters are measured:

image noise
high contrast spatial resolution
imaged slice thickness (z-sensitivity)
irradiated slice thickness (dose profile)
multi-slice surface dose
$CTDI_{air}$
$CTDI_{CTPX}$

An overall dosimetry efficiency performance figure, Q, is calculated from some of these parameters.

■ Conditions of Measurement- Scan Parameters

The performance parameters are measured using scanning parameters recommended by the manufacturer for the clinical scans described above. Some of the scan parameters that can be encountered are listed below, however not all scan parameters are applicable to every scanner. They are often presented in different ways to the operator, and some are automatically changed by the scanner software due to their dependence on another scanning parameter.

The dependency of the performance parameters on the more fundamental scan parameters, such as kV, mA, and reconstruction algorithm, is addressed in a separate section. The less common scan parameters and their influence on performance parameters are discussed in the individual scanner reports.

■ **General scan parameters for axial scanning**

tube voltage (kVp)
tube current (mA)
slice thickness
field of view
convolution kernel or filter
image filters
matrix size
beam hardening algorithm
motion correction algorithm

- shaped filter
- additional flat filter
- focal spot
- number of samples
- quarter detector shift implemented
- moving focal spot

■ **Scan parameters specific for conventional axial scanning**

scan time (single slice)

- angle of tube rotation

couch increment
tube isocentre distance (two older scanners only)

■ **Scan parameters specific for spiral scanning**

pitch (or pitch factor)
table speed
total acquisition time
interpolation algorithm
image reconstruction increment

■ **Scan parameters for scan projection radiographs**

- couch speed
- slice thickness

tube current (mA)
tube voltage (kVp)
distance scanned
couch height
projection (AP/PA/lateral)

- These parameters are rarely set by the operator, and often tend to be implemented by their dependency on another parameter.

■ Definition of Pitch

For any discussion of spiral scanning 'pitch' needs to be defined. In CT it is referred to as the ratio of the distance (d) moved by the couch in one 360^0 tube rotation, divided by the nominal slice width (sw). Strictly speaking, pitch refers to a distance between two corresponding points on a helix, therefore the definition above is of a different term, and perhaps should be described (as one CT manufacturer does) as the pitch factor. However, generally in CT, both the pitch and pitch factor are used with the same definition, as given here.

$$Pitch = \frac{d}{sw}$$

Equation 1. Definition of Pitch (or Pitch Factor).

Where: d = distance moved by couch in one tube rotation
sw = nominal slice width

The equivalent ratio of couch increment (ci) to slice width (sw) is found in the conventional slice scanning mode, and this is the inverse of what is sometimes called the packing factor.

$$\frac{1}{Packing\_Factor} = \frac{ci}{sw}$$

Equation 2. Definition of a Packing Factor

Where: ci = distance moved by couch between each slice
sw = nominal slice width

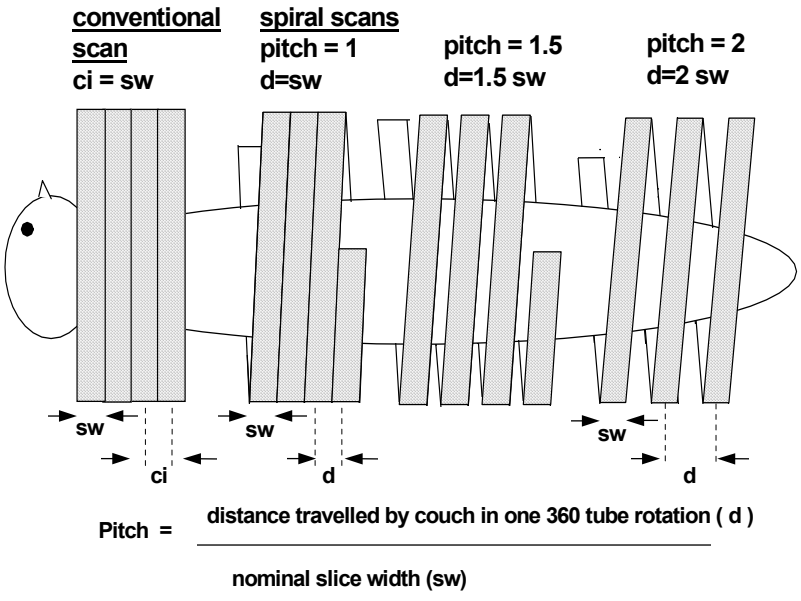


Figure 2. Definition of pitch in spiral scanning.

Methodology of Scanner Assessment

■ Rationale for Selection of Scan Parameters

■ General

To intercompare scanner models, ideally one would select identical scanning parameters for each scanner evaluation. However, due to many differences between scanners such as tube focus-to-isocentre, tube focus-to-detector distances, amount of filtration, kVp, convolution algorithms, and matrix size, it is not possible to standardise on all scan parameters.

Even where standardisation of the value of a parameter can occur, scanners are used with different set-ups in clinical practice and so standardisation does not give a realistic assessment of how a scanner would perform in clinical use. Therefore, in order to be able to compare the performance of different scanners, some baseline has to be established. The approach we have taken is to use standard clinical studies as a framework and, wherever possible, to use similar scan parameters to aide inter-comparison.

Discussions with each manufacturer are arranged prior to the assessment to determine their recommended scanning parameters for a range of typical clinical studies. For the head, these studies include a standard head scan (perhaps more representative of a posterior fossa slice), a low contrast soft tissue study (perhaps more representative of standard brain) and a high resolution study (for example typical of inner ear examinations). For the body, there is a standard abdominal scan and two small field of view scans (typical of spinal studies investigating high contrast features in one case and soft tissue features in the other). High resolution lung and paediatric scanning parameters are to be added in future evaluations. Where spiral scanning is used as standard these are looked at in the relevant application.

These clinical scans are outlined in this section, and for each type of scan a particular performance parameter, which is of most importance and relevance, is indicated.

In Comparison Reports, for example Issue 6[1], tables are presented which compare the performance of each scanner for the six typical clinical studies. These 6 clinical studies form the basis of the summary of performance for every individual scanner assessment.

It is important to understand that the clinical scan data does not necessarily indicate the limits of best performance of the scanner, or even the best use of the scanner. Data is primarily presented according to the manufacturer's recommendation of scanning parameters for the particular type of clinical scan. More data may be available as an appendix to a report. The more complete data sets may be useful for users when considering other scan protocols.

■ Spiral Scanning

Spiral scanning has now become a popular mode of scanning for a wide range of studies.

The original advantages of spiral data acquisition were seen in abdomen studies where the speed of acquiring an axial length of scan data in one breath hold greatly reduced movement artefacts and misregistration. The additional advantage of spiral CT, over conventional CT, is that after scanning the reconstruction position and spacing of reconstructed images can be chosen retrospectively.

Some centres keep head studies as non-spiral examinations since some of the benefits of spiral scanning, such as single breath hold examination, do not apply to the head region, and disadvantages can then become of higher priority. Examples of these are the slight degradation (depending on the pitch and interpolation algorithm) of the slice width, and the appearance of artefacts where high contrast objects, such as the bony regions in the skull, are changing sharply in the z-axis direction. (The z-axis conventionally describes the axis of rotation of the scanner, perpendicular to the slice plane).

It can be advantageous to acquire some high resolution studies in non-spiral mode because a well defined narrow slice is needed to maintain the resolution along the z-axis. Also, where only 3 or 4 slices are required with large intervals between them such as for high resolution lung studies it is unnecessary to perform a spiral irradiation of the whole volume.

For those scanners that have spiral scanning capability, the spiral scanning as well as the standard scanning mode is assessed for the more routine clinical applications.

■ Standard Head (Posterior Fossa)

The main requirement for the ‘Standard Head’ examination is to obtain an optimal compromise between high contrast spatial resolution and low contrast differentiation, within a reasonable scan time. The dose efficiency value, Q, is therefore used as a measure of success for this study.

Clinically, a standard head examination tends to utilise a 'standard' convolution filter for the particular scanner, and a wide or medium slice thickness. The wide slice thickness is used for the major part of the study, and referred to as a ‘standard brain scan’, but when traversing the posterior fossa the medium slice is often used clinically to reduce artefacts from that region. For this latter region it is often referred to as the ‘posterior fossa’ scan. This differentiation for standard scanning of the two regions of the head is given in the ‘Standard Head’ and the ‘Low Noise’ head as used in the ImPACT evaluations.

For the ImPACT Standard Head scan the medium slice thickness is used. The scanner’s standard head field of view is used, and this is usually 250 mm. Data is also collected for a spiral Standard Head scan using the same irradiation parameters where applicable.

■ Low Noise Head, Soft Tissue Examinations (Standard Brain)

The ImPACT Low Noise Head scan relates best to the clinical 'standard brain' scan. In the ‘standard brain’ examination it is the soft tissue differentiation of extended masses that is important, and low noise is required in the image to maximise the low contrast resolution.

The measured statistical noise can be used as a key comparative parameter, with the radiation dose per slice also being considered. Therefore, in addition to the noise value, the dose efficiency figure, Q, is a good figure of comparison for this type of scan.

For a standard brain examination, in addition to using a wider slice than the posterior fossa scans, a different convolution filter giving rise to a smoother image is often utilised to achieve lower noise, but this is at the expense of lower resolution. Lower noise is also achieved without reducing the spatial resolution, by using a higher mAs and hence a higher radiation dose to the patient. Often a combination of the two approaches, a smoother reconstruction algorithm and higher mAs, is used.

Clinically, and in the ImPACT evaluation for the Low Noise Head scan, the widest available slice is usually used, allowing for maximum noise reduction in the image.

■ High Resolution Head (Inner Ear)

In clinical practice, fine detail is often sought in high contrast situations such as bone-tissue and tissue-air interfaces. Higher noise is clinically acceptable in these types of images, compared to standard scanning, since it is the high contrast interfaces that are of interest and these can be used with a wide window setting. In clinical use it is important to match the increase in spatial frequency in the scan plane with a corresponding reduction in slice width to avoid loss of contrast for small objects or loss of resolution for edges oblique to the scan plane.

Two values of resolution performance are compared, the 50% response of the MTF, and the 10% response. For these high resolution studies it is often the 50% value which is the most reliable indicator of performance since statistical uncertainties are less at the 50% than the 10% levels on the MTF curve. However, the 10% figure has more value in relating the numerical performance to visual assessment of resolution[3].

When an image is created using a high resolution, or ‘sharp’, convolution filter, the high spatial frequencies obtained lead to increased image noise. The narrow slice width also contributes to increased image noise as the X-ray flux to the detectors is reduced. The noise can be reduced if necessary by increasing the X-ray exposure, either by increasing the mA or the scan time.

For the ImPACT evaluation the narrowest slice thickness is usually used for the High Resolution Head scan. Where the narrowest slice is 1 mm the next narrowest slice (e.g. 1.5 mm, 2 mm) is also assessed since the dose increase for a 1 mm slice thickness in a multi-slice irradiation can be significant due the difficulty in collimating to 1 mm. A reconstructed field of view of 120mm is used.

Manufacturers have differing opinions as to an acceptable high resolution algorithm for clinical use, in that the acceptability of the associated noise, and artefacts from aliasing, is viewed differently by different manufacturers.

We have attempted to use a fair approach, and where a higher resolution algorithm is available, but perhaps not usually clinically recommended, this is referred to in the report, and/or inserted into the tables together with the recommended algorithm to show the scanner's limiting resolution capability.

■ Standard Body (Abdomen)

Movement artefacts are a problem for body studies, so short scan times are desirable. In this study, as in the Standard Head scans, a compromise between noise and spatial resolution is sought. Therefore, the Q value has been used as a measure of performance.

Clinically, and in the ImPACT study, the widest slice thickness is usually employed along with a standard convolution filter. The field of view used by ImPACT is between 350 mm and 400 mm, the specific dimension depends on which is available on the scanner.

A spiral standard body scan is also part of this protocol since spiral acquisition is now routine for standard body scanning. The same scan parameters are used as for the standard axial scan where applicable.

■ Small Field Of View Body Scan (Spine)

Spinal studies are normally carried out with a narrow slice thickness and, due to the curvature of the spine, often with gantry angulation. A small field of view is normally used. There are two types of spinal study examined in these type tests. One is used to investigate bony structures where the high contrast resolution is of interest. The other is where soft tissue differentiation is required.

■ High Resolution Spine

The high resolution spine scans are compared according to the 50% MTF value. Although two values of resolution performance are given, the 50% response of the MTF, and the 10% response, for these high resolution studies it is often the 50% value which is the most reliable indicator of performance since statistical uncertainties are less at the 50% than the 10% levels on the MTF curve. However, the 10% figure has more value in relating the numerical performance to visual assessment of resolution[3].

Clinically, and in the ImPACT tests, a narrow, though not the narrowest, slice width is usually used with a sharp reconstruction algorithm. Gantry angulation is not used for the ImPACT study. A reconstructed field of view with a diameter of 120 mm is normally used.

■ Low Noise Spine

The soft tissue spine studies are compared according to noise values as well as the dose efficiency value Q.

Clinically, and in the ImPACT tests, a medium slice width is usually used for this study with a smooth reconstruction algorithm. Gantry angulation is not used for the ImPACT study. A reconstructed field of view with a diameter of 120 mm is normally used.

■ Scan Projection Radiograph (SPR)

The scan projection radiograph (manufacturers' terminology includes; 'Scoutview', 'Scanogram', 'Topogram', 'Surview' etc.) is not only used to allow the radiographer to prescribe the position of the slices in examination, but also used in its own right for pelvimetry distance measurements.

The scan parameters used in the assessment are those commonly used for both routine purposes and for pelvimetry measurements, if different. Distance measurement accuracy and dose are the two performance parameters of interest.

■ Summary of Performance Parameters for Axial Scans

The following table indicates a level of significance for the performance parameters, with relation to the particular clinical applications discussed above. Of course dose values are always important, and the dose efficiency value, Q, is dependent on a combination of noise, dose and resolution. However the purpose of this table is to give at least some indication of importance when considering image quality issues in the context of clinical application.

Table 1. Summary of imaging performance parameters used in ImPACT 'clinical scans' results.

ImPACT Scan Type	Noise	Spatial Resolution	Dose	Q
Standard Head	1	1	1	1
Low Noise Head	1	2	1	1
High Resolution Head	3	1	2	-
Standard Body	1	1	1	1
High Resolution Spine	3	1	2	-
Low Noise Spine	1	2	1	1

Key:

Primary Performance Parameter	1
Secondary Performance Parameter	2
Tertiary Performance Parameter	3

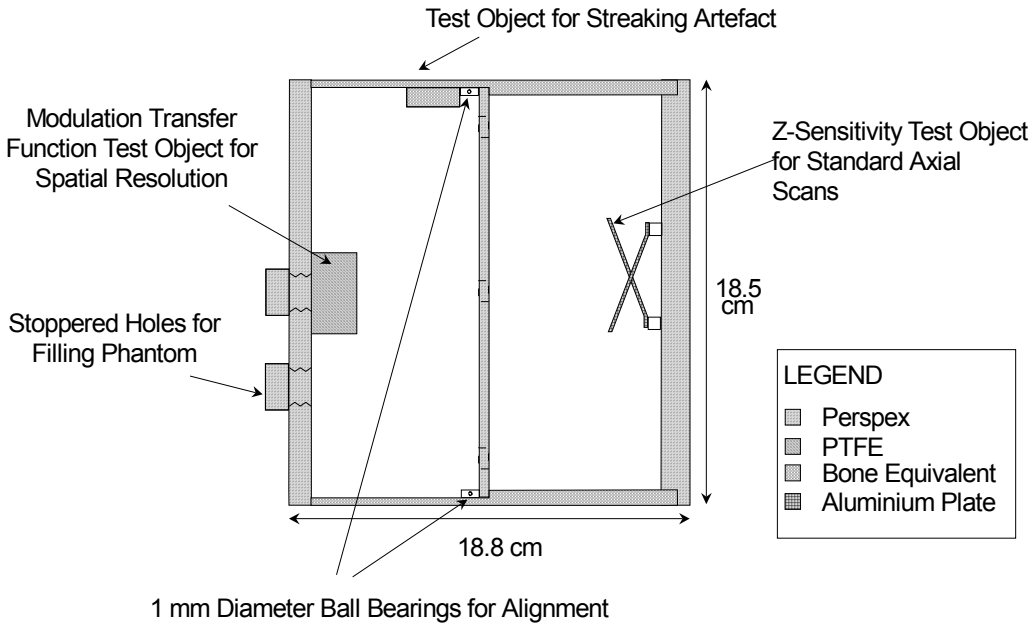


Figure 3. ImPACT head phantom.

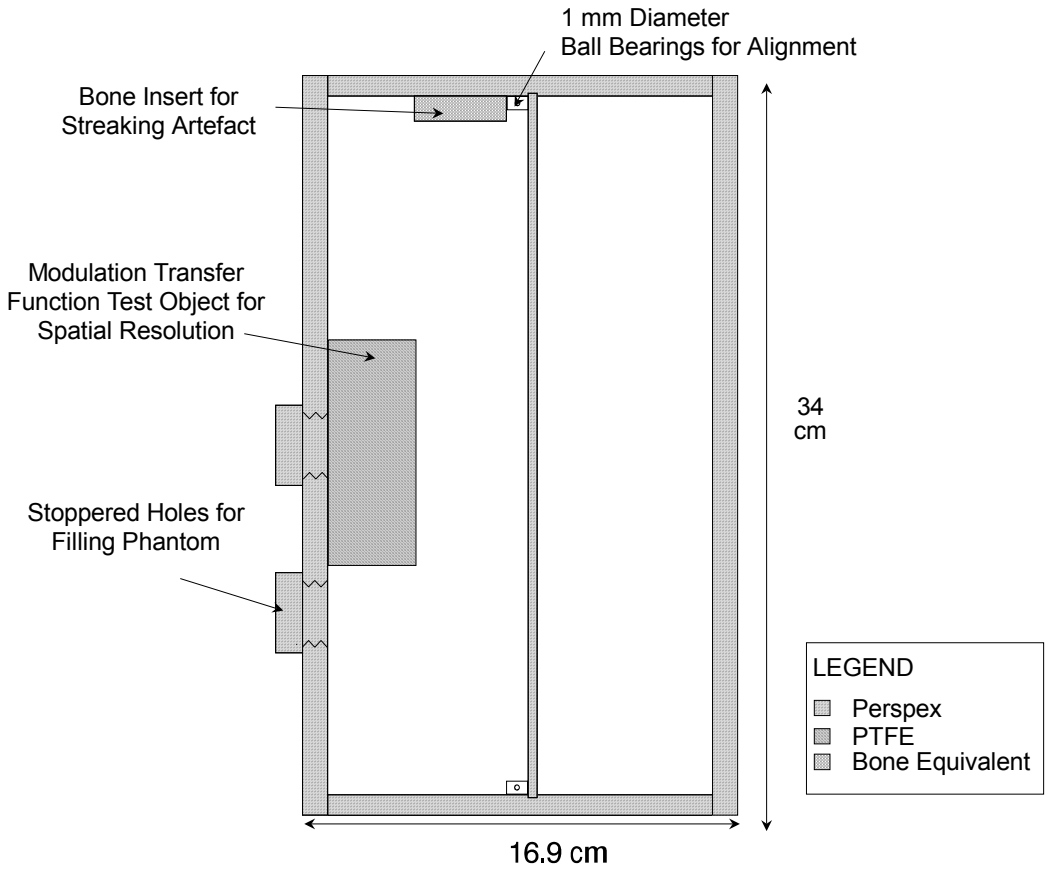


Figure 4. ImPACT body phantom.

Two main sets of phantoms are used, a water filled head and body set for the main image quality performance parameters and a further perspex set for the Computed Tomography Dose Index (CTDI) measurements. In addition, separate phantoms for spiral z-sensitivity and for determination of the scan projection radiograph accuracy are used.

■ Image Quality Phantoms

The ImPACT head phantom is water filled with a 185 mm outside diameter. It has a bone equivalent shell of 3 mm thickness for one half of its length and 8 mm thickness for the other half (figure 3). The ImPACT body phantom has a 340 mm outside diameter, is also water filled, but with a perspex shell (figure 4).

The phantoms contain a range of test objects to measure the range of performance parameters investigated in the scanner evaluations, these are shown in figure 5, and are discussed in the relevant chapters of this report.

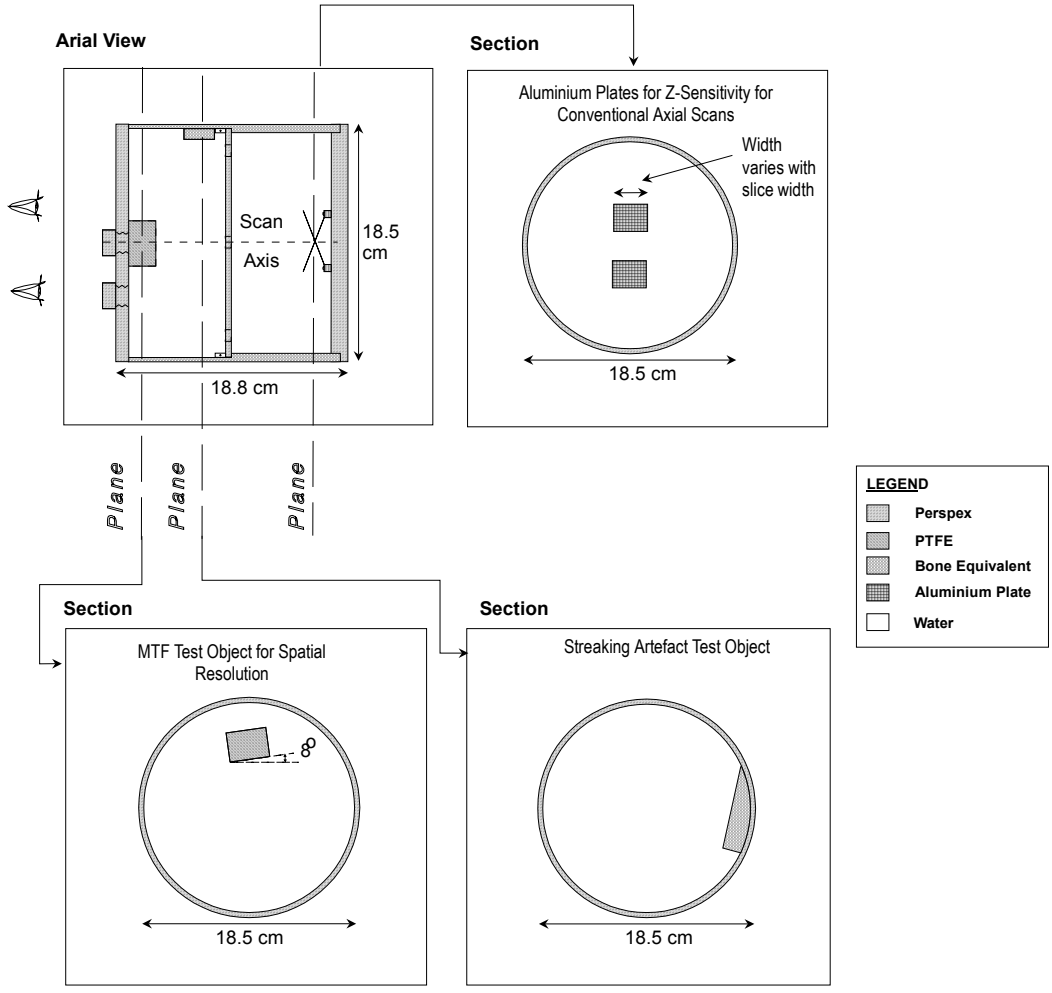
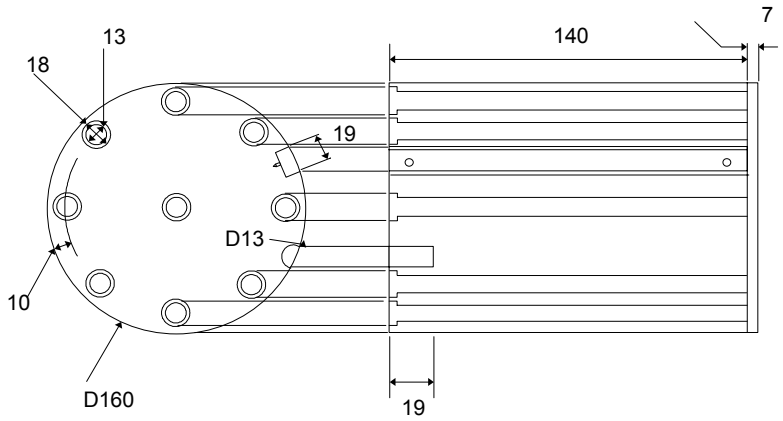


Figure 5. ImPACT head phantom: arial and cross sectional views.

■ CTDI Measurement Phantoms

The phantoms used to measure the Computed Tomography Dose Index are made to the dimensions and composition specified by the Food and Drugs Administration (FDA)[4] in the USA. They have become an internationally accepted design of phantom and the same dimensions and construction are recommended within the IEC standard for constancy testing for Computed Tomography[5].

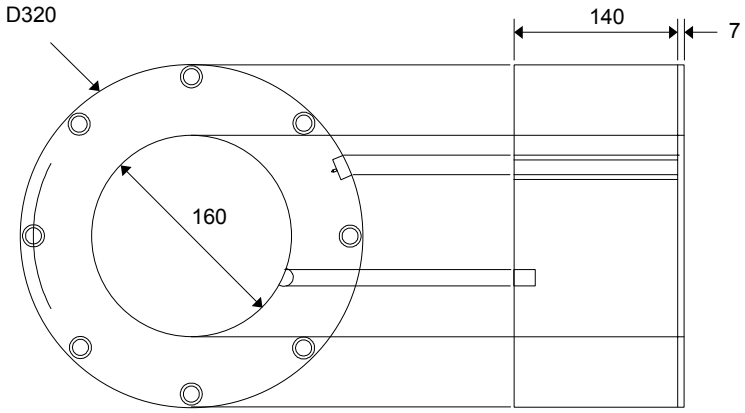
These phantoms are constructed using polymethylmethacrylate (perspex) and contain a number of removable plugs. On removal of one of these plugs a suitable dosemeter, such as an ion chamber or an insert containing thermoluminescent dosimeters, can be inserted into the resultant hole. The head phantom is 160 mm in diameter, and the body phantom, 320 mm (figures 6 and 7).



All measurements in mm

Key: D - Diameter

Figure 6. Perspex head phantom for CTDI measurements.



All measurements in mm

Key: D = diameter

Figure 7. Perspex body phantom for CTDI measurements (annulus that fits over head phantom.)

Special alignment inserts[6] and the addition of a surface dose position are included on those phantoms used by ImPACT, and shown in figures 8 and 9.

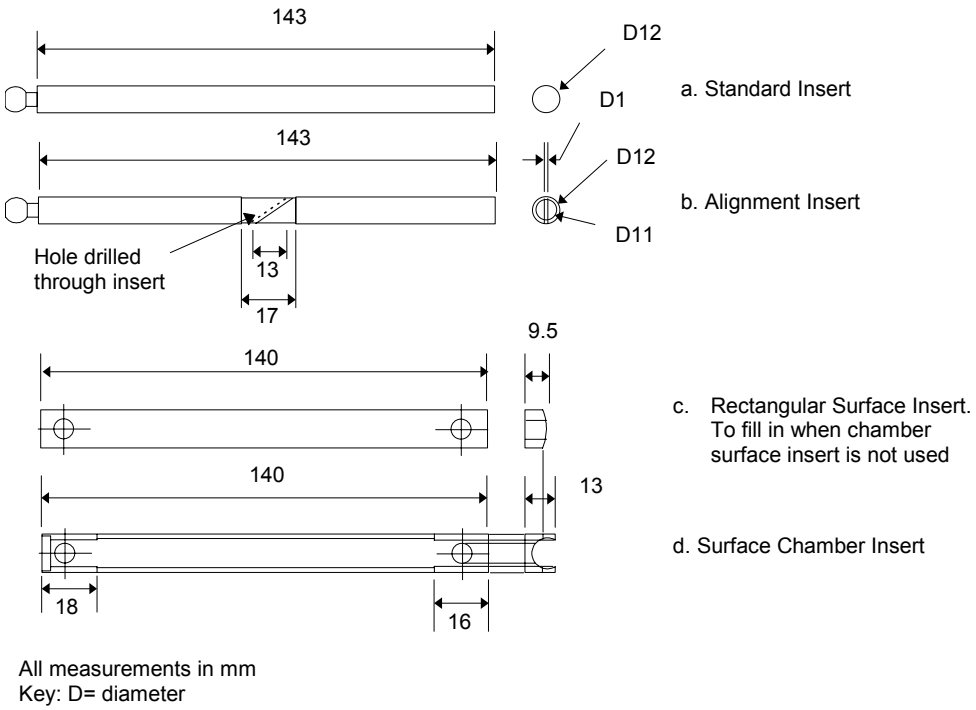


Figure 8. Inserts for perspex phantoms for CTDI measurements.

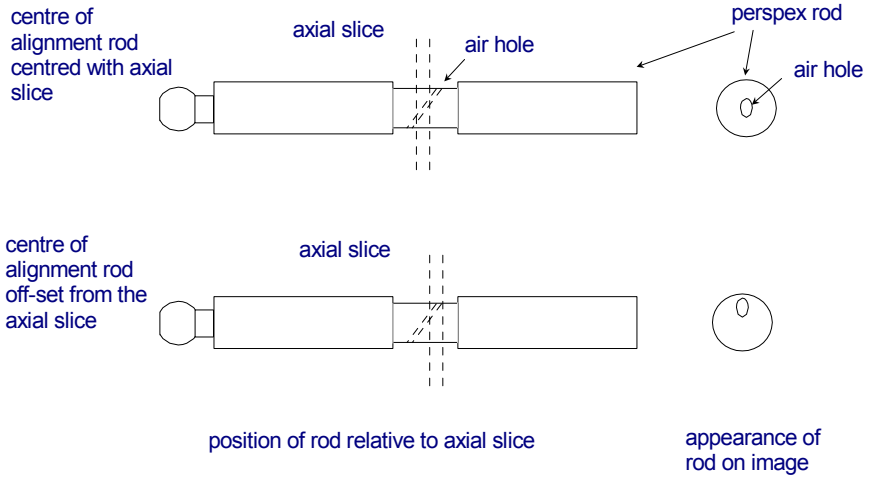
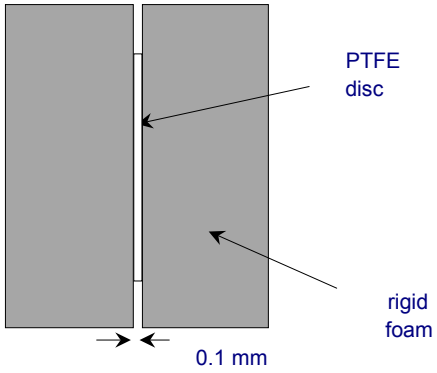


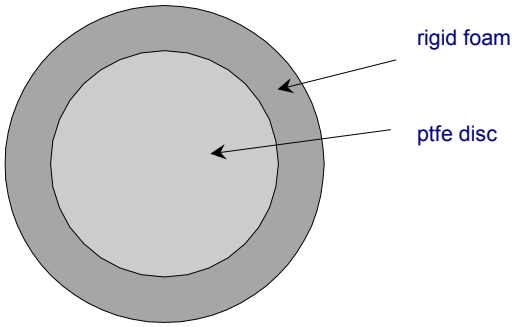
Figure 9. Detail of alignment inserts for CTDI phantom.

■ Z-sensitivity phantom

A separate z-sensitivity phantom is used for the evaluation of the spiral imaged slice thicknesses. This consists of a thin disc of PTFE (polytetrafluoroethylene) sandwiched between pieces of rigid foam[7] (figure 10). The partial volume effects of the PTFE disc and the relative sensitivity of sequential imaged slices to this thin piece of PTFE are used to identify the z-sensitivity. More details are given in the chapter on performance parameters.



a. Plan View



b. Cross-sectional View

Figure 10. Phantom for z-axis resolution.

■ SPR Distance Phantom

The accuracy of the distance measurement is checked on the scan projection radiograph (SPR) scans using a test object usually used for testing image intensifiers[8]. A test object produced by FAXiL is used (FAXiL is the Facility for the Assessment of X-ray Imaging in Leeds, another Medical Devices Agency Evaluation Group). It is part of a set of circular test objects of different diameters usually used for testing image intensifiers. The discs are made of perspex and contain a grid of copper wires (figure 11). The particular one that ImPACT uses has a diameter of 380 mm.

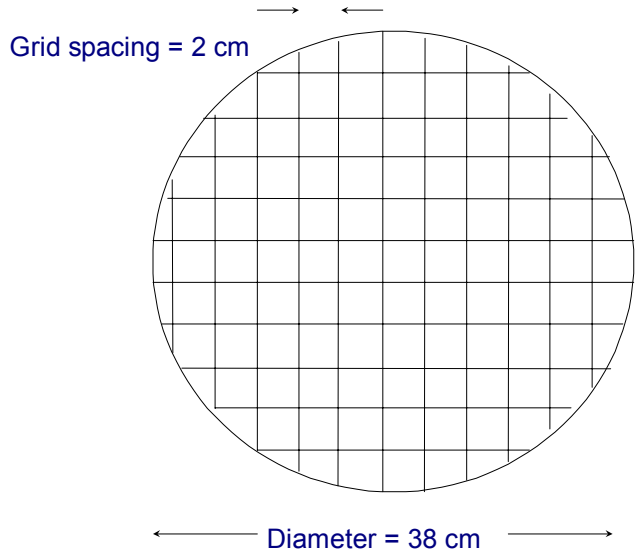


Figure 11. Leeds test object for SPR measurements,

■ Dosemeters

For phantom surface dosimetry, lithium fluoride thermoluminescent dosemeters (TLD's), 3.2 mm x 3.2 mm x 0.9 mm, are used.

A 10 cm long pencil ion chamber is used for CTDI measurements which are carried out both 'free in-air' and in (and on) the perspex phantoms.

Two types of envelope wrapped film are used. For dose profile measurements, which are made in air at the iso-centre, Kodak X-OMAT V is used. This is a rectangular film (25 cm x 31 cm) which is often used for radiotherapy field verification.

To establish the variation of dose around the circumference of a phantom Agfa Structurix DP2 is used[41]. This film, primarily designed for industrial purposes, consists of a long roll, about 5 cm wide, which can be cut to required lengths. The rectangular X-OMAT V can also be used, but two or three films have to be taped together. Special processing arrangements are made for the Agfa Structurix film since the processing times are different and from those used for common diagnostic films and, therefore, the automatic processor cannot be used.

Performance Parameters used in ImPACT 'Clinical Scan' Results

■ Introduction

This section considers the parameters which are used to describe the image quality for the clinical scans following the rationale given in the Methodology of Scanner Assessment section. These form the basis of the summary imaging performance tables given in individual scanner reports.

Except for the spiral z-sensitivity measurements, the main phantoms used to obtain the parameters in this section are the ImPACT head and body phantoms. Image data is collected from the scanner and, for independent evaluation, is analysed on a Sun workstation using an in-house programme called ImPASS.

A brief description of each performance parameter and the methods of measurement is given. Further information is found in the chapter entitled 'Effect of Scan Parameters on Performance Parameters'. Estimates are given on the precision of the particular measurements made. Currently, for most parameters, no estimate is made of the variation between individual scanners of a particular model. Some estimate of dose variation is given.

■ Spiral Scanning

The performance parameters are initially described for the evaluation of non-spiral scans, however where techniques have had to be adapted specifically for the spiral scanning, this is indicated. Usually where the test object is uniform along the z-axis the technique can be used for the images produced from a spiral dataset.

When using an appropriate region in an existing phantom care must be taken to ensure that the image is not reconstructed from projection data taken from outside that region. For example, in figure 12, an image of the uniform water region must not be reconstructed from attenuation data taken from the non-uniform areas of the phantom which occur either side of the uniform water region.

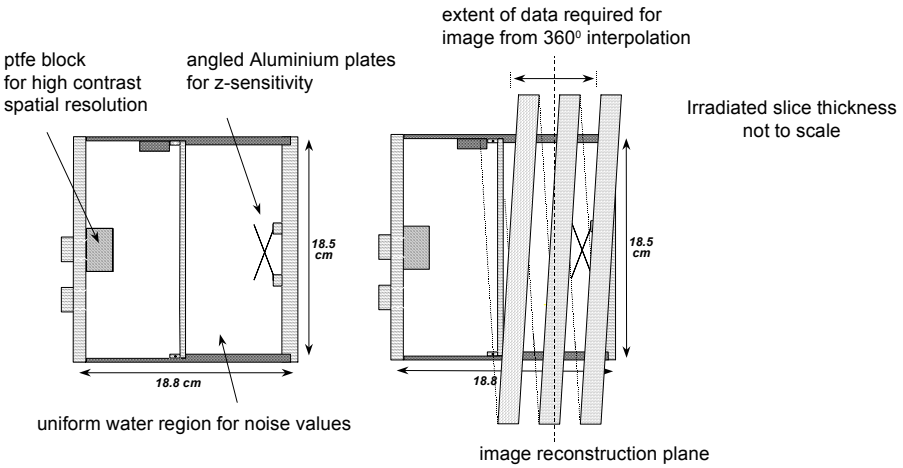


Figure 12. Restrictions on use of phantom for spiral data acquisition.

■ Noise (σ%)

The noise parameter describes the amount of statistical variability of the CT number in an image. It does not differentiate between random or structured noise. Noise is quoted as the percentage standard deviation of the CT numbers of a group of pixels with respect to the difference in CT number between water and air[9]. The water region of the phantom is scanned and a region of interest (ROI) placed at the centre of the image (figure 13).

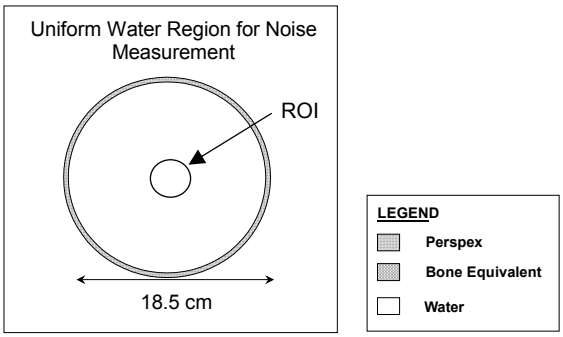


Figure 13. Cross-sectional view of uniform water region.

The (ROI) used has an area of approximately 480 mm^2, which is kept constant for all fields of view (FOV). This is expressed as an area, as opposed to a radius, since on the scanner console, area information, but not the radius, is usually given when forming ROIs. The area chosen is reasonably arbitrary, but based on a region large enough to obtain reasonable statistics whilst small enough to avoid non-uniformity effects in the image.

The percentage noise is given as:

$$\sigma(\%) = \frac{\sigma_{roi} \times 100}{CT_{H_2O} - CT_{air}} \qquad \textbf{Equation 3}$$

where: σ_{roi} = standard deviation of region of interest placed in centre of image
CT_{H_2O} = CT number of water
CT_{air} = CT number of air

Normally, the difference in CT number between water and air is 1000, however in certain instances, particularly with the use of high spatial resolution algorithms, this range may be compressed to 500 or 250.

The repeatability of the noise from successive scans is only expected to be within 3% due to the variation in output, however on some scanners a range of 10-14% is found.

In scanners without slip rings where the tube, and detectors on third generation scanners, are constrained by the length of the cables connected to them, the tube (and detectors) need to rotate first in one direction, and then back in the other direction for the next scan. Tube rotations in alternate directions sometimes give rise to slight variations in noise values (and CT numbers).

■ Spiral scanning

The above method of measurement of noise can be used for spiral scanning provided the region over which data is collected and interpolated to create the image is uniform along the z-axis. Differences between noise values from standard axial images and those from spiral acquisitions are discussed in the 'Effect of Scan Parameters on Performance Parameters' chapter.

■ High Contrast Spatial Resolution ($f_{50\%}$), ($f_{10\%}$)

High contrast spatial resolution in the scan plane can be evaluated in a number of ways and an objective method is to measure the modulation transfer function (MTF). This describes how well object spatial frequencies in the scan plane are transferred into the image, with higher frequencies relating to higher spatial resolution.

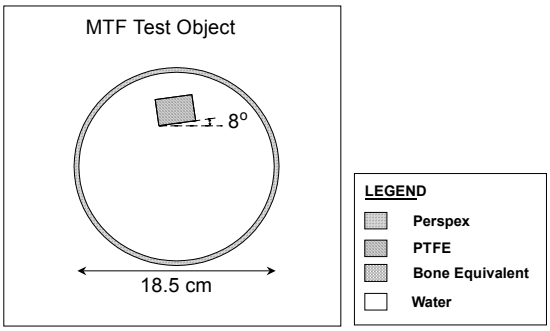


Figure 14. Cross-sectional view of resolution test object.

The method used by ImPACT utilises a profile of CT numbers taken across a high contrast sharp edge, from water to PTFE, figure 14. This is the edge spread function. By differentiation the line spread function (LSF) is achieved and the modulus of the fourier transform gives the MTF[9,10].

The PTFE block is inclined at a small angle to the pixel array in order to obtain data points at intervals less than the pixel width for the edge spread function. The technique employed enables frequencies beyond the Nyquist limit, imposed by the matrix size and field of view, to be investigated[9,10]. The Nyquist frequency equals the reciprocal of twice the pixel size. Since this technique essentially overcomes the limit imposed by sampling at the pixel size, it results in the 'pre-sampling' MTF.

The shape of this curve varies according to the convolution filter used (figure 15). In the scanner evaluations, the spatial frequencies at which the modulation transfer function (MTF) falls to 50%, and to 10%, of its initial value are quoted. The two values allow a more complete description of the modulation transfer curve because of the differing shapes. The 50% and 10% MTF levels are indicated on figure 15, with the frequency at the 50% MTF level shown for the smooth filter (a) and the high resolution filter (b).

The value at the 10% level of the MTF is more closely related to the visual assessment of the resolution, which is at about the 5% level of the MTF[3], and therefore has scope for easier interpretation by converting the spatial frequency to an object size and associated spacing. However the value at the 10% level is determined with less certainty, particularly in the evaluation of high resolution scans, due to statistical variations.

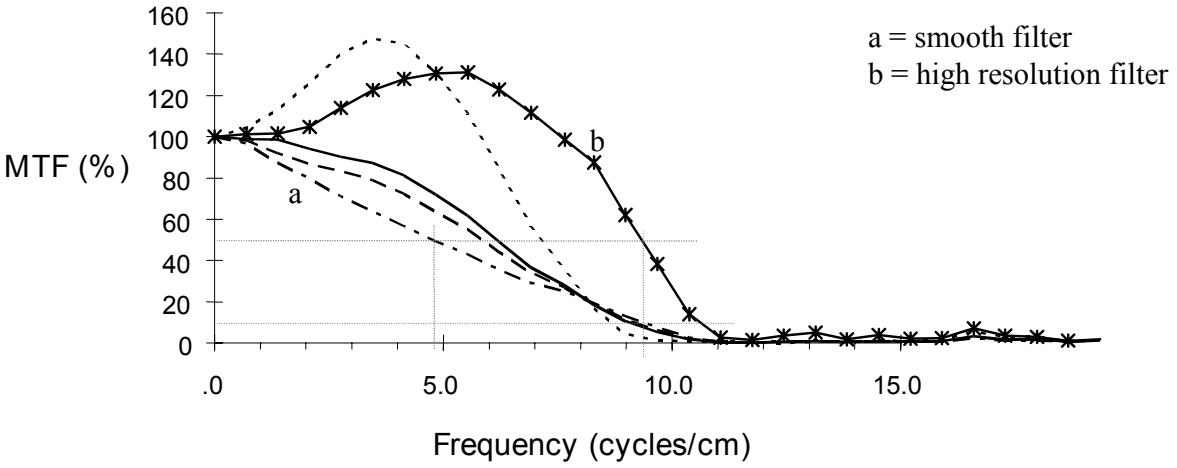


Figure 15. Variation of MTF shapes with convolution filter.

High spatial resolution images generally tend to be difficult to analyse because of the noise associated with the high spatial frequency content, as demonstrated in the example in figure 16 by the use of low mAs. The resulting noise on the MTF curve produces a greater uncertainty in both the 50% and the 10% values compared to standard resolution images. A standard resolution image, using the same levels of mAs, is given in figure 17, showing no degradation in the curves. Where possible a high mAs is used for high resolution scans to reduce the error in the measurement.

The accuracy of MTF values quoted is about +/-0.1 cycles/cm for images reconstructed using smooth or standard algorithms to between +/-0.5 cycles/cm and +/-1 cycles/cm for the high resolution filters.

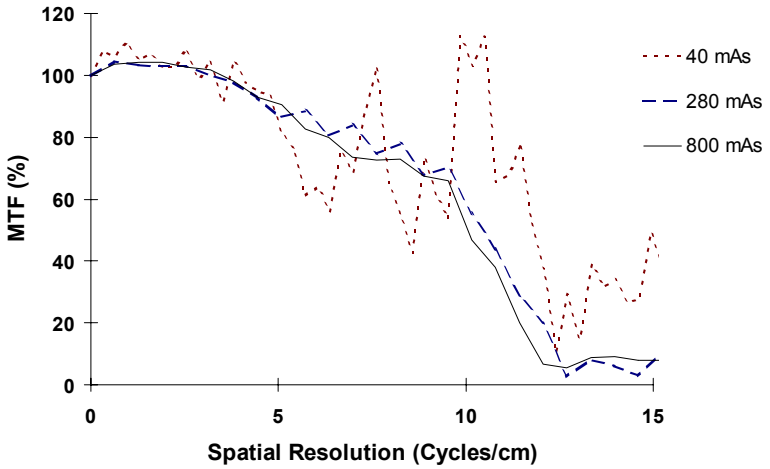


Figure 16. Degradation of measurement of the MTF from high resolution images due to low mAs (head phantom used).

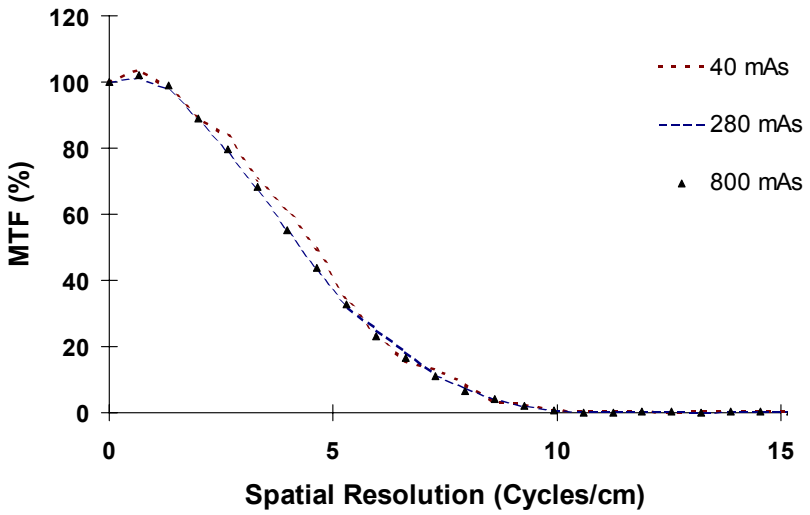


Figure 17. MTFs from images obtained with a standard resolution filter (head scan), showing no ill-effect from low mAs.

■ Spiral scanning

This method of measurement of resolution will not be affected by spiral scanning since the test object is uniform along the z-axis, as long as the interpolation distance is less than the extent of the test object along the z-axis.

Differences in spatial resolution between spiral and standard axial scans may however occur where the reconstruction algorithms applied for spiral imaging are different than those for the standard imaging.

■ Z - Sensitivity (z) - Imaged Slice Thickness

The z-sensitivity is the value defining the width of the imaged slice along the z-axis. The z-axis conventionally describes the axis of rotation of the scanner, perpendicular to the slice plane.

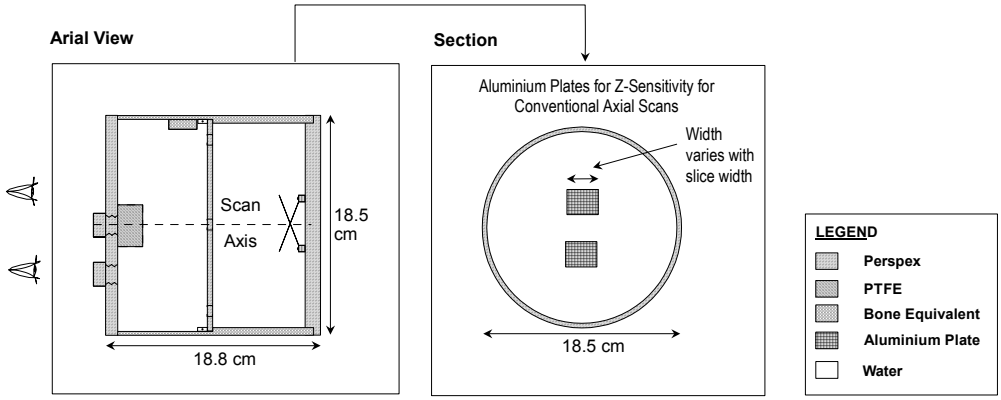


Figure 18. Arial and cross-sectional view of aluminium plates for z-axis sensitivity.

The z-sensitivity for non-spiral scans is measured using an aluminium plate inclined at an angle of 30 degrees with respect to the scan plane[9] (figure 18). It is quoted as the full width at half maximum (FWHM) of the profile of CT numbers taken across the image of this plate with an appropriate correction factor for the angle and the thickness of the aluminium plate. This is a standard type of test and, in this phantom, the measurements are taken as closely as possible to the centre of the field of view. This is because the z-sensitivity is not necessarily uniform over the whole scan field. The results from the two plates inclined in opposing directions, shown in the diagram, can be averaged to correct for any non-alignment of the phantom in the scan plane.

The uncertainty in this measurement is determined by the pixel size. Images are reconstructed with small fields of view with pixel sizes of typically about 0.2 mm, giving rise to an uncertainty of approximately 0.1 mm due to interpolation of the profile values.

■ Spiral scanning

Published studies[7], as well as our own findings on spiral scanners[11], have cast doubt on the validity of the method of inclined ramps for investigating the reconstructed slice thickness from spiral scanners. An alternative method for evaluating the z-sensitivity of spiral scans is therefore used[7].

A thin disc, about 0.1 mm, of a high CT number material, polytetrafluoroethylene (PTFE), is sandwiched between two thick discs of rigid foam. A spiral scan is performed and images are reconstructed at small intervals (between 0.1 mm and 1 mm depending on the slice thickness). A region of interest is placed over the central portion of each image, and the CT number obtained.

For images reconstructed at the z-axis position of the PTFE disc the CT number measured will be high relative to the background, representing the partial volume of the disc in the central region of the slice. For images reconstructed away from the z-axis position of the disc there will be a decreasing CT number depending on the sensitivity of the slice at the position of the disc (figure 19). The mean CT number value of the region of interest is plotted against distance to give a line spread function in the z-direction. This technique is shown in figures 20 and 21.

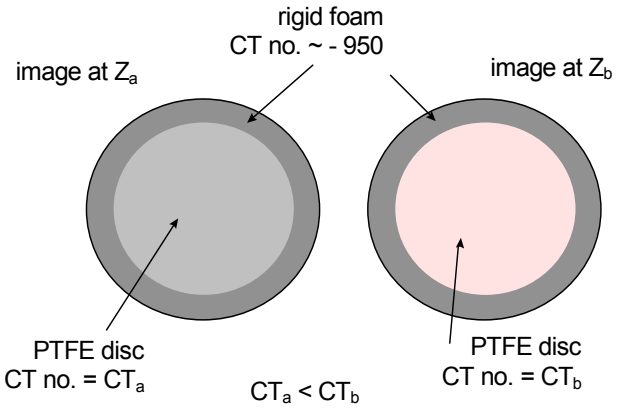


Figure 19. Appearance of images from z-sensitivity disc phantom.

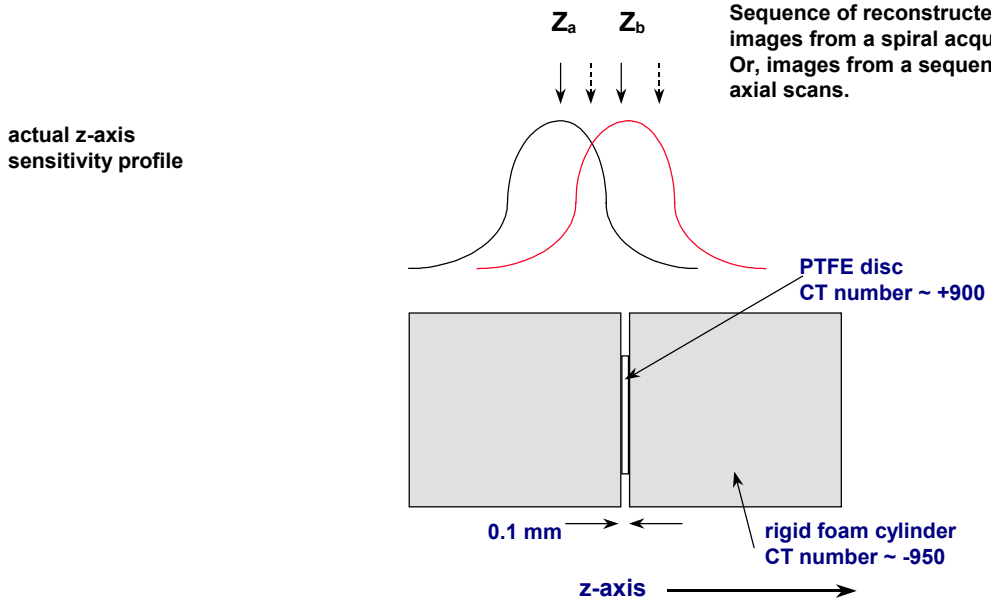


Figure 20. Imaging sequence for spiral z-axis sensitivity.

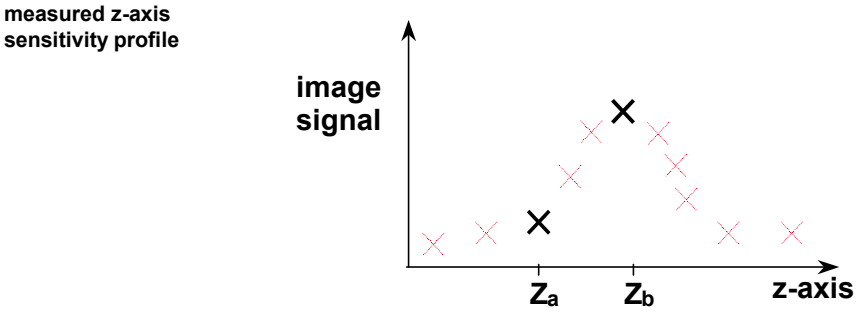


Figure 21. Z-axis sensitivity disc method showing plot of CT numbers.

The same approach can also be applied to a high contrast bead with a sub-millimetre diameter. This has some alignment advantages over the disc method[12], however ImPACT have used the disc method successfully and currently use it as the standard method.

This method can also be used for standard axial scans by incrementing the test object by small intervals along the z-axis and scanning at each position to obtain a series of images.

The conventional descriptor, the full width at half maximum (FWHM) of the z-axis sensitivity profile is a measure of the imaged slice width. However the FWHM is not a complete description of the z-sensitivity. The shape of the z-sensitivity profiles broadens and changes for images from a spiral acquisition relative to those from standard axial scans, and also depends on pitch and interpolation algorithm. Therefore a fuller description needs to be given of the z-sensitivity profile.

The full width tenth maximum (FWTM), the full width tenth area (FWTA) or the slice profile quality index (SPQI) have been suggested as additional indices[13] (figure 22) and ImPACT will be looking at incorporating these values into the scanner reports.

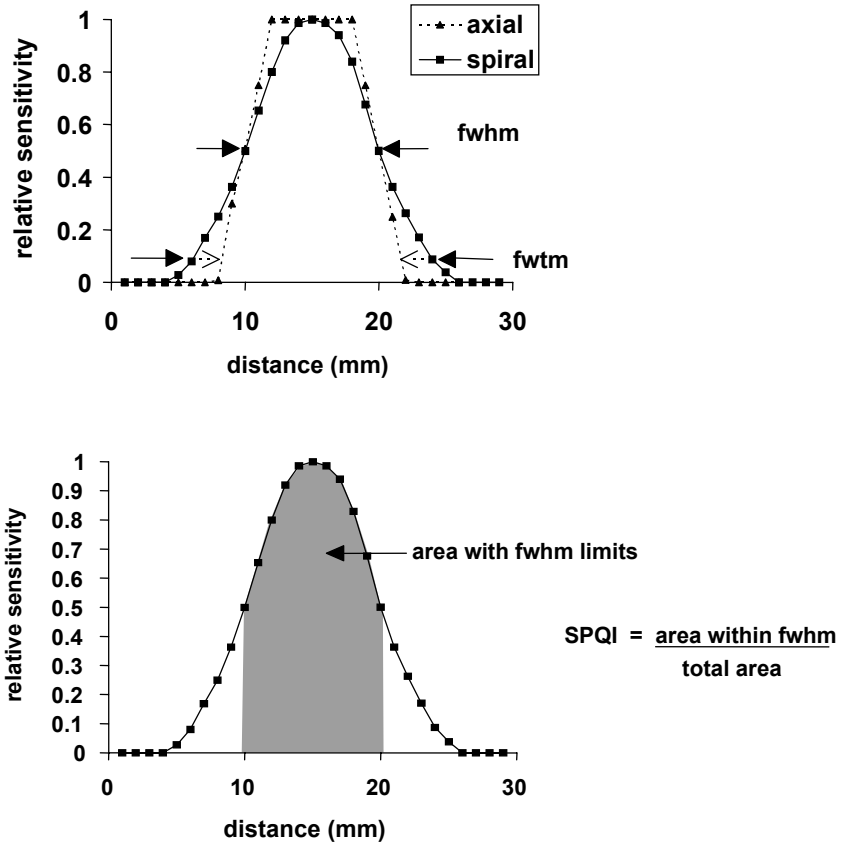


Figure 22. Descriptors for z-axis sensitivity. (Based on a figure from reference 13, with permission from AAPM and Advanced Medical Publishing[13])

■ Dose Profile Full Width Half Maximum - Irradiated Slice Thickness

This measurement describes the width of the irradiated slice.

The irradiated slice thickness is not the imaged slice thickness. The difference will depend on pre- and post-patient collimation settings. If measured at identical positions in the scan plane, the irradiated slice thickness will always be equal to or greater than the imaged slice thickness.

Specifically the irradiated slice thickness is given as the full width half maximum (FWHM) of the dose profile along the z-axis obtained from a single scan. This can be obtained using either envelope wrapped radiotherapy verification film or lithium fluoride thermoluminescent dosimeters (TLD's).

The dose profile measurements are made in air along the scanner axis of rotation at the isocentre (figure 23). This is to enable a direct comparison with the imaged slice thickness, the z-sensitivity, at the same position in the scan plane. No scattering medium is used[a]. The TLD readings or the optical density (converted to dose) of the film is plotted to show the irradiated dose profile (figure 24) and the full width at half maximum (FWHM) is quoted.

[a] In earlier scanner assessments dose profile measurements were made on the phantom surface. Comparisons could only be made with the z-sensitivity by geometrically extrapolating the surface dose profile FWHM to the axial position. This could not take into account the differing scatter contribution which led to uncertainties in the extrapolated values. Measurements on the phantom surface are no longer included in the evaluations.

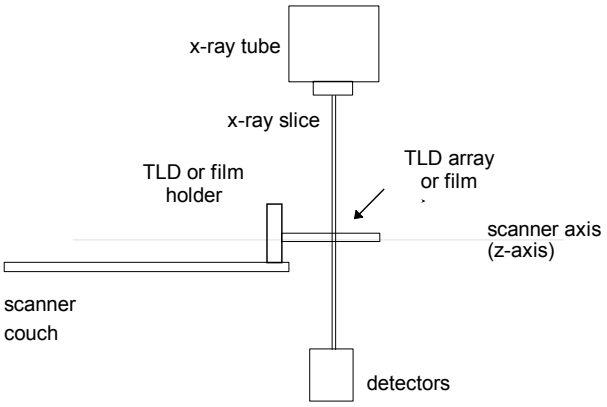


Figure 23. Lateral view of arrangement for dose profile measurements.

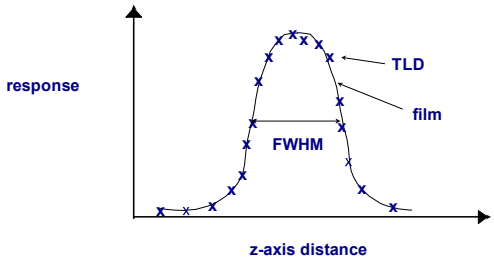


Figure 24. Dose profile FWHM measured by film and TLD.

Film microdensitometry is considered to produce more accurate FWHM measurements than the TLD profiles[14], and the thickness of the TLD (with a minimum edge thickness of 0.9mm) means that narrow slices cannot easily be evaluated. Film is now used as the method of preference. The optical densities are converted to dose using a calibration curve, although in reality, with the film and exposure conditions used by ImPACT, this presents only a small correction to the measured full width half maximum. The accuracy of this measurement is about 0.1 mm.

■ Spiral scanning

The measurement of the single slice dose profile is not applicable for spiral irradiation. The assumption is usually made that, for the same set nominal slice width, the irradiated width collimation is the same for a standard axial scan as for a spiral irradiation. However, where there is uncertainty a film can be placed as though to obtain a single slice dose profile, but this time fixed independently of the couch and its movement. A spiral irradiation (of short length and with low mA to avoid saturating the film) can then be carried out, and the developed film assessed.

A scanner may alter the collimator settings when moving from a standard axial scan to what is thought to be an equivalent spiral irradiation, particularly where the setting up procedure refers to the imaged or effective slice width. The imaged slice width is dependent not only on the irradiated width but also on the pitch and the interpolation algorithm, and in this instance the irradiated profile may need to be investigated.

■ Multi-Slice Surface Dose (D)

The multi-slice surface dose, D, is the average dose (averaged over a nominal slice width) measured on the surface of the phantom from a series of contiguous slices in conventional slice scanning. It is equivalent to the multiple scan average dose (MSAD)[15].

The measurement, D, is obtained by placing a tray of thermoluminescent dosimeters (TLD's) on the phantom surface. Each TLD measures 3.2 mm x 3.2 mm x 0.9 mm. The TLDs are placed flat on the phantom surface. A series of seven contiguous scans is performed, incrementing by the nominal slice width. The fourth scan coincides with the centre of the TLD array. In this way the net dose from a series of slices is measured (figure 25).

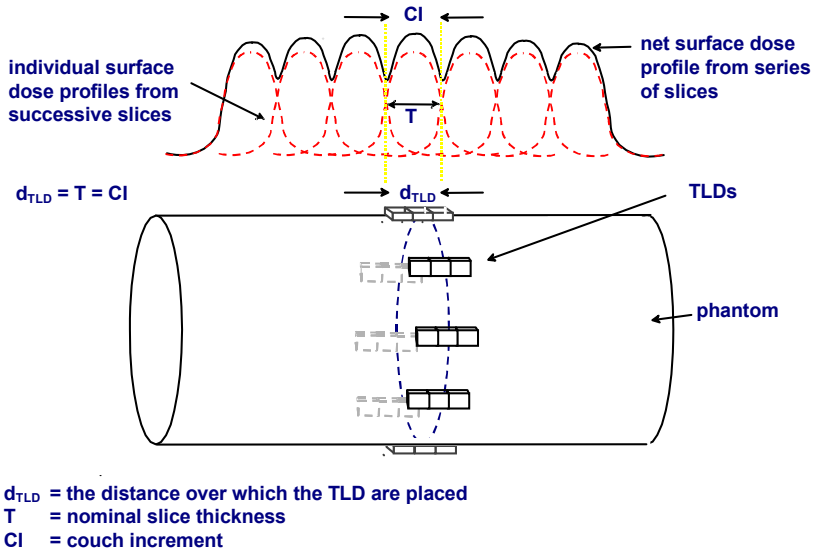


Figure 25. TLDs on Phantom for multi-slice dose measurement.

To obtain a true average dose, the number of TLD's in each tray is chosen to ensure that the length of TLD material matches as closely as possible to an integral number of nominal slices. For example for a 10 mm slice, 3 TLD chips, giving a length of 9.6 mm, are used. For dual detector irradiation, where two slices are imaged simultaneously, the TLDs cover a length equal to twice the nominal imaged slice thickness.

The average surface dose from a series of scans can be significantly over or under estimated if the TLD length (d_{TLD}) does not match the distance of an integral number of nominal slices. This is because of the peaks or troughs that can occur in the multi-slice surface dose. The nominal slice width is usually matched to the irradiated slice width at the isocentre. However, on the phantom surface, because of beam divergence the irradiated slice width will be narrower than the nominal slice width, giving rise to troughs in the multi-slice dose profile between the irradiated slices. Where the system is poorly collimated the irradiated width may be wider than the nominal slice width giving rise to peaks in the multi-slice dose profile.

The dose is measured at eight positions around the circumference of the phantom and the mean value of these eight dose readings, D, is given in the 'clinical scan' tables in each scanner report.

Detailed information on the calibration conditions of the TLD used for the multi-slice dose measurement is given in Appendix 2.

The overall uncertainty of the measurement of D is 5%, mainly due to the inherent sensitivity variations in TLDs.

Measurement of Overscan

The eight dose values are plotted to show any non-uniformity around the phantom surface. Some non-uniformity will be apparent due to the couch or the head rest but larger amounts may be due to 'overscan', where the X-ray exposure is over an angular range which is greater than 360^0. This non-uniformity is of importance where sensitive regions such as the eyes are scanned. For a multi-slice irradiation the overscan may not necessarily be at the same angular location for each slice. This is because, although sometimes the position at which the tube starts irradiating is constant, sometimes the tube starts irradiating at one of two positions, and sometimes it has a random start. The overscan therefore, as recorded by the TLDs, will be the net effect from a series of slices where the overscan, may, or may not, be at the same angular position for each irradiated slice.

The single slice overscan can be determined if necessary. Where the tube starts its irradiation in the same position around the gantry for each scan, the non-uniformity distribution can be demonstrated by CTDI measurements, described in the next section, performed at different angular positions within a phantom.

However where the angular start position of the irradiation is not known, or is variable, it is better to use TLDs, as with the multi-slice irradiation, or wrapped film. The types of film used are described in the section on 'Phantoms and Dosemeters'. One of the types of film used is primarily designed for industrial purposes, and consists of a long roll, about 5 cm wide, which can be cut to required lengths[41]. Where rectangular film is used two or three films have to be taped together.

One, or more, of the multi-slice TLD data, the single slice CTDI measurements or single slice film densitometry measurements, from around the phantom circumference, are incorporated in some reports.

■ Spiral scanning

A similar approach is undertaken for spiral scanning, except a distance is scanned which provides the same imaged length as the seven slices from the contiguous standard axial scanning. Because interpolation techniques are involved this will necessitate extra rotations of the tube at either end of the imaged length. The extent of this extra irradiation will depend on the interpolation technique. For example, where images are reconstructed using the 180^0 linear interpolation, half a tube rotation, at either end of the imaged series, is required. A full rotation at either end is needed for a 360^0 linear interpolated image, and sometimes more for the non-linear interpolations. However, this extra irradiation does not significantly affect the dose at the centre of an irradiated length when using at least the number of rotations as used in the method described above.

The spiral length irradiation, using a pitch of 1, should result in a similar dose to that from the multi-slice surface dose measurement for the same mAs per tube revolution, provided there is no significant 'overscan' beyond 360^0 for the standard axial scans.

In order to avoid inaccuracies due to the troughs and peaks of the surface irradiation, for spiral data collection, as for consecutive slice scanning, the length of TLD's must match an integral number of nominal slice thicknesses for a pitch of 1. For pitches greater than 1, the TLDs cover a length equal to the nominal width times the pitch factor (figure 26).

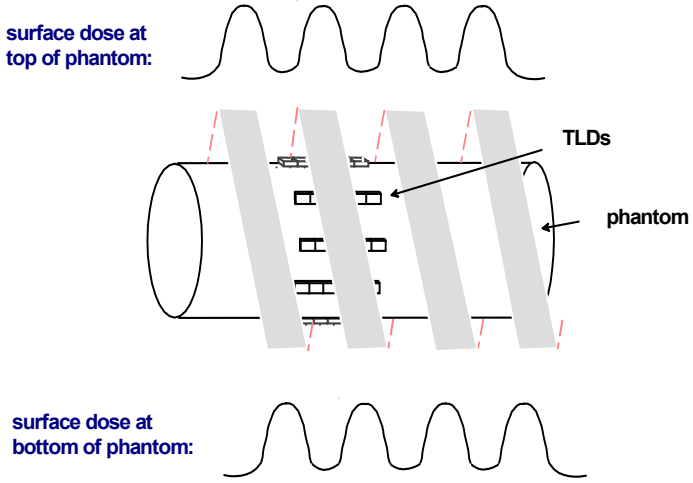


Figure 26. Surface dosimetry for spiral irradiation at a pitch of 2.

■ Dose Efficiency and Imaging Performance Parameter (Q)

Statistical noise, spatial resolution and slice width are the fundamental parameters which describe the amount of object information retrievable from an image, or its image quality. X-ray dose can be regarded as a cost of this information. In general, it is meaningless to quote any one of these measurements without reference to the others. The Q-value incorporates the dose, noise, spatial resolution and slice width into one number. This figure has been derived from a relationship between image quality and dose received[16,17,18], and is defined as follows:

$$Q = \sqrt{0.9 \times \frac{f_{50\%}^{3}}{\sigma^2 z_0 D}} \qquad \textbf{Equation 4}^{b}$$

where z_0 is the nominal slice thickness in cm, σ is the percentage noise, $f_{50\%}$ is the 50% value of the modulation transfer function in cycles/cm, and D is the dose in water in mGy.

The Q-factor is in part empirical and it should be used with caution. It is not an absolute figure. Its derivation relies on assumptions of the shape of convolution filter used. Comparisons between scanners will be more reliable when comparing scans reconstructed with similar convolution filters. It is of most importance when considering the standard scans for head or body. The uncertainty in this value is about 15%.

[b] The equation given in the Comparison Report Issue 6[1] (STD/88/36) uses exposure (in Roentgen) not dose, hence the inclusion of the factor of 0.9 to ensure the consistency of the quoted Q values within our reports.

A dose efficiency factor can take a number of forms depending on how the various parameters are measured and quoted. The basic approach is that a dose efficient scanner will produce good resolution at minimum dose and noise.

Starting in 1998, a new Q value will be introduced into ImPACT CT reports.

One of the changes is that the nominal slice width is replaced by the full width half maximum of the measured imaged width (z-sensitivity). This is more true to the original derivation of Q, since the effect on noise from the thickness of the slice is from the imaged, as opposed to the nominal, slice width. Where the imaged width is very close to the nominal slice width the Q will not be significantly affected, however it is particularly important to use the imaged slice in spiral scanning as the imaged slice thickness can be quite different from the nominal slice thickness because of the dependency on both pitch and interpolation algorithm.

Other changes to the calculation of Q involve the dosimetry and resolution parameters. The dose value used in the original formulation of Q is the surface dose to a phantom. However a cross-sectional dose value is more representative of the overall dose to the phantom or patient. The weighted average of the CTDI (described in the next section) as measured in standard perspex phantoms will be introduced into the Q equation.

Rather than using a single value to represent the resolution, a more complete description of the resolution takes into account the full modulation transfer function over all frequencies. A new resolution value based on an average of the 50% and the 10% values of the modulation transfer function will be used.

Any new Q values will be clearly indicated in all reports to prevent any confusion with the existing value currently used.

Computed Tomography Dose Index (CTDI)

■ Introduction

The Computed Tomography Dose Index (CTDI) is a measure of the total dose from a single slice irradiation. It has a general mathematical definition[15] whose details, for practical reasons, change for different applications[19]. These applications of the general formula differ in terms of the integration limits and medium of measurement.

The general formula is given as

$$\mathrm{CTDI} = \frac{1}{n.T}\int_{z1}^{z2} D(z)dz \quad \textbf{mGy} \qquad \textbf{Equation 5}$$

Where z_1, z_2 = the limits of integration
$D(z)$ = the single slice dose profile
T = the nominal slice thickness in cm
n = the number of slices irradiated simultaneously (usually n = 1, except for early EMI scanners and one current manufacturer where n = 2).

ImPACT has adopted a notation using suffixes to identify the integration distance, the phantom in which the measurement is made and, where the quoted dose value is for a medium different to that of the phantom, this too can be given.

$$\mathrm{CTDI}_{\Delta Z,\ \mathrm{phantom},\ \mathrm{medium}}$$

Where
ΔZ = integration distance
phantom = measurement in specified phantom, or air
medium = medium of quoted dose, if different from measurement medium.

In some cases, an alternative subscript may be used which encompasses both the integration distance, the phantom, and the quoted dose medium:

$$\mathrm{CTDI}_{\mathrm{label}}$$

Where label = an incorporated definition of ΔZ, phantom, medium.

An example of this is the $\mathrm{CTDI}_{\mathrm{FDA}}$.

■ ImPACT CTDI Measurements

A 10 cm long pencil ion chamber irradiated by a single slice across its centre effectively measures the integral of the single slice dose profile over a distance of 10 cm. This measurement can be used to calculate the $\mathrm{CTDI}_{10\mathrm{cm}}$. ImPACT measure this either in air, given as the $\mathrm{CTDI}_{10\mathrm{cm,air}}$, or in the perspex CT phantoms, where, using the letters CTPX to indicate these phantoms, the notation $\mathrm{CTDI}_{10\mathrm{cm,CTPX}}$ is used.

■ $CTDI_{10cm,air}$

The $CTDI_{10cm,air}$ is primarily measured at two positions in air using a single 360^0 rotation scan. The first position is at the axis of rotation (figure 27), and the second is at 10 cm off axis in the scan plane. This is most simply achieved by raising the couch 10 cm, but where there are limitations in the couch height range, the chamber is moved sideways.

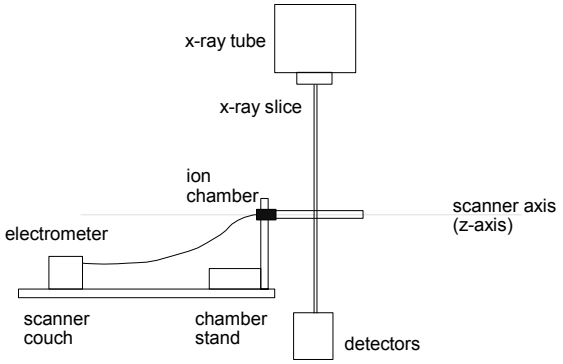


Figure 27. Lateral view of chamber position for $CTDI_{air}$ at the iso-centre.

The first measurement position records the output of the tube through the central axis in the scan plane, and the second records an integral of the off-axis output of the tube. This second measurement has evolved through the need to characterise the beam shaping filter as well as the output of the tube. By placing the chamber off-axis, as the tube rotates a measurement is made of the tube output through a continuously varying region of the beam shaping filter. (A schematic view of this position can be seen in figure 31). This integral effect can be used to characterise the beam shaping filter. If there is no beam shaping filter the off axis and on-axis values, for an axial 360^0 scan, should be nearly identical.

Another, more direct, way of characterising the beam shaping filter is to perform axial and off-axis measurements with a stationary tube. As this is not always possible without an engineer present the same effect is achieved using the scan projection radiograph (SPR). This is described in a later section on the SPR.

For in air measurements the chamber is calibrated using a diagnostic X-ray set operating at 125 kVp with extra filtration resulting in a beam with a half value layer of 8.6 mm Al. Doses are quoted as mGy in air using an f-factor of 8.76 x 10^{-3} mGy/mR. See Appendix 2 for further information.

The $CTDI_{10cm,air}$ is calculated as follows :

$$\mathrm{CTDI}_{10cm,air} = R \times cf_E \times 8.76 \times 10^{-3} \times \frac{L}{T} \ \mathbf{mGy} \qquad \textbf{Equation 6}$$

Where	R	= ion chamber reading in 'mR'
	cf_E	= calibration factor of ion chamber at appropriate energy (55 keV).
	$8.76x10^{-3}$	= factor to convert exposure (mR) to dose in air (mGy)[21]
	L	= length of ion chamber (= 10 cm)
	T	= nominal slice thickness (in same units as L; cm)

■ $CTDI_{10cm,CTPX}$

The $CTDI_{10cm,CTPX}$ is obtained from measurements using a 10 cm chamber within the 16 cm diameter or 32 cm diameter perspex phantoms made to the specifications of the FDA[4] with some additional modifications to assist in alignment[6]. See figures 6-9. The phantom is placed at the scanner isocentre (figure 28) and measurements are made at the centre of the phantom and at the periphery positions with sometimes a surface measurement being made as well.

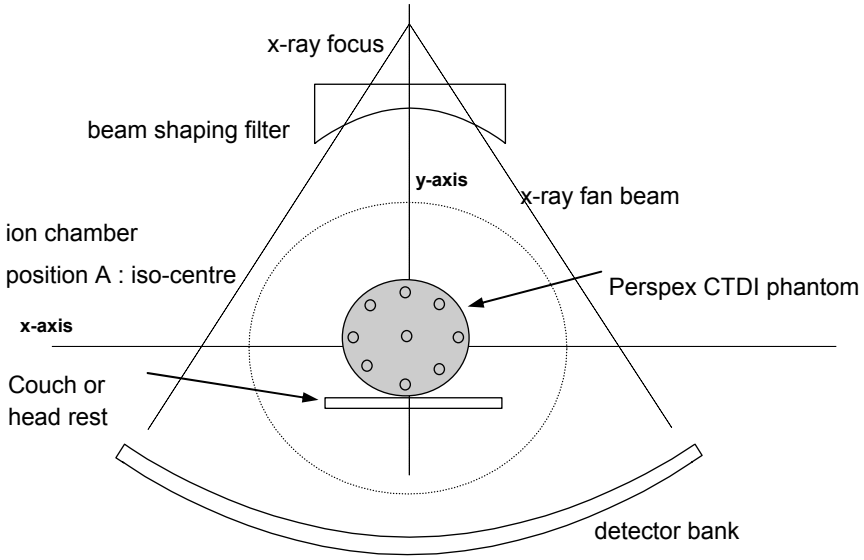


Figure 28. Position of perspex CTDI phantom in gantry.

The $CTDI_{10cm,CTPX}$ is calculated as follows :

$$CTDI_{10_{cm,CTPX}} = R \times cf_E \times 7.8 \times 10^{-3} \times \frac{L}{T} \text{ mGy} \qquad \textbf{Equation 7}$$

Where R = ion chamber reading in 'mR'
cf_E = calibration factor of ion chamber at appropriate energy (70keV).
7.8×10^{-3} = factor to convert exposure (mR) to dose in perspex (mGy) at 70 keV[21,22]
L = length of ion chamber (= 10 cm)
T = nominal slice thickness (in same units as L; cm)

A single value, called the weighted CTDI ($CTDI_w$), can be quoted, which is a weighted average of the centre and periphery values, where one third of the centre value and two thirds of the periphery value is used[39] (equation 8). In this instance the suffixes indicate the measurement position within the perspex CT phantoms.

$$CTDI_w = \frac{1}{3} CTDI_{10cm,centre} + \frac{2}{3} CTDI_{10cm,periphery} \text{ mGy} \qquad \textbf{Equation 8}$$

where $CTDI_{10cm,\ centre}$ is measured at the centre of the phantom
and $CTDI_{10cm,\ periphery}$ is measured at the periphery of the phantom.

In the equation as given here the CTDI are measured for a given tube current and scan time (mAs), but they can also be expressed as the CTDI per mAs, to give normalised CTDI values in mGy/mAs. Where overscan (ie deliberate irradiation beyond 360^0) is used, scanners often give the nominal scan instead of the actual scan time (for example 1sec instead of 1.1 sec where, for example, a 10% overscan is implemented). If the normalised CTDI is calculated using the nominal scan time, and then used to calculate CTDI values for other, nominal or actual, mAs values, errors will be introduced in the quoted doses.

Care is taken in ImPACT reports, where possible, to present data as the CTDI normalised to the true, rather than the nominal, mAs.

For the phantom measurements, the chamber is calibrated using a diagnostic X-ray set operating at 125 kVp with extra filtration resulting in a beam with a half value layer of 11.6 mm Al. Doses are quoted in mGy, as dose in perspex using an f-factor of 7.8×10^{-3} mGy/mR. See Appendix 2 for further information.

■ CTDI for Spiral Scanning

The CTDI cannot be measured for spiral scanning since the definition involves a single slice irradiation.

Comparative data can however be easily achieved using the 10cm ion chamber. A spiral irradiation and a conventional multi-slice examination are undertaken covering the whole length of the chamber (figure 29). Parameters are matched to be equivalent, for example the same mAs per revolution is set, and the pitch is made equivalent to the slice to couch increment ratio. (Normally a pitch of 1 for the spiral irradiation , and contiguous slices for the multi-slice irradiation).

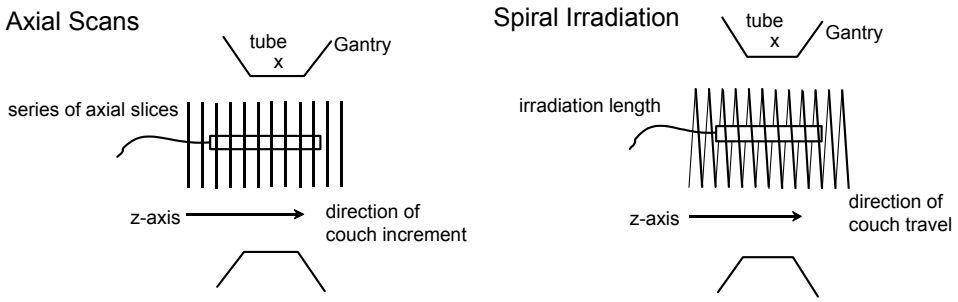


Figure 29. Comparison dosimetry between axial and spiral irradiation.

In the ImPACT assessment, comparisons as outlined here are performed.

With matching parameters the two sets of data would normally agree. However the spiral irradiation will give a lower value where the single slice irradiation incorporates some ‘overscan’, i.e. where, for each slice, the tube irradiates beyond the nominal 360 degrees.

Sometimes this discrepancy can be picked up from the scan parameters, where a nominal time is set, and the actual scan time, which takes into account the 'overscan', is indicated somewhere on the monitor. The dosimetry between the standard axial scan and the spiral scan can then be easily related. On some scanners the 'overscan' on the standard axial scan is minimal, however, on other scanners there is a deliberate overscan of 10 to 36 degrees, usually for body scans, which is utilised to correct for artefacts due to patient movement.

Discrepancies could also arise where there are inaccuracies in couch incrementation (standard axial scans) or in couch speed (spiral irradiation).

The measurements described here are essentially two multi-scan average dose measurements where the dose has been averaged over a measurement distance of 10cm, and the same couch increment is used, whether per slice for the standard axial irradiation or per tube rotation for the spiral irradiation.

To further test the spiral doses the chamber can be irradiated using different pitches. The average dose, as given by the chamber, is expected to be inversely proportional to the pitch. What is mainly under test, with the different pitch irradiations, is the software, terminology, and other factors such as appropriate couch speed for the programmed pitch.

■ $CTDI_{FDA}$

In their Performance Standard for Diagnostic X-ray Systems[4], the Food and Drug Administration (FDA) in the USA use an integration distance of 14 times the nominal slice thickness (14T) for their adapted version of the general CTDI formula. It is measured within the specified 16 cm diameter or 32 cm diameter perspex phantoms, shown in figures 3 and 4. According to the FDA, this data would be 'the absorbed dose in PMMA (polymethylmethacrylate, i.e. perspex)[20]. It should therefore be quoted as dose in perspex and have the appropriate dose conversion factors applied.

Although not necessarily an ideal dose descriptor, almost all manufacturers quote their dose data in this format due to the regulations in the USA. Discrepancies can occur where some manufacturers do not quote this data as dose in perspex. This is addressed in each individual report. Using the notation in this report, this CTDI data is referred to as $CTDI_{14T,CTPX}$ or with the all encompassing label, $CTDI_{FDA}$.

The $CTDI_{FDA}$ is similar to the $CTDI_{10cm,CTPX}$ but with different integration distances. The integration distance of 14T only matches 10cm for a slice thickness of 7 mm. The extent of scatter within the phantom means that the $CTDI_{10cm,CTPX}$ will over-estimate or under-estimate the $CTDI_{FDA}$ by a factor which relates to the difference between 10 cm and 14T[19]. The factors are given in the section on 'Comparison with Manufacturers' Data'.

The 1994 IEC report on constancy testing for CT Scanners[5] has adopted the FDA application for the CTDI. ImPACT however consider the 14 slice thickness distance for integration to be unnecessarily difficult to measure for practical constancy testing, and suggest that a single slice axial irradiation in air ($CTDI_{10cm,air}$) is better suited for this purpose.

■ $CTDI_{NRPB}$

The computed tomography dose index as used by the NRPB (UK National Radiological Protection Board) in their National CT Dosimetry survey[23] was measured using thermoluminescent dosimeters stacked, with the narrow edge parallel to the scan plane, in a holder to measure the dose profile, and peak dose, in air. The profile and dose data was used to calculate the computed tomography dose index, with the dose quoted as absorbed dose to ICRU muscle. The integration distance, x, depended on the size of the capsule used, which in turn depended on the width of slice being measured, and was typically between 1 and 3 cm. In the terminology used in this report, this computed tomography dose index can be referred to as $CTDI_{xcm,air,ICRUmuscle}$, or, more simply, the $CTDI_{NRPB}$.

It should be noted that the $CTDI_{10cm,\ air}$ measurements quoted in the ImPACT reports are quoted as dose in air, and this needs to be taken into account to enable any comparison with $CTDI_{NRPB}$ data.

With the appropriate conversion to dose in ICRU muscle, the value of $CTDI_{NRPB}$ should match that of the $CTDI_{10cm,air}$ measured by ImPACT using the 10 cm long pencil ion chamber, provided calibration conditions for the TLD and the chamber are equivalent. This is because the dose profile in air does not extend much beyond the nominal slice width due to the absence of a scattering medium, and the different integration distances will have negligible effect on the measured data.

■ Multi-Slice Surface Dose (D) - $MSAD_{ImPACT}$

The multi-slice surface dose measurement (D) described earlier is equivalent to the multiple scan average dose (MSAD) which is a form of the CTDI. The measurement, D, is made at the central position of a series of seven contiguous slices, each of thickness T, and this is equivalent to the integration of a single slice, over an integration distance of 7T, which by definition is the MSAD[15]. This is equivalent to the CTDI where the couch increment equals the slice thickness and where the integration distance on the CTDI matches the scanned distance on the MSAD. In the notation introduced here it could be called $MSAD_{ImPACT}$, or, for a more full description, $MSAD_{7T,ImPACT,water}$.

■ Dose in Air

The 10 cm pencil ion chamber is aligned free-in-air along the z-axis of the scanner, and a scan projection radiograph (SPR) carried out. The complete chamber is covered by the length of the irradiation (figure 30). The chamber is then moved, laterally or vertically depending on the position of the X-ray tube for the SPR, and the irradiation repeated. These off-axis measurements are taken at 5 cm intervals, and the aim of acquiring them is to characterise the attenuation of the beam shaping filter (figure 31).

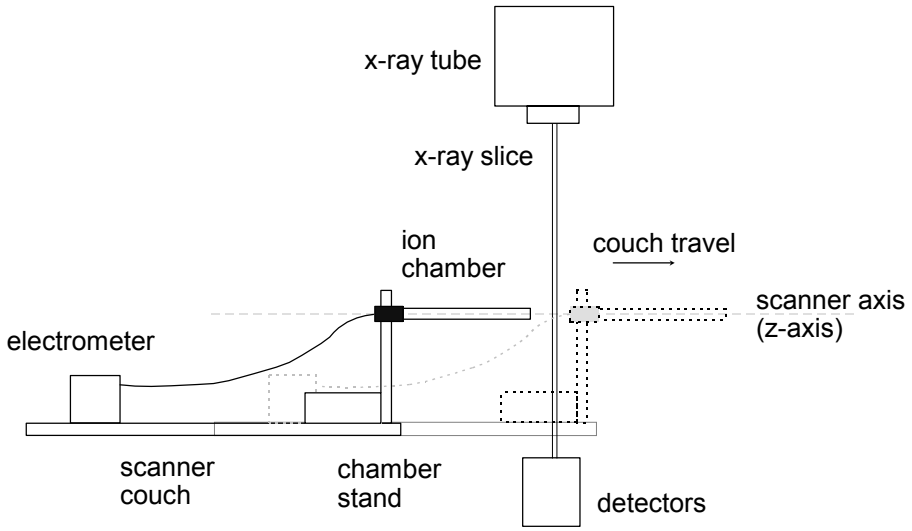


Figure 30. Lateral view for dose in air SPR measurements.

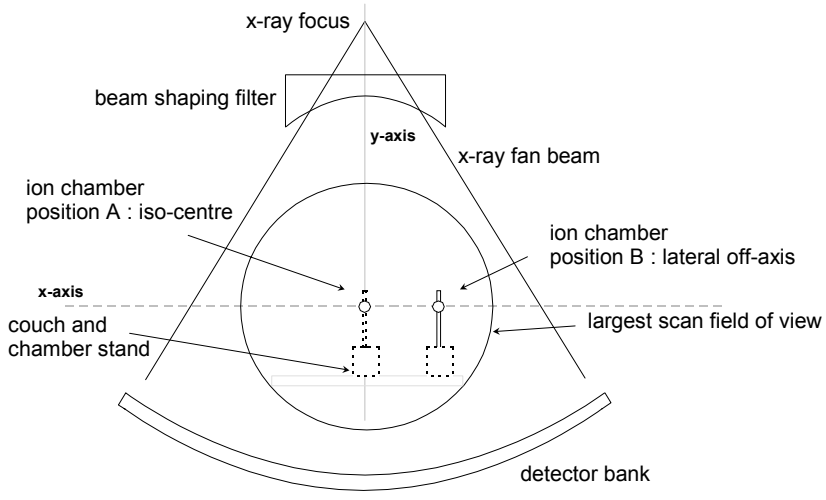
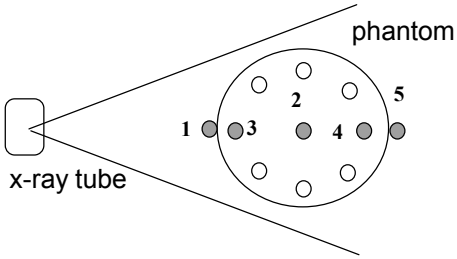


Figure 31. Cross-sectional view for dose in air SPR measurements.

■ Dose to a Phantom

The standard 32 cm CTDI phantom, is used to measure doses in a phantom from the SPR (figure 32). The SPR is performed using parameters given for a pelvimetry irradiation. The 10 cm pencil ion chamber is placed in the relevant positions in the phantom. The complete chamber is covered by the length of the irradiation. A minimum data set consists of an entrance dose and a centre phantom measurement. Other positions in the phantom can be measured, for example the 1 cm below surface both entrance and exit doses, however the exit doses are negligible.



Key:

1 Entrance: taped to Phantom surface
2 Centre phantom position
3 1 cm entrance periphery
4 1 cm periphery exit
5 Exit - taped to Phantom surface

Figure 32. Positions of ion chamber for phantom SPR measurements.

■ Distance measurements

The accuracy of distance measurements can be measured relatively simply using a grid from the LEEDS test objects which are normally used in the evaluation of fluoroscopic equipment[8], see figure 11.

The grid test object is placed horizontally on the scanner axis. An SPR scan is performed with the tube in the AP position. The grid is then moved vertically off-axis, along the y-axis, and another SPR scan performed with the tube in the same position. Figures 33 and 34.

Measurements are made on the images of the grid, along the z-axis direction and across the fan beam, in the x-direction. Figures 35-37.

The measured distances are generally calibrated by the manufacturer to be exact in the plane of the isocentre. However, systematic errors associated with field length have been observed for a scanner with pre-set scan lengths[24].

The across beam, along the x-axis, measurements will vary according to the geometrical factor from the divergent beam caused by moving the test object closer to the tube and further from the detectors. An expected change for a given distance off-axis can be calculated using the focus to isocentre distance. For example for a scanner with a tube to isocentre distance of 50 cm, a 10% change in the x-axis measurement is found when the test object is moved 5 cm nearer or further from the tube.

The z-axis measurements are not expected to change with the test object distance from the tube since this measurement is not dependent on the geometry but on couch speed, and distances in the z-axis direction can be found to be accurate within 0.5%[24].

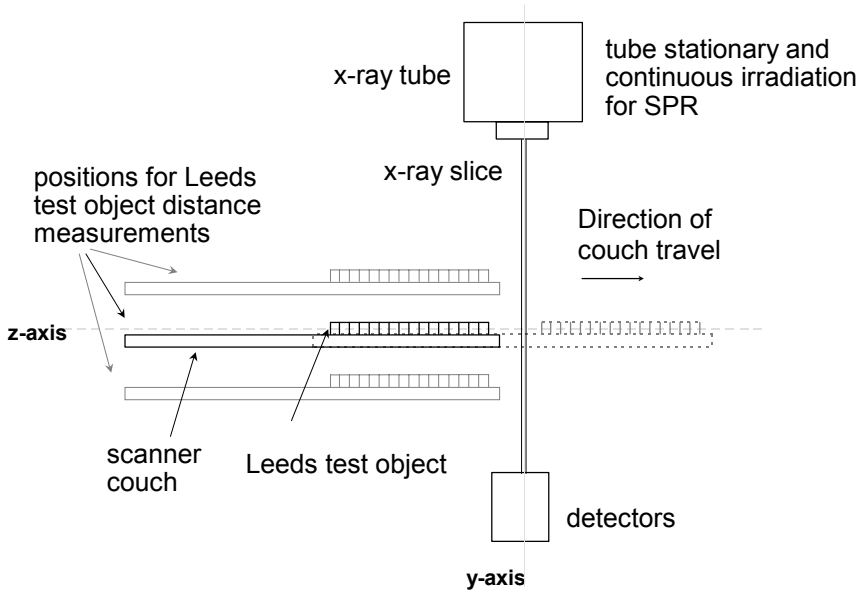


Figure 33. Lateral view for SPR distance measurement test.

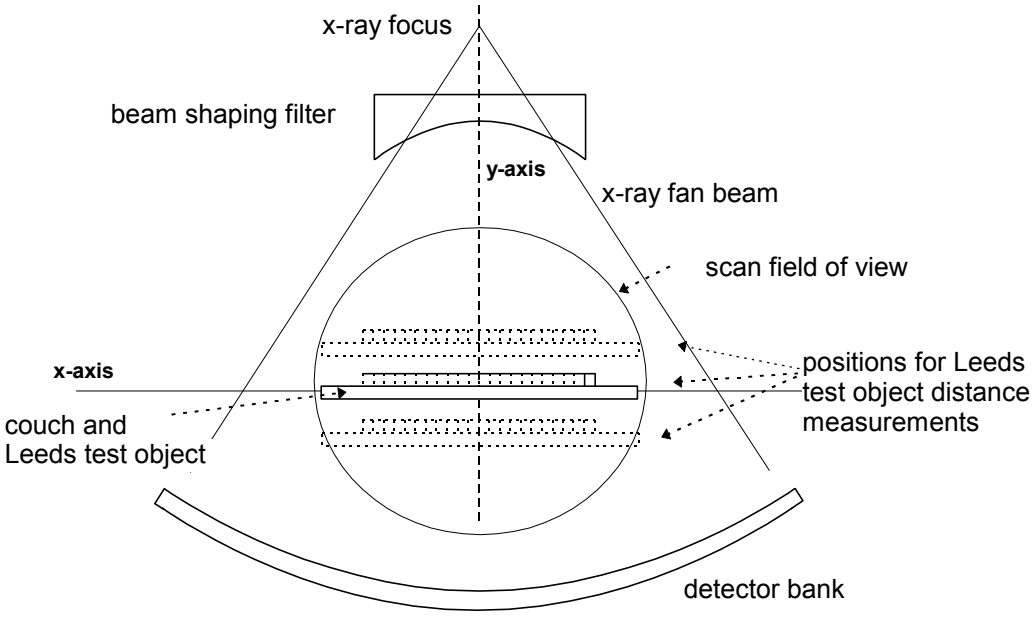


Figure 34. Cross-sectional view for SPR distance measurement test.

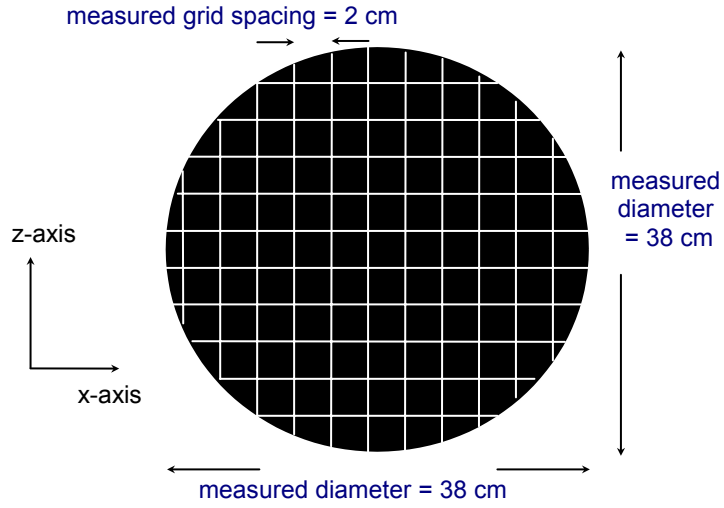


Figure 35. SPR image of Leeds test object placed at isocentre.

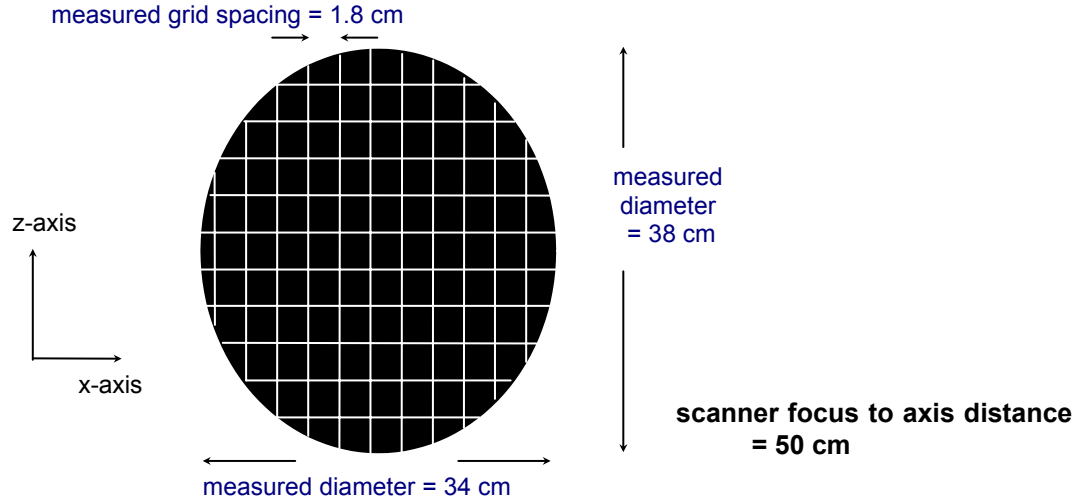


Figure 36. SPR image of Leeds test object displaced 5 cm away from tube.

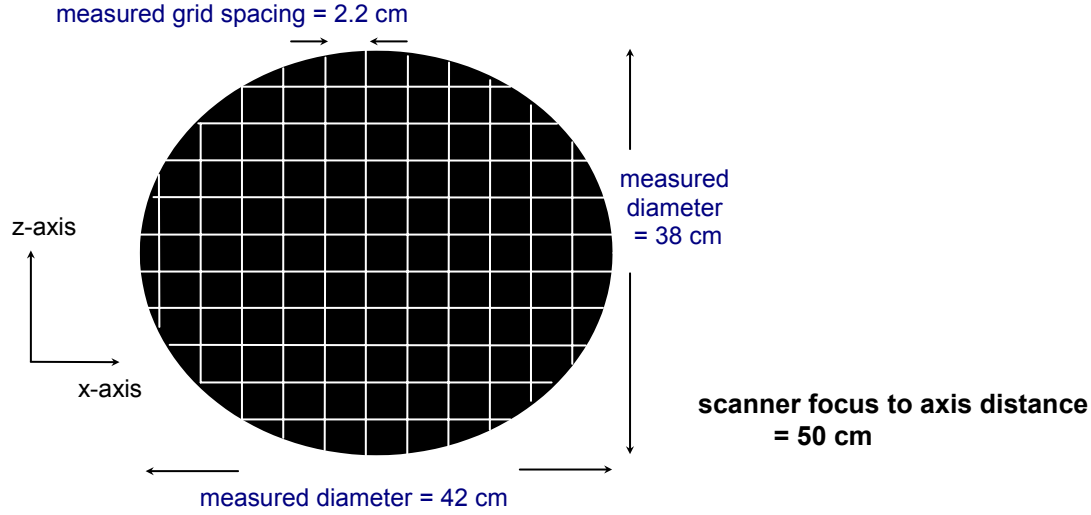


Figure 37. SPR image of Leeds test object displaced 5 cm towards tube.

Effect of Scan Parameters on Performance Parameters

■ Introduction

Scan parameters that can be altered on most scanners include scan time, mA, (together giving mAs), slice thickness, kV, reconstruction algorithm, image filter, focal spot, reconstruction field of view, and reconstruction matrix size.

In addition, some scanners have different beam shaping filters, and some non-spiral models incorporate different X-ray focus to patient distances depending on the field of view. Some scanners have an 'overscan' where the tube rotates beyond 360^0 as a standard procedure for scanning in body mode, whilst others allow this to be switched on or off. Most scanners have the ability to create an image using a 'partial scan', where the tube irradiates for less than 360^0, usually between 220^0 and 240^0.

Other functions, such as beam hardening algorithms for head, shoulder or pelvis scans, and motion correction algorithms for body scans, are either incorporated as standard features, or are provided as options to be included according to user preference. ImPACT evaluate some of these options which are then discussed in each individual scanner report, however no general comment is made in this report.

In addition to the main reconstruction algorithms that are applied to the initial attenuation data ('raw data'), many scanners offer post-processing filters that can be applied to smooth or sharpen the final displayed image for viewing. As these filters can be offered in a wide variety of forms and sometimes the images with these filters applied cannot be saved, they are, in general, not investigated or applied in the 'clinical scan' results unless they are routinely used. Comment may be made on them in the main body of the text of an individual report.

On spiral scanners there are additional parameters of pitch and interpolation algorithm which can affect image quality. The pitch is defined as the ratio of the distance of couch travel, in one tube revolution, to the nominal slice thickness. An example of a pitch of 1 is a nominal slice of 10 mm combined with a couch speed of 10 mm/sec, where a complete tube revolution occurs in 1 second.

For the reconstruction of images from spiral data acquisition there are two basic linear interpolations. These use attenuation profiles taken from up to a half (180^0) or a full (360^0) rotation of the tube either side of the required image plane. These may be referred to by the straight descriptive term e.g. '180^0 linear interpolation' (180LI), or '360^0 linear interpolation' (360LI). Alternatively a manufacturer may have a particular terminology such as 'slim' and 'wide', or 'interpolation 1' and 'interpolation 2'. In addition to the two basic linear interpolations, non-linear, higher order interpolations may also be available and are sometimes used as standard. The 360^0 linear interpolation is not often used as standard as it gives rise to a significantly wider imaged slice width than the corresponding standard axial scan using the same collimated slice width.

■ Noise

The random statistical noise arising from the X-ray exposure and detection is inversely proportional to the square root of the mAs, and hence also inversely proportional to the square root of the dose[25]. Deviations from this relationship may be due to electronic noise[26] or a deliberate modifying function applied by the manufacturer[27]. See figure 38.

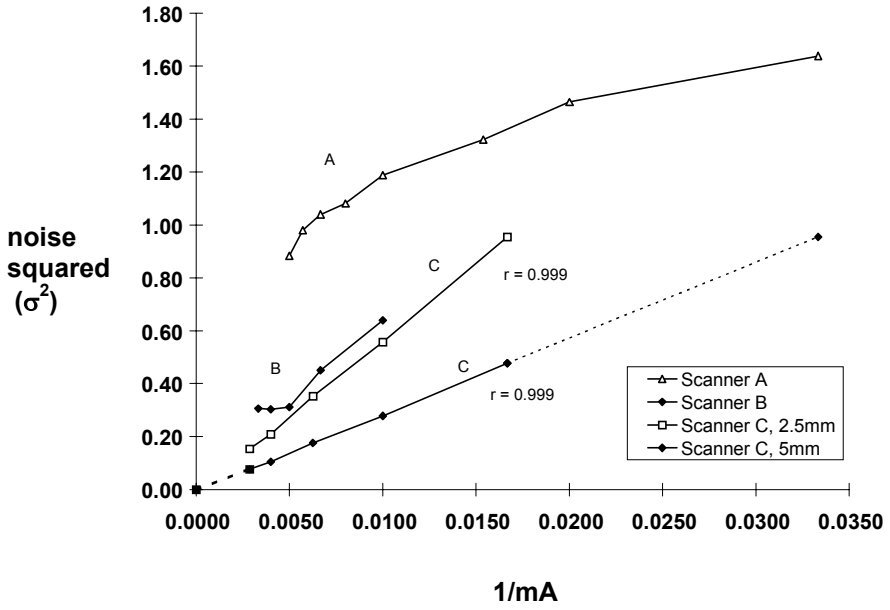


Figure 38. Noise versus mA relationship for 3 scanners.

Convolution filters used in image reconstruction will affect noise, with the softer (lower resolution) filters giving reduced noise relative to the sharper (higher resolution) filters.

The definition for noise utilises the difference in CT number between water and air, which is normally 1000. However in certain instances, particularly with the use of high spatial resolution convolution filters, this range may be compressed to 500 or 250.

For perfectly rectangular imaged slice profiles, noise will be inversely proportional to the square root of the slice thickness[25].

The matrix size may affect the noise on the displayed image where the reconstruction matrix does not match the display matrix and interpolation is used between the two matrix sizes.

Whilst it might be expected, for all other parameters the same, that image noise increases with smaller reconstructed fields of view (and therefore smaller pixel sizes), in practice ImPACT evaluations do not always find this to be the case. On some scanners, and for some filters, noise is unaffected by the reconstructed field of view, however on other scanners, or with different filters, the noise may either increase, or decrease, with smaller fields of view. Individual scanner reports need to be consulted.

■ Spiral scanning

Differences in noise values between those from the conventional slice scans and those from the spiral data collection, for the same mAs per revolution and the same convolution kernel or filter, will be determined by the z-axis interpolation algorithms used for reconstructing the spiral scans. The noise values from the 360^0 linear interpolators (360LI) will be lower than from the standard axial scans and the noise from the 180^0 linear interpolators (180LI) will be higher[12,13], see table 2.

The noise will be even higher when using a 180^0 higher order interpolation algorithm. An example of a higher order algorithm is also given table 2. The extent of effect from a higher order algorithm will differ from manufacturer to manufacturer.

Differences in noise values between spiral and non-spiral images, beyond those determined by the z-axis interpolator, may simply occur where the reconstruction algorithms applied for the image reconstruction from the spiral imaging are different from those for the standard imaging.

Table 2. Relative noise values: spiral scans reconstructed with different interpolation algorithms[12,13].

Interpolation Algorithm	Relative Noise
-(axial)	1.00
360 LI	0.83
180 LI	1.12
180 HI	1.29

Noise values, for images from spiral data acquisitions of a uniform water region, do not change with pitch (all other parameters such as interpolation algorithm, mA etc being equal). Examples for pitches less than 1 are given in table 3, however the same values would be expected for pitches greater than 1, since the same principle (noise being dependent on dose to the detectors) applies regardless of the pitch. However scanners with more than one detector bank may not follow this principle, since data from adjacent detector banks could be theoretically interchanged. Individual scanner reports need to be consulted.

Table 3. Relative noise values: spiral scans scanned with different pitches (360LI example given)[13].

Pitch	Relative Noise
0.0 (axial)	1.00
0.2	0.83
0.5	0.83
0.8	0.82
1.0	0.83

A single reconstructed image from a spiral acquisition may have non-uniform noise values around the periphery of the image[12,28]. The non-uniformity is a similar effect to that observed for partial conventional axial scans.

The extent of this variation depends on the type of interpolation algorithm[28]. The reference states that the maximum to minimum noise ratio within an image is about 1.12 for images from the 360^0 linear interpolation (360LI) algorithm to about 1.20 for images form the 180^0LI. Images from higher order 180^0 interpolations give ratios of about 1.2 or higher. A standard axial scan has no variation (ratio of 1) whilst a partial scan has a ratio greater than 1.5.

This non-uniformity changes its distribution within the image according to the z-axis position of the reconstructed slice, and is thought to depend on the angular position of the tube relative to the z-axis position of reconstruction. In a sequence of reconstructed images from a spiral acquisition, reconstructed at z-axis increments less than the distance moved by the table in one tube rotation, the noise value at a particular position within the set of images will appear to change in a cyclical manner[12].

ImPACT has not yet investigated these findings extensively, though will be including some analysis in scanner assessments. Results from one scanner[29] show this non-uniformity, and resulting cyclical variation, not only at the periphery of the phantom image, but also at the centre of the image. This is the data shown in figure 39. This variation is still thought to depend on the angular position of the tube relative to the z-axis position of reconstruction, but caused at the centre of the image by extra attenuation of the primary irradiation by the couch top as the tube travels under the couch.

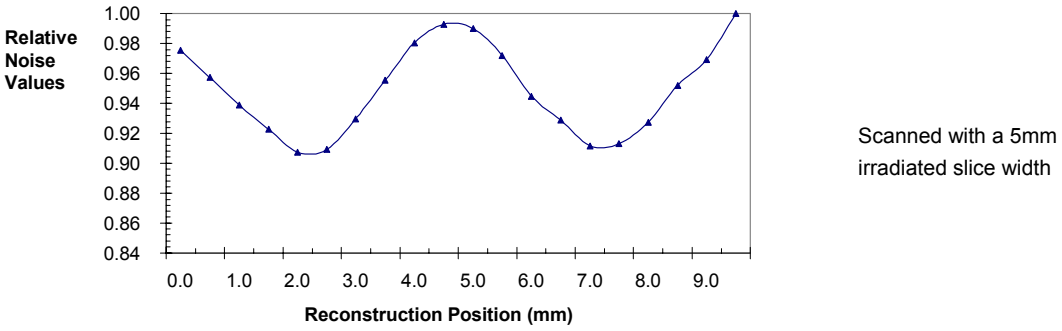


Figure 39. Relative noise values for images reconstructed at different z-axis positions from spiral data.

■ Spatial Resolution

Convolution filters affect spatial resolution, sharper filters giving increased resolution relative to the softer or standard ones.

Spatial resolution may be affected by scan time where the number of data samples collected is dependent on scan time. The spatial resolution may therefore increase with longer scan times, however, this may only show in the limiting resolution case of the high resolution algorithms.

The focal spot size affects the resolution, and where two focal spot sizes are available the highest spatial resolution will be achieved with the small focal spot.

Intrinsically the matrix size should not affect the spatial resolution unless the resolution of the image is limited by the size of the pixels. To avoid this loss of spatial resolution, the pixel size can be reduced by reconstructing with a smaller field of view or, if available, using a larger reconstruction matrix to increase the numbers of pixels.

In the method of resolution analysis used by ImPACT[10], however, there is rarely a pixel limitation since the method of using an inclined plane across the matrix enables data points to be acquired at less than one pixel width intervals. The analysis therefore gives MTF values beyond the frequency values limited by the pixel size (the Nyquist frequency). The values quoted, however, can therefore be slightly misleading, as resolutions quoted beyond the Nyquist frequency are not accurately represented in the final image matrix.

In practice, the perceived resolution is not usually limited by the pixel size for standard resolution filters. A head field of view, displayed on a 512 matrix, has a pixel size of about 0.5 mm, and a corresponding Nyquist frequency of about 10 cycles per cm. A typical standard head scan has a 10% MTF value of about 7 cycles per cm, which means that the Nyquist frequency falls nicely just beyond the extent of the pre-sampling modulation transfer function (see figure 17). A typical body field of view (350 mm, 512 matrix) has a pixel size of about 0.7 mm with a corresponding Nyquist frequency of 7 cycles per cm. This is just greater than the 10% modulation transfer function value, of about 6 cycles per cm, from a typical standard body scan. The Nyquist frequency is therefore about equal to the MTF cut-off frequency for these types of scans. With higher resolution convolution filters than these, the field of view needs to be reduced to prevent this pixel size, Nyquist frequency, limitation.

■ Spiral scanning

The method of measurement of image plane resolution used by ImPACT should not be affected by spiral scanning since the test object is uniform along the z-axis. Therefore, for the same reconstruction or convolution filter one would expect the same scan plane resolution for both spiral and axial scanning. However, differences in the scan plane resolution, between conventional and spiral scanning, may occur where the reconstruction kernels applied to the interpolated spiral data are different from those used in the standard axial imaging.

In non spiral CT the spatial resolution will be reduced towards the periphery of the scan field relative to the centre. For a full 360^0 scan there should be spatial symmetry about the axis within an image, for a partial scan, however, this spatial symmetry is not present[12,28]. In spiral scanning, using a 360^0 linear interpolation, one might also expect spatial symmetry within a reconstructed image, but not so for a 180^0 interpolation[12,28]. This is found to be the case, with approximately a 2% variation for the 180^0 linear interpolation given in the reference[23], between 4% and 15% for non-linear 180^0 interpolations, and about 20% for a partial standard axial scan.

In a similar manner to the noise, the distribution of this non-uniformity of resolution across an image depends on the angular position of the tube relative to the z-axis position of reconstruction. In a sequence of reconstructed images the resolution, at an off-centre position in the image, therefore changes according to the z-axis position of the reconstructed slice[28]. As with the cyclical noise variation ImPACT has not yet investigated these findings extensively. Results from one scanner however do show a small cyclical variation of the resolution, using the resolution test object which is situated in an off-axis position[29] (figure 40).

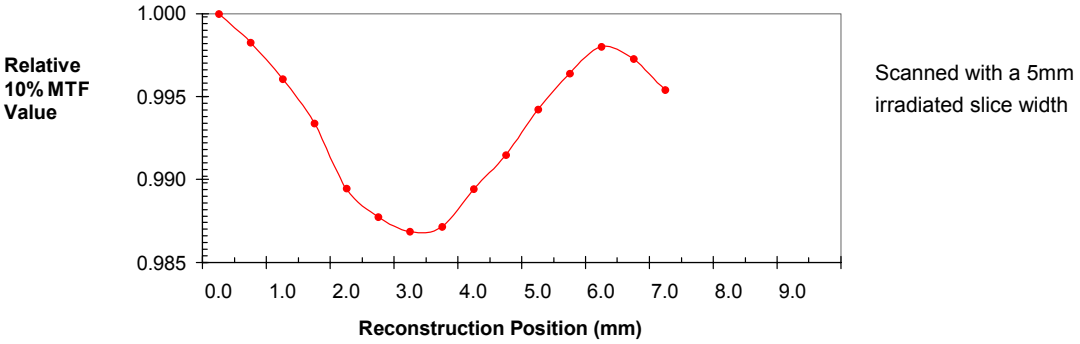


Figure 40. Relative resolution values: images reconstructed at different z-axis positions from spiral data.

■ Z-Sensitivity - Imaged Slice Thickness

This is mainly affected by the nominal slice thickness set. In its measurement it may be affected by the field of view where a geometrical shift of the X-ray tube relative to the isocentre is employed, requiring different settings to collimators.

The convolution filter should not affect this measurement. A standard algorithm (neither too sharp nor soft) is used as preference when evaluating the z-axis sensitivity.

The mAs value, whilst not intrinsically affecting the z-axis sensitivity, influences its measurement since at low mAs values image noise increases and, therefore, the error.

A small field of view (together with a large matrix if there is a choice), to give a small pixel size, provides the best spatial sampling for assessing this parameter.

■ Spiral scanning

The imaged slice width in spiral scanning is partially determined by the nominal slice width, which defines the width of both the pre-patient and post-patient (if used) collimators for the irradiation. It is also determined by the pitch and the interpolation algorithm. The interpolation algorithm affects the image width both in terms of how many rotations either side of the required imaged plane are used for the interpolation, and also by whether a linear or weighted interpolation algorithm is used.

The imaged slice width increases with pitch. This is because data is collected over a greater length to create the image (figure 41).

For the same pitch and collimation an image created from a 360^0 linear interpolation will be wider than that from a 180^0 linear interpolation because data is collected from a greater distance either side of the required image plane (figure 42). Non-linear interpolations result in narrower imaged slices relative to the linear counterpart. For example the imaged slice produced by a 360^0 non-linear interpolation will be narrower than that from the 360^0 linear interpolation, and may even be narrower than that from the 180^0 linear interpolation.

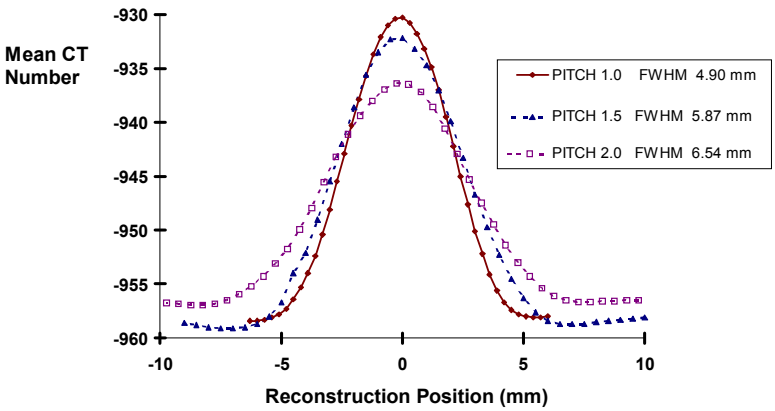


Figure 41. Z-axis sensitivity profiles: images from spiral data acquisitions at different pitches.

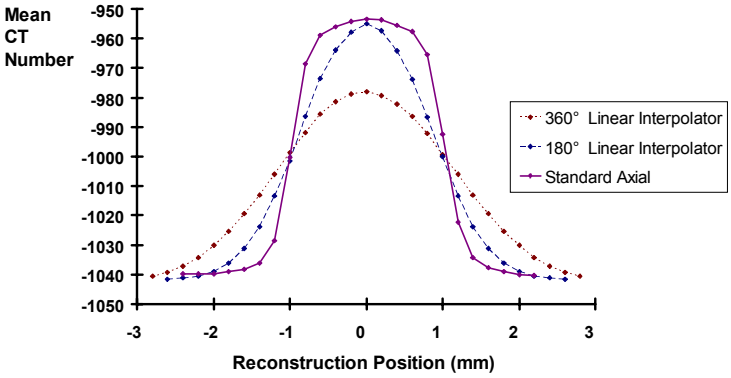


Figure 42. Z-axis sensitivity profiles for a standard axial scan and images from a spiral data acquisition (180LI and 360LI).

The imaged width of a slice reconstructed from a spiral acquisition can be predicted for different pitches and for the common interpolation algorithms[13]. Data is given for one slice width (5 mm) from the reference, but the same ratios can be used for other slice widths, and is presented for two other slice widths (table 4).

Table 4. Spiral imaged slice widths[13].

Nominal Slice FWHM (mm)	Imaged Slice FWHM (mm)				
	Axial	Spiral Pitch 1		Spiral Pitch 2	
		360LI	180LI	360LI	180LI
2	2	2.5	2	4.3	2.6
5	5	6.3	5	10.8	6.5
10	10	12.6	10	21.6	13

■ Dose Profile FWHM - Irradiated Slice Thickness

This is affected by the nominal slice thickness. The exception is where post-patient collimation is used to achieve a 1 mm or 2 mm imaged slice from a wider irradiated slice. The measured irradiated slice width for the 1 mm slice will then be the same as for the wider slice. For 1 mm slices one might expect the manufacturer to use the next largest irradiated slice width, usually the 2 mm, however in some instances a 3 mm irradiated slice width is used. This has obvious dose consequences which is readily picked up in the CTDI measurement.

The irradiated slice width may be affected by the field of view where a geometrical shift of the X-ray tube relative to the isocentre is employed, requiring different settings to collimators to achieve the same required width at the isocentre. For a properly calibrated system the measurement should not change.

Where there are two focal spots, the smaller focal spot will cause the irradiated profile of thin slices to be more clearly defined as there will be less penumbra. (The focal spot size is often selected automatically according to the setting of parameters such as slice width, algorithm, mAs, or a combination of these).

The dose profile is measured using film and care is taken in selecting mAs values used for exposures, since the measurement accuracy of the dose profile from the film can be affected by too high an exposure.

■ Spiral scanning

The irradiated slice width is a measure of the collimation, and is usually set by choosing the nominal slice width. In most spiral scanners the nominal slice width is selected in the same way (even though the final imaged slice width depends on the pitch and the interpolation algorithm), therefore the irradiated width should be the same as for the conventional axial slice.

ImPACT do not routinely check this. However, on some scanners there may be a deliberate change in collimation, for example to a narrower slice to conserve dose with pitches less than one, and ImPACT will investigate to verify. In this instance film is used, or a series of chamber measurements to give relative information.

■ CTDI and Multi-Slice Dosimetry

Dose should be directly proportional to mAs.

The multi-slice surface dose is largely unaffected by nominal slice thickness where there is close agreement of the irradiated slice width with the nominal slice thickness. However, where there is only post-patient collimation achieving the narrower imaged slices there will be significant dose increase. For example the dose is usually doubled for a 1 mm slice with post-patient collimation since the pre-patient collimation is as for the 2 mm slice.

On the majority of scanners the dose is unaffected by field of view, however, there are scanners with particular features associated with the field of view which do affect the dose. These are identified in each report and include features such as geometrical shift, physical collimation of the X-ray fan beam, asymmetrical fan beams, or different flat or beam shaping filters.

Where a scanner irradiates beyond the nominal 360^0 for a conventional axial scan, this is demonstrated in the dose distributions given around the phantom.

The kV selected will affect the dose. The dose tends to increase by a factor of about 1.4 for irradiations from 120 kVp to 140 kVp. The specific factors are given in each report.

The multi-slice doses are affected by changing the couch increment relative to the slice width. The slice width divided by the couch increment can be called the packing factor. The multi-slice average dose is proportional to this packing ratio.

■ Spiral scanning

Spiral scanning will not normally affect dose compared to standard axial scanning where the same mAs per tube revolution is used, the irradiated length is the same, and where the pitch for the spiral scan is equivalent to the packing ratio on the standard axial scan. For example a pitch of 1, and a packing ratio of 1.

The exception is for standard scans where an overscan is used. The mAs stated is often for the 360^0 of rotation and does not take into account the longer exposure due to the overscan. In these situations the standard axial scan will have a non-uniform dose distribution around the phantom, with a higher average dose.

For pitches other than 1 the average dose over an irradiated length will be inversely proportional to the pitch. For example, a pitch of 0.5 will double the average dose, a pitch of 1.5 will give a net effect of 67% of the original dose, and a pitch of 2 should halve the average dose.

For scanners with more than one detector bank care must be taken to correctly interpret the manufacturers definition of pitch. This will be discussed in the relevant reports.

For a general discussion on spiral dosimetry see Appendix 1.

■ Scan Projection Radiograph (SPR)

Parameters such as tube current (mA), irradiated slice thickness and table speed need to be considered for the SPR dose measurements. An increased table speed will decrease the dose by an amount inversely proportional to the table speed. Dose is proportional to the mA and for a constant table speed and mA the dose will be higher for wider slice thicknesses.

The measurement of distances should be independent of all parameters except the object distance from the tube (couch height for an AP or PA SPR). The system is calibrated for all measurements to be correct in the horizontal plane of the isocentre. However, even in this plane, systematic errors have been observed for a scanner with pre-set scan lengths[24].

Comparison with Manufacturers' Data

■ Introduction

Manufacturers use a wide variety of phantoms and techniques for quoting their performance parameters, and these may not be directly comparable with values obtained from the ImPACT protocols.

In general, for measurements such as noise and multi-slice surface dose, it is not possible to compare the ImPACT measurements with the specification data supplied by the manufacturer, due to differing phantom sizes and composition. Manufacturers use a number of different phantoms for their full specification data set, consisting of a combination of their own and commercially available phantoms.

■ Noise

Noise is dependent on phantom size and composition, and ImPACT values cannot, therefore, easily be compared with the manufacturer's data.

■ High Contrast Spatial Resolution

These results are usually more easily comparable since the spatial resolution is not dependent on the phantom size. The only exception is where the phantom size unnecessarily contributes to noise on the image which then gives rise to large errors in the determination of the modulation transfer function.

Different analysis techniques, however, can give variations in results, particularly for high resolution algorithms[3].

■ Z-Sensitivity (z) - Imaged Slice Thickness

Imaged slice thickness (z-sensitivity) measurements are generally independent of the phantom used, and the technique of inclined plates or wires for standard axial scans is fairly universal. The full width at half maximum (FWHM) of the z-sensitivity profile, at or near the centre of the scan field, is the basis of most manufacturers' nominal slice thickness data.

Direct comparison can, therefore, be made between ImPACT and the manufacturers' measurement of this parameter.

■ Dose Profile FWHM - Irradiated Slice Thickness

This is usually specified by the manufacturer at the isocentre in air. Measurements made by ImPACT are now also in air at the iso-centre, enabling direct comparison[c].

[c] In earlier reports measurements were made on the surface of the DH/ImPACT phantom. To compare with manufacturer's data, correction factors to the ImPACT data had to be applied in order to give an estimate for the in air at isocentre measurement. This factor consisted of a geometrical factor due to the different tube to isocentre distance, and an additional empirical estimate, of up to 5%, which took into account the scattering medium of the phantom. This is no longer necessary to apply since data is now acquired directly in air.

■ Multi-Slice Surface Dose (D)

The ImPACT multi-slice doses are quoted as dose in water and are made on the surface of the ImPACT phantom. There is no comparable data supplied by the manufacturer.

The ImPACT multi-slice doses, carried out for 7 contiguous slices, are equivalent to the MSAD, which, where the couch increment equals the nominal slice thickness, is similar to the CTDI integrated over 7 times the nominal slice thickness. The $CTDI_{FDA}$ data, supplied by the manufacturer, is measured within a phantom of different material and dimensions to the ImPACT phantom. The integration length is over 14 times the nominal slice thickness.

Therefore due to the different phantom sizes and composition, the different measurement positions, the different distances over which the data is acquired, as well as the different media in which the doses are quoted, it is impractical to attempt to compare the near surface $CTDI_{FDA}$ measurements of the manufacturer to the ImPACT surface MSAD.

■ $CTDI_{10cm}$

The simplest parameter for measurement, and therefore for comparison of data, is the $CTDI_{10cm,air}$, however manufacturers do not generally quote this value.

The next most suitable parameter for measurement and ease of comparison is the $CTDI_{10cm,CTPX}$. Again, manufacturers do not generally quote this value. However, they do quote the $CTDI_{FDA}$ in their technical literature. All manufacturers that market their scanners in the USA will have acquired the $CTDI_{FDA}$ data for their scanner model. The difference between the $CTDI_{10cm,CTPX}$ and the $CTDI_{FDA}$ is the integration distances over which the CTDI is calculated.

The $CTDI_{FDA}$ data supplied by the manufacturer, is measured at the centre, and 1 cm below the surface, of the perspex CT phantoms. According to the FDA regulations these values should be quoted as absorbed dose in perspex, and measured over an integration distance of 14 x the nominal slice thickness. For the 1cm below surface, periphery, position usually either the maximum, or the average, dose is quoted.

To match as closely as possible to the manufacturers quoted specification ImPACT measure the CTDI in the same positions within the perspex CT phantoms and also quote as absorbed dose in perspex. However, because the integration distance of 14 x the nominal slice thickness is difficult to achieve practically, correction factors have been established to apply to the 10 cm chamber measurements in order to facilitate an easy comparison.

These correction factors have been calculated from the integrals of dose profile measurements obtained with TLDs[19,31]. The correction factor is the ratio of the two integrals, of the single slice dose profile, taken over the different integration distances. An example is given in figure 43 where for a 10mm irradiated slice the integral taken over 14cm is required for the $CTDI_{FDA}$, and the integral over a distance of 10cm is measured by the ion chamber for the $CTDI_{10cm}$.

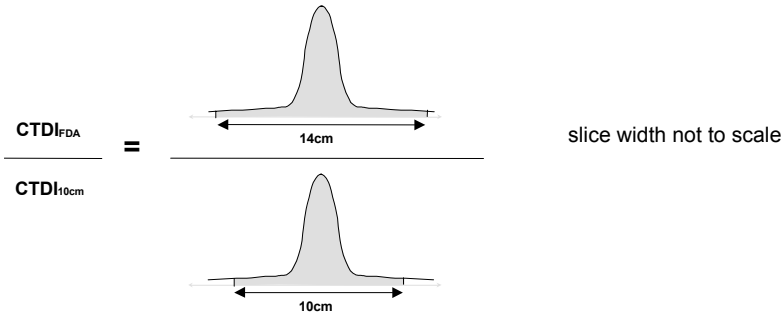


Figure 43. Correction factor for 10 mm irradiated slice width.

A straightforward comparison can then be made with the manufacturer's quoted doses of $CTDI_{FDA}$ by applying the correction factor to the $CTDI_{10cm,CTPX}$, as measured in the perspex phantoms with the pencil ion chamber. These correction factors depend on the phantom size, the position of measurement, and the slice thickness. The correction factors are shown in table 5. The labels C, P and S are the positions in the CTDI phantom as given in figure 44 (as well as figures 6 and 7), relating to the centre, periphery and surface positions respectively.

Table 5. Ratio of $CTDI_{FDA}$ to $CTDI_{10cm,CTPX}$

Ratio CTDI $_{FDA}$ / $CTDI_{10cm,CTPX}$						
Slice	**Head Phantom**			**Body Phantom**		
(mm)	**C**	**P**	**S**	**C**	**P**	**S**
10	1.10	1.05	1.04	1.16	1.05	1.04
5	0.87*	0.93	0.95	0.81	0.93	0.93
3	0.70	0.84	0.86	0.60	0.84	0.85
2	0.55	0.75	0.80	0.43	0.76	0.80
1.5	0.51	0.71	0.75	0.39	0.69	0.76
1	0.36	-	-	0.29	-	-

- Please note in some earlier reports this value was incorrectly given as 0.97.

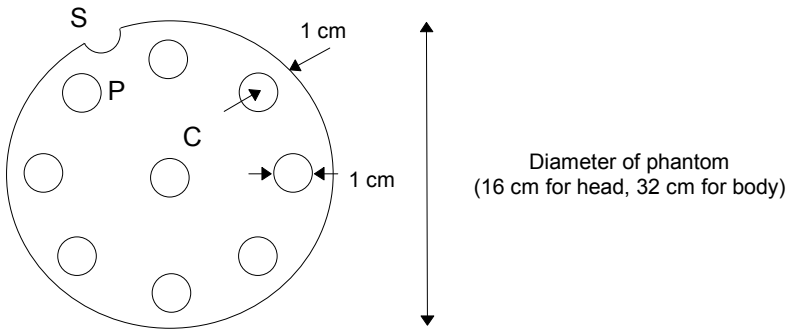


Figure 44. Positions S, P and C on the CTDI perspex phantoms.

Whilst it is useful to have the full set of correction factors, in practice since the aim is primarily to check the tube output, ImPACT usually look at the 10 mm or 5 mm slice width dose data for comparison with the manufacturer's data. Comparison of the relative slice width profiles or the relative values of the $CTDI_{10cm}$ supply the necessary information about the dosimetry arising from a series of smaller slices.

The manufacturers allow quite a large range for the $CTDI_{FDA}$ data, often +/-20%, however ImPACT generally find much closer agreement. Large differences in ImPACT measured data compared to the manufacturer's quoted data has occurred where a manufacturer has not quoted the doses as dose to perspex. This can result in an approximately 17%[30,21] discrepancy between data. This is addressed within individual reports where relevant.

Appendix 1. Comments on Dosimetry

■ Patient Dosimetry

The ImPACT reports do not supply information on specific organ doses or radiation risk to the patient. However, the multi-slice doses quoted in the ImPACT reports can be used as an estimate of patient skin doses, and the CTDI values can be used for system to system, or inter-technique, comparison purposes.

Risk to the patient is best evaluated by calculating effective dose[40]. This takes into account the extent of the irradiation (the length of patient scanned, and the inter-scan spacing or spiral pitch), as well as the weighting factors for organs irradiated. Calculation of both organ doses and effective dose has been undertaken by the National Radiological Protection Board for some older scanner models[23]. The data sets are available[32], which, upon incorporation into a suitable software package[33] , and input of the in air $CTDI_{10cm,air,ICRUmuscle}$, enable users to calculate organ doses and effective dose for individual scanning procedures. Scanners that have been introduced since the publication of the NRPB data need to be classified to match scanners for which there is data available. A project is currently in progress at ImPACT to enable this to be achieved easily.

For comparison of systems, however, the length and region of patient irradiated are not relevant since these are not scanner dependent. Data such as that presented in the ImPACT reports can be helpful for comparison of systems, or, with a simple incorporation of scan length and inter-scan spacing, of techniques.

■ Spiral Scanning Patient Dosimetry

The spiral 'multi-slice' doses quoted in the ImPACT reports can be used as an estimate of patient skin doses from a spiral irradiation. Where an equivalent scanner can be used from NRPB datasets, organ dose and effective dose can be calculated by matching the pitch to an equivalent inter-slice spacing. The $CTDI_{10cm,air,ICRUmuscle}$ can be considered as equivalent for both the standard axial and the spiral scan, provided consideration has been given to any overscan in the standard axial mode. If there is any overscan in the standard axial mode (discussed in this report in the computed tomography dose index section) then the CTDI used for the organ and effective dose calculation needs to be corrected for the overscan, as it will not be present in the equivalent spiral irradiation.

Two conflicting misconceptions have arisen about dose from spiral scanning; the first is that it delivers a lower dose compared to conventional scanning, and the second that it delivers a higher dose. From discussions in the main body of this report it can be seen that spiral scanning actually gives essentially the same dose as standard axial scanning when comparable scanning parameters are used. The two opposing views on spiral dosimetry have arisen from the way scanners are used rather from the technology itself.

The first idea that spiral scanners produce lower doses, comes from the limitations of the tubes in use. For a spiral run, a tube must be able to irradiate for longer continuous times than for non-spiral and, therefore, needs a high heat capacity or high heat dissipation rate (or both). The early spiral scanners usually used the same tube as that for the conventional non-spiral version of the scanner and those tubes were not adequate for the task. The solution was to operate the tube at a lower mAs per revolution than would have been used for the equivalent non-spiral scan, thus giving a lower dose. This still applies to some of the current lower specification scanners where the tube supplied may limit the spiral applications of the scanner. Spiral scanners can also operate with an increased pitch hence reducing the dose to the patient.

In general, therefore, for all other parameters equivalent, spiral scanners do not intrinsically give lower dose. However there is one exception to this which is where there is significant overscan (ie where the tube rotates through more than 360^0) in the non-spiral scan. Each standard axial slice will therefore be increased in dose, compared to a single tube rotation in the spiral mode, by a fractional amount equal to the ratio of the total irradiation angle to 360^0. The overscan generally increases the overall dose by between 5 and 10%. Sometimes the manufacturers give the total irradiation time exactly so this can easily be calculated.

The second view, that spiral scanners give higher dose, comes from the use of the scanner in its spiral modality. This arises from the increased capability of the scanner, particularly with the more recent higher heat capacity tubes and the newer more flexible versions of software, which allow the scanning of longer lengths more quickly as well as two passes for contrast studies. Therefore even with the same mAs per revolution as the conventional scanning situation, more of the patient may be scanned, more times, thus leading to higher dose received by the patient for the examination.

The interpretation that spiral scanning gives higher dose has arisen again not from the technology itself, but from its application.

An additional, though less common, scenario where spiral scanning may give rise to higher dose is in situations where wrong or inappropriate scanning parameters are entered for the scanning procedure. In a sequence of non-spiral scans a mistake can be identified in the first few scans and rectified for the rest of the examination. However, in a spiral irradiation a mistake can only be identified once the first image has appeared, by which time the complete irradiation is usually finished. A complete second irradiation may therefore be necessary. Whilst not a regular occurrence, this potential source of unnecessary irradiation needs to be identified so that users are aware of its possibility, and can perhaps develop procedures to prevent this scenario occurring.

Appendix 2. Calibration Conditions and Dose Calculations

■ Effective Energy of the CT Scanner

■ In air

ImPACT calibrations for in-air measurements are based on a CT scanner operating in the 120 kVp region and having an estimated half value layer (HVL) of between 7 and 9 mm Al[34]. This approximates to an effective energy range of 50-59 keV for a monoenergetic beam[30].

■ In phantom

For phantom dose measurements ImPACT calibrations are based on an effective energy of 70-75 keV[34,35,36]. This corresponds to 11-12 mm Al HVL for a monoenergetic beam[30].

■ Ion Chamber

■ For in air measurements (~55 keV)

The chamber is calibrated using a diagnostic X-ray set operating at 125 kVp with added filtration resulting in a beam with a half value layer of 8.6 mm Al. The chamber is calibrated in air against a secondary standard using the substitution method.

Scanner measurements in air are quoted in mGy, using a factor of 8.76×10^{-3} mGy/mR to convert exposure (mR) to dose in air (mGy)[21].

■ For in phantom measurements (~70-75 keV)

The chamber is calibrated using a diagnostic X-ray set operating at 125 kVp with extra filtration giving a half value layer of 11.6 mm Al. The chamber is calibrated in air against a secondary standard using the substitution method.

CTDI Phantom measurements when converted to doses are quoted as mGy to perspex, using an f-factor of 7.8×10^{-3} to convert exposure (mR) to dose in perspex (mGy) at approximately 70 keV[22].

■ LiF Thermoluminescent Dosimeters

■ Phantom surface measurements

~55 keV Calibration

The TLD's are calibrated using a diagnostic X-ray set operating at 125 kVp, with additional filtration to give an HVL of 8 mm Al. The TLD's are calibrated on a perspex back scatter material with a chamber irradiated simultaneously. The chamber is calibrated against a secondary standard.

In order to convert to absorbed dose in water, in mGy from exposure in mR, an f-factor of 9×10^{-3} mGy/mR is used, derived from tables based on an effective energy of 55 keV[37].

Historically within ImPACT the calibration energy of 55 keV has been used for all TLD dose measurements dating from the early scanners, even though this is more representative of the effective energy for free-in-air dose measurements, whereas for phantom dose measurements the effective energy of the scanner is more closely in the region of 70-75 keV.

Since ImPACT TLD dose data is used relatively, as a comparison with previous ImPACT data, the present calibration energy for TLD measurements remains at 55 keV.

TLD sensitivities differ by about 5% for the two effective energies of 55 keV and 70 keV, and combined with the relevant f-factor there is an absolute error in calculated dose of about 9%[38]. However, this only needs to be noted if users are comparing ImPACT data with their own.

~70-75 keV Calibration

Where absolute measurements are required, and there is no need for comparison with previous data, the TLD's are calibrated using a diagnostic X-ray set operating at 125 kVp, with additional filtration to give an HVL of 11.7 mm Al. The TLD's are calibrated on a perspex back scatter material with a chamber irradiated simultaneously. The chamber is calibrated against a secondary standard.

In order to obtain absorbed dose in water an f-factor of 9.3×10^{-3} mGy/mR is used, derived from tables based on an effective energy of 70 keV[37].

Appendix 3. MDA and Evaluation Centre Information

■ Background

One of the roles of the Medical Devices Agency (MDA) is to support evaluation programmes for medical devices and equipment. The MDA Device Evaluation and Publications Business Unit is responsible for this work. The programme includes evaluation of X-ray Computed Tomography Equipment currently available on the UK market.

MDA aims to ensure that evaluation techniques keep abreast of improvements in CT imaging performance and that MDA reports present evaluation information that is timely, useful and readily understood.

■ Evaluation Facility

The evaluations are undertaken, and the reports written, by ImPACT (Imaging Performance Assessment of CT Scanners) which is MDA's CT evaluation facility. It is based at St George's Hospital, London which is part of St George's Healthcare NHS Trust.

The individual reports contain the results from a type test of the image quality and dosimetry of a CT scanner. Type testing is a procedure whereby one machine, that has been identified by the manufacturer or supplier as operating within specification, is assessed and the results taken to be representative of that model.

Clinical user evaluations are also undertaken by the ImPACT group and features such as ergonomics and ease of use are reported on in User Reports of the Scanners.

■ Manufacturer Involvement

Measurements are carried out with the active participation of the manufacturer, at a mutually acceptable site and on a current CT system running the latest release of software. This helps to ensure that the performance is within specification and is representative of the current capabilities of the model.

Acknowledgement is made to all the CT manufacturers for supporting MDA's CT evaluation programme.

Before publication, a copy of each report is sent to the manufacturer or supplier for comment. Comments received are either incorporated in a revised draft or presented in the report as appropriate.

■ Support to Purchasers and Users

■ ImPACT

The ImPACT team are available to answer any queries with regard to details within the reports:

Technical (Imaging and Dosimetry) Reports:
S. Edyvean, M.A. Lewis, J.F. Carden

User Reports:
L.A. Donnison

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■ MDA

The Medical Devices Agency can be contacted for the availability of CT reports and general information on other MDA Reports.

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■ World Wide Web

MDA: http://www.medical-devices.gov.uk
ImPACT: http://www.sghphy.demon.co.uk/impact.html

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