February 2005

Medicines and Healthcare products Regulatory Agency

EVALUATION

Report 05016

CT scanner automatic exposure control systems

Department of Health

About MHRA evaluation reports.

What you can expect.

The Device Evaluation Service (DES) aims to provide independent and objective evaluations of medical devices available on the UK market. Specialist centres, mainly in NHS Trusts, do the evaluations under long term contract to, and in accordance with protocols approved by, the Medicines and Healthcare products Regulatory Agency (MHRA). The evaluations are usually of a unit supplied by the manufacturer. We would expect this unit to be representative of the product on the market but cannot guarantee this. Prospective purchasers should satisfy themselves with respect to any modifications that might be made to the product type after MHRA's evaluation. The reports are intended to supplement, not replace, information already available to prospective purchasers.

The MHRA DES does not have access to any information held by the Agency in its capacity as the Competent Authority for the UK, apart from any information already in the public domain. The reports will contain data given by the manufacturer on the regulatory status of their devices but, apart from this, they are not an indicator of the regulatory status of a product. Occasionally, DES refers products to the regulatory arm of the MHRA for considerations of breaches of the legislation governing medical devices. DES plays no further part in any regulatory investigation that ensues and does not have advance notification of any regulatory action that may follow.

How to obtain MHRA reports.
To order evaluation reports, a copy of the publications catalogue or to sign up for our e-mail alert service contact:

MHRA
Orders Department,
Room 1207,
Hannibal House,
Elephant & Castle,
London,
SE1 6TQ.

T: 020 7972 8181.
F: 020 7972 8105.
E: des@mhra.gsi.gov.uk

Visit www.mhra.gov.uk for a comprehensive list of publications, details of forthcoming evaluations, services and contacts.

Colour reports.
Full colour versions of all reports published after 2002 are available from the NHSnet at:
www.medical-devices.gov.uk

CT scanner automatic exposure control systems

Nicholas Keat

ImPACT
Bence Jones Offices
St George's Hospital
London SW17 0QT

Tel: 020 8725 3366
Fax: 020 8725 3969
e-mail: impact@impactscan.org

For more information on ImPACT visit www.impactscan.org

The MHRA is an executive agency of the Department of Health.

ISBN 1 84182 947 1

Contents

Introduction

Automatic exposure control (AEC) systems for CT scanners are now available from all the major scanner manufacturers. These systems have different capabilities and operate in a variety of ways, but their main purpose is to adjust the x-ray tube current to compensate for different levels of attenuation of the scanner's x-ray beam. This has a number of potential advantages, including consistency of image quality, better control of patient radiation dose, avoidance of certain types of image artefact and reduced load on the x-ray tube, leading to extended scan run lengths. As well as the advantages of AEC systems, there is a responsibility to use them correctly, and this requires some education about the capabilities of particular systems, and the introduction of new concepts for controlling their operation.

This emphasis of this report is in two areas; to give the reader general information on the issues surrounding CT AEC, and to report on the results of ImPACT's testing on the manufacturers' current systems. The report discusses the CT AEC systems in current use, covering the technology and methods of operation. It lists the capabilities of present systems and includes the results of testing on current 16 slice scanners. Because of the differing approaches of each AEC system, the results section is not a direct like-with-like comparison. Instead it investigates the operation of each system, and its response to a range of variables. Information is provided on approaches for the introduction of AEC into clinical practice, as well as techniques for optimising their use.

Background

The effect of patient size and composition on image quality in x-ray images

One of the primary goals of any diagnostic x-ray department is to consistently produce high quality patient images. In order to do so, a range of exposure parameters can be adjusted, primarily tube voltage (kV), tube current (mA), and x-ray exposure time, as well as other factors. These general factors hold true for all x-ray techniques, whether analogue or digital, from standard planar x-rays and image intensifiers to mammography and CT.

The reason that a single set of exposure parameters cannot be used for all patients is that the variations in size and composition of the anatomical regions being examined affect image quality.

The penetration of x-rays through a patient depends on a range of factors, but as an approximate guide, the intensity of a CT x-ray beam will approximately halve for each 3 cm of soft tissue it passes through[1]. This distance is known as the half value thickness. For x-rays of two patients, one of whom has a cross sectional thickness 6 cm greater than the other, the intensity of the x-ray beam at a detector would differ by a factor of about four (two half value thicknesses). This would result in different levels of exposure to an x-ray film, or different levels of image noise in a digital system.

Patient composition also plays a role in image quality, as some tissues are more attenuating than others. The main examples in the body are bone and lung tissue, which are respectively more and less attenuating than soft tissue. This affects the intensity of an x-ray beam in a similar way to that of patient thickness.

Automatic exposure controls in general diagnostic radiology

AEC systems have been available in general diagnostic radiology departments for many years. Phototimers were first used in the 1940s [2] to terminate x-ray beams at a pre-determined exposure level, measured at the film. This is designed to ensure a consistent optical density of the x-ray film, avoiding over- and under-exposure that might result in non-diagnostic images and a repeat examination. When carrying out x-rays on a series of patients using an AEC, the tube voltage and current are set according to exam type, but the exposure time is automatically varied to compensate for differences in size and composition of each patient.

The effect of patient size and composition on image quality in CT

CT scanners produce digital images that are mathematically reconstructed from a series of attenuation measurements made during the course of the rotation of an x-ray tube and detectors around the patient. These images do not suffer from under- and over-exposure problems in the same way as film based images. As with all digital images, the use of window level and width controls allows viewing in a highly flexible manner.

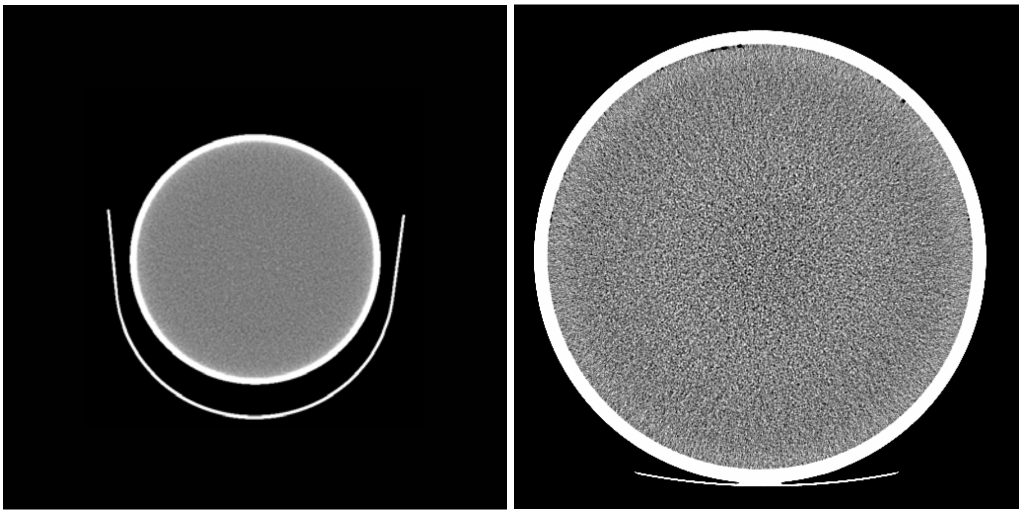
As discussed above, the effect of differences in patient size and composition is to increase or reduce the intensity of the x-ray beam at the CT detectors. Because of the statistical

---

[1] 120kV beam, 7° anode angle, 8 mm Al filtration, HVL in soft tissue = 30 mm, as calculated using [1]

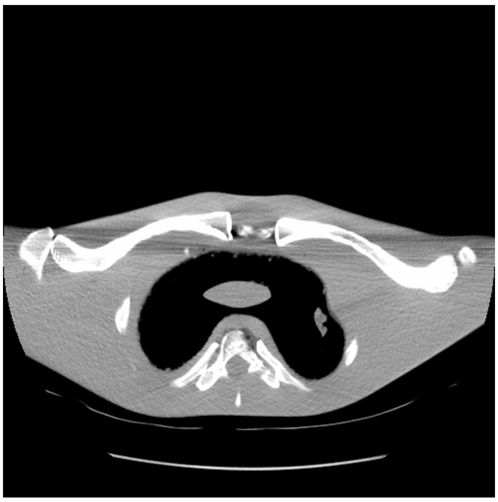
nature of x-ray transmission, the accuracy of the attenuation measurements made by the detectors depends on the beam intensity. When relatively few x-ray photons are detected, the accuracy of the attenuation measurement is poorer. Images reconstructed from this data, will have higher noise, and will appear more ‘mottled’ (Figure 1).

Figure 1. Head and body sized phantoms imaged using the same kV and mAs settings. Noise in the body sized phantom is much higher due to increased attenuation of the x-ray beam, and fewer photons at the detector



In some cases, the increase in noise can be non-uniform across the image, and ‘streaking’ artefacts can appear. This is particularly common in asymmetric regions of the body, such as the shoulders, where the anterior-posterior (AP) attenuation is much less than in the lateral scan angle, which goes through the thickest cross section of patient. In addition, the shoulders have a higher than usual amount of bony material, which increases the attenuation further. This results in high levels of uncertainty in individual measurements of attenuation through the lateral shoulder region. When images are reconstructed from this data, the errors in the lateral views dominate the noise patterns in the image, as in Figure 2 These streaks are known as ‘photon starvation’ artefacts, as they result from a lack of x-ray photons in individual attenuation measurements in the lateral direction.

Figure 2. 'Photon starvation' streaks across a CT image through the shoulders of a body phantom



Patient radiation dose from CT

With x-ray examinations, improved image quality is generally associated with an increase in the radiation dose to the patient. The process of defining the optimal examination protocol involves balancing the benefit of improved image quality with the radiation dose incurred.

CT is a major contributor to the total radiation dose from medical exposures. Reports from the UK's National Radiological Protection Board (NRPB) in 1989 [3], [4] estimated that although CT exposure represented only 2% of all radiological examinations by number, it contributed approximately 20% to the total patient radiation dose. These estimates were updated in 1999 [5] to 4% of examinations and 40% of radiation dose. The introduction of multi-slice scanners in 1999 has added capabilities to CT scanning that could be expected to increase these doses further.

Radiation dose from CT scanning has recently received considerable interest, particularly for paediatric applications. Brenner's paper [6] on estimated fatal cancers induced by CT scanning of paediatric patients generated front page headlines in at least one US national newspaper [7]. One of the key points that Brenner made was that exposure parameters are often not adjusted from those used for adults when children are scanned. When this is combined with the knowledge that younger patients receive higher effective doses for the same CT exposure parameters [8], and that the risks from radiation are thought to be higher for children than adults [9], it is clear that paediatric radiation dose from CT is of particular concern.

Existing methods for setting exposure levels in CT

The usual method for adjusting exposure levels in CT to compensate for patient size is to adjust the tube current or rotation time in order to change the mAs (tube current – time product). Although it would be possible to achieve this by altering either the x-ray tube current or the rotation time, the tube current is normally the parameter that is varied. This is mainly due to the wide range of available tube current settings, typically from around 50 to 500 mA on a modern scanner. In addition, variations in rotation time are restricted by the need to perform most scans within a limited time frame, such as a single patient breath hold.

All CT scanners have a range of pre-programmed protocols for different examination types, with set values for tube potential, current, rotation time, slice width etc. These will generally be set up for an 'average' sized patient, and the operator of the scanner can vary these parameters on a patient-by-patient basis. The degree to which the parameters are altered depends on the institution, but in many cases it is left to the judgement of the operator.

A more repeatable way of adjusting tube current for different sized patients is to relate it to some measured characteristic such as height and/or weight, body mass index or lateral width. Table 1 shows an example of this type of approach [10], using the lateral patient width at the level of the liver as the measure of patient size. Standard mA values are set in scan protocols for a patient width of 40 cm, and relative tube current settings for larger and smaller patients are shown in the table. This particular technique doubles the mA for each additional 10 cm patient width, which gives variable levels of image noise depending upon the patient size. In order to keep image noise constant, the mA would have to be doubled approximately every 3 cm.

Table 1. An example of a technique chart based on patient size, giving relative mA settings for different sizes [10]

Lateral Patient width (cm)	Relative mA
21-25	0.4
26-30	0.5
31-35	0.7
36-40	1.0
41-45	1.4
46-50	2.0

Examinations that cover more than one region of the body will often go through areas that have different attenuation characteristics. For example, a study that covers both chest and abdomen will see an increase in attenuation when moving from the mainly air filled thorax to the more dense abdominal region. Some centres use different tube currents in each of these regions to more selectively control image quality along the patient's length. A higher tube current can be used in the abdomen region to avoid the reduction in image quality that would normally occur compared to the images in the thorax region.

In addition to the mA, it is also possible to alter the x-ray tube potential in order to adjust the exposure level. Typically, CT scanners will use a tube potential of 120 kV, with other options available at one or more of 80, 100 and 140 kV. Changing the tube potential has two effects, firstly it will change the effective energy of the x-ray beam (and with it the penetration), and secondly it alters the number of x-ray photons produced.

Low kV exposures are usually used for smaller patient cross sections and in situations where the exposure level that is required is lower than can be produced using 120 kV and the minimum tube current. For equal image noise, low kV exposures result in greater patient dose than a higher kV exposure, although improved image contrast may make lower kV the better option in some cases [11].

Exposures using tube voltages above 120kV are usually limited to dense regions of the body, particularly bony regions, and thicker cross sections through large patients.

CT scanner AEC

Automatic exposure controls

The general aim of an AEC system for CT is to significantly reduce, or eliminate variations in image quality between different images. This also reduces the variation in radiation doses to different sized patient cross sections. On present systems, this is achieved through the control of the x-ray tube current to achieve the required level of image noise.

This can work at three levels:

Patient size AEC:
The AEC system adjusts the tube current based upon the overall size of the patient. The same mA is used for an entire examination or scan series. The aim is to reduce the variation in image quality from patient to patient that was discussed in the previous section.

Z-axis AEC:
The tube current is adjusted for each rotation of the x-ray tube, taking into account the variation of the attenuation along the patient's z-axis (along the scanner couch). The goal is to reduce the variation in image quality of images from the same series. This would be particularly useful for the scan series discussed in the previous chapter covering both the thorax and the abdomen. Here, the mA would be relatively low through the thorax region, which is less attenuating due to the presence of the air filled lungs, and higher through the abdomen, which is denser.

Rotational AEC:
The tube current is decreased and increased rapidly (modulated) during the course of each rotation to compensate for differences in attenuation between lateral (left-right) and A-P (anterior-posterior) projections [12], [13]. In general, lateral projections are more attenuating than A-P, particularly in asymmetric regions of the body, such as the shoulders or pelvis.

The amplitude of mA modulation during rotational AEC reflects the patient asymmetry, with less modulation occurring in regions where the patient is more circular, such as in the head. More modulation occurs in asymmetric regions. Through shoulder phantoms, the optimal modulation for the AP views has been reported to use less than 10% of the mA used through the lateral views [13].

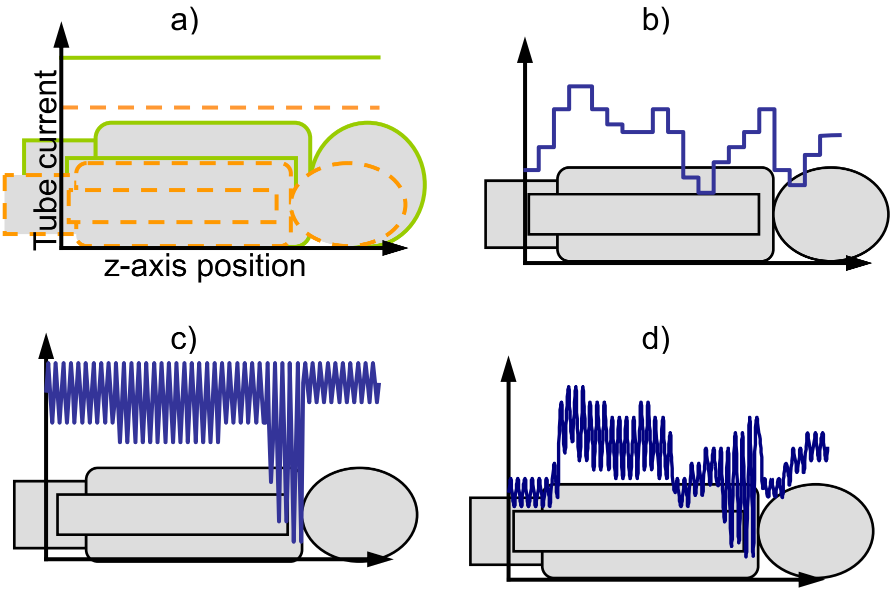
Rotational AEC requires x-ray tubes and generators that can vary the output rapidly and accurately, especially for sub-second rotation times that are now commonly used on multi-slice scanners.

Image noise is affected by rotational AEC in a different way to patient and z-axis AEC. Noise in any CT image is a function of the uncertainty of all the measurements that contribute to each pixel. The attenuation measurements with the greatest effect upon image noise are those with the greatest uncertainty, these are the ones where the fewest number of x-rays reach the detectors. Rotational AEC attempts to reduce the variation in uncertainty of attenuation measurements by increasing the tube current through the most attenuating projection angles, and reducing the mA where the attenuation is lowest. The effect on the image is to even out variations in image noise across the field of view. This also reduces the severity of photon starvation artefacts through asymmetric body regions.

Combination of AEC functions

The three levels of AEC are illustrated in Figure 3. In practice, two or three of these levels of AEC can be combined, as in Figure 3d, to allow the scanner to have the optimum exposure at each point during the scan acquisition.

Figure 3. Three levels of Automatic Exposure Control: a) patient size AEC: higher mA is used for larger patient, b) z-axis AEC: higher mA used at more attenuating z-axis positions, c) rotational AEC: the degree of modulation depends on asymmetry at each z-axis position, d) combined effects of using all three levels of AEC



Benefits of CT scanner AEC

Consistent image quality

It is important that medical imaging systems can be consistently relied upon to provide images of sufficient quality to provide accurate diagnosis. In order to achieve this in CT, scan acquisition and reconstruction variables have to be set correctly and with reference to each other. Relevant acquisition parameters include tube voltage, tube current, rotation time, collimation and helical pitch, reconstruction parameters include slice thickness, convolution kernel and reconstruction field of view. The introduction of CT AEC systems shifts the focus of controlling the scanner away from the exposure parameters and towards measures that are related directly to the output of the scanner. This has the benefit of simplifying parameter selection and improving the consistency of image quality, although it does not answer the question of what level of image quality is required for the diagnostic task.

When a CT AEC system is used, depending on its capabilities, there should be less image noise variation between patients and within a single scan series. With rotational AEC, there is also a slight reduction in the variation of noise across the field of view.

Potential for dose reduction through exposure optimisation

AEC systems by themselves will not automatically lead to a reduction in patient dose. However, when used correctly, their introduction should generally tend to result in reduced doses, for a number of reasons.

When CT protocols are defined, there can be a tendency to set tube currents higher than may be needed for average sized patients in order to avoid poor image quality for larger patients. This has the benefit of reducing the need for repeat examinations, but will lead to larger doses than may be necessary for average sized and smaller patients. Even when tube current is manually adjusted for patient size, the tendency is to reduce the mA for smaller patients less than would be required to maintain a constant level of image quality. AEC systems are generally controlled with reference to image quality, with the tube current adapted for each patient. This avoids the need to set the tube current conservatively high.

For examinations that cover more than one body region, such as chest and abdomen or head and neck, there are two possibilities with a non-AEC based CT scanner. One option is to divide the examination into two or more sections, each of which has a different mA, appropriate for that body region. Alternatively, the same mA is used for the whole examination, with the knowledge that in some body regions either image noise will be too high, or the radiation dose will be higher than necessary. Z-axis AEC overcomes this problem by setting the mA for each rotation to give a predictable level of image quality, and a more optimal radiation exposure to the patient.

Rotational AEC offers the most clear cut dose reduction of CT AEC systems. Pixel noise in CT images is related to the accuracy of the thousands of individual attenuation measurements that are used in the image formation process. The most attenuating projections have the greatest influence on image noise; as a result, it is possible to reduce the tube current for the least attenuating projections without significantly increasing the image noise.

Reduced tube loading

Multi-slice CT scanners generally have fewer problems with tube cooling than their single slice equivalents. This is due to the fact that they have high heat capacity x-ray tubes with fast cooling rates, and use wider x-ray beams that require shorter exposure times for a complete exam. Tube load considerations can still be relevant for some multi-slice examinations, but in general, if the radiation exposure to the patient is reduced by lowering tube currents, the heating of the x-ray tube is also reduced.

Extended scan runs

Some extended CT examinations can still be limited by the heat capacity of the x-ray tube. For example, in the case where the entire body and legs are imaged in an angiographic run off examination, an AEC system would reduce the tube current significantly when imaging the legs, which have a relatively low attenuation compared to the body. Using a reduced current for the extended leg region might allow the entire scan run to be imaged, whereas without AEC, a series prescribed with a single set tube current may not be able to do this due to the x-ray tube overload.

Reduction in photon starvation artefact

One benefit of rotational AEC is a potential reduction in photon starvation artefact. Streak artefacts result from inaccurate attenuation measurements at some scan angles, due to a relatively small number of x-rays being transmitted through the patient and reaching the detectors. Because the tube current is varied during the course of the rotation, it can be increased for the most attenuating scan angles (e.g. laterally through the shoulders) in order to increase the accuracy of these attenuation measurements.

Methods for AEC system control

On non-AEC based systems, the user sets the tube current for each examination, either through selection of a pre-set protocol, or by manually entering an mA or mAs value. For AEC systems to operate, a method is needed for the user to set the desired image quality level. A number of approaches are currently used, each of which has its own advantages and disadvantages.

Standard deviation based AEC control
Using this method, the user controls the AEC by specifying image quality in terms of the resultant standard deviation (SD) of pixel values. Setting a high SD value gives a noisy image; low SD settings give low noise images. The scanner aims to set the tube current that is required to achieve the requested standard deviation on an image by image basis.

One advantage of this method is that the image quality resulting from protocols from different scanners can be compared more easily, although when comparing settings from different manufacturers it is important to be aware of differences in the implementation of the systems.

Using an SD based system, the AEC is controlled by setting image quality, rather than using tube current, which is a radiation exposure related measure. Users will have to familiarise themselves with this new way of working with the scanner, as an understanding of the standard deviation of an image is not intuitive. It is important to ensure that the SD chosen is appropriate for the clinical task. It would easily be possible to enter an SD which is lower than would be needed, resulting in higher patient doses than the ones achieved without AEC. Users also need to understand that image noise is inversely proportional to the square of the tube current, so halving the SD results in an increase in the mA, and therefore the patient dose, by a factor of 4.

Reference mAs AEC control
This method of AEC control uses the familiar concept of setting an mA (or mAs) related value for a scanner protocol, in this case a 'reference' mAs is used. This is the value that would be used on an average sized patient. The AEC system assesses the size of the patient cross-section being scanned, and adjusts the tube current relative to the reference value.

The reference mAs concept permits more flexible adjustment of tube current than with standard deviation AEC controls. With SD based systems, the AEC response to different patient sizes is pre-defined, because the aim is always to keep the image noise constant. Reference mAs systems can vary their response, depending on the image quality requirements. For example, it is possible to 'under correct' for changes in patient size, so that images of small patients are less noisy than the standard patient, and those for larger patients are noisier than the standard. Smaller patients generally require better image quality than larger patients, who have a different fat distribution, making the organs easier to visualise.

It is not as straightforward to make comparisons between reference mAs protocols as it is with SD based techniques. This is because the tube current that is used for a standard patient depends upon scanner design features, such as beam filtration and scanner geometry, as well as the definition of a standard patient. However, users are generally familiar with typical mAs values for their scanners, and the move to equivalent mAs is an easier step to make than to an SD based AEC.

Reference image AEC control
The third approach that is currently used for controlling AEC systems is to use a 'reference image' that has previously been scanned and judged to be of appropriate quality for a particular clinical task. The scanner then attempts to adjust the tube current to match the noise in the reference image. The main advantage of this system is that when setting it up for use, the required image quality is expressed using an existing clinical image, rather than an abstract value of standard deviation.

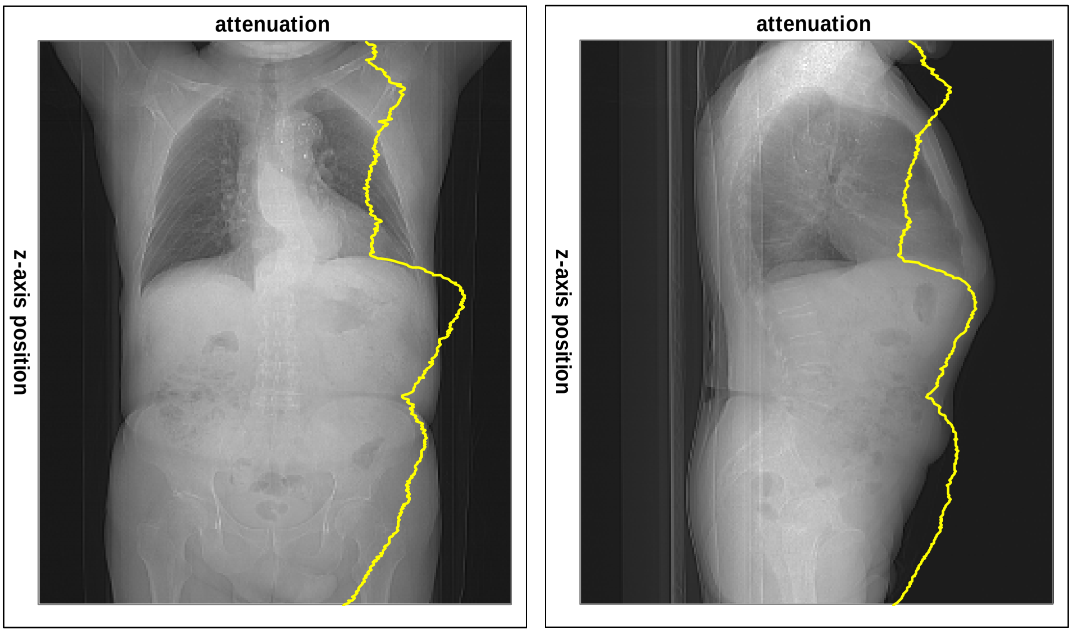
One possible drawback is that the temptation will be to pick a 'pretty' image, with low noise rather than one which has been judged to be good enough for the task. This could potentially lead to the system attempting to match higher quality images than are needed and using higher doses than are necessary. It is also difficult to compare scan protocols, even for two scanners of the same model, as there is no value associated with the image quality in the reference image.

Approaches for AEC operation

Whatever method the AEC uses to control the exposure or image quality level, there needs to be a way for the system to assess the attenuation of each patient, and calculate the tube current required.

Patient size and z-axis AEC
Scan projection radiographs (SPRs, known as scout, scanogram or topogram views) are the main way that AEC systems assess the attenuation of the patient in order to set the tube current. Figure 4 shows an AP and a lateral SPR view from the same patient, overlaid with a graph of the summed attenuation of the patient at each z-axis position. This information can be used as the basis for patient and z-axis AEC. In practice, AEC systems tend to operate using a single SPR view, to be compatible with existing clinical practice.

Figure 4. Patient attenuation measured along the z-axis from AP and lateral SPR views



Rotational AEC

In order to supply the information required for rotational AEC, data is required on how asymmetric the patient is at each position. It is possible to make estimates of this from the pattern of the attenuation profile across the patient at each z-axis position, and vary the tube current accordingly.

Another way of gaining information on the patient's rotational symmetry is to use feedback from the measurements during the course of the CT scan. Changes in the patient profile generally occur gradually along the z-axis, so the shape of the attenuation profile at each angle during each rotation can be used to control the tube current during the next rotation. In practice, the feedback can come from the scan data acquired in the previous 180° in order to reduce the lag between the change in the patient's shape, and the system's response. It is possible to use a predefined function, such as a sine wave, to vary the current between the maximum and minimum value in each rotation. Alternatively, a function that adapts more closely to the variation of attenuation with projection angle can be used [12], [13].

AEC systems on present CT scanners

GE

GE currently has two elements to their AEC system. AutomA provides the patient and z-axis AEC elements for all LightSpeed scanners. Rotational AEC is available on the LightSpeed Pro 16 only, in the form of the SmartmA function. AutomA and SmartmA can be used for both axial and helical scanning.

AutomA allows the user to set a desired image quality by entering a ‘Noise Index’ (NI). The system then aims to achieve the same level of noise in each image. The AutomA software aims to provide images with a standard deviation equal to the NI value when they are reconstructed with the ‘Standard’ convolution kernel. Images reconstructed with other kernels will result in different standard deviation values.

SmartmA varies the tube current sinusoidally during the course of each rotation, based upon calculations of asymmetry from the SPR view.

The calculations required to operate both aspects of the AEC operation are made from a single SPR view.

Philips

Philips’ AEC system is known as DoseRight, and has two elements; DoseRight ACS (Automatic Current Selector) which provides patient based AEC, and DoseRight DOM (Dose Modulation) which provides rotational AEC. DoseRight is currently available on the Mx8000 and Brilliance ranges of CT scanners. DoseRight currently has no z-axis AEC, but Philips state that DoseRight Z-DOM will fulfil this function in version 2 of their Brilliance scanner software, due for release in 2005. DoseRight can be used for axial and helical scanning.

DoseRight ACS is controlled by use of a reference image, which is assigned by the user to be used for individual scan protocols. If no reference image exists for a particular protocol, DoseRight ACS will not operate. DoseRight ACS uses an SPR view to assess the patient attenuation in order to set the tube current, which is used at all z-axis positions. The tube current is set so that 90% of images will have equal or lower noise than the reference image, with the remaining 10% of images in a series having equal or higher noise than the reference image. DoseRight DOM uses feedback from the previous rotation to assess the amplitude of mA modulation used for rotational AEC.

Siemens

Siemens’ AEC system is known as CARE Dose 4D. CARE Dose 4D is available on all Emotion and Sensation scanners with software version VB10 or higher. This is a development of the previous CARE Dose system, and adds patient and z-axis AEC to the rotational AEC that was available. The previous CARE Dose system is available on Siemens multi-slice scanners with software versions below VB10. CARE Dose is part of CARE Dose 4D and may also be selected by the user. The choice of clinical protocol determines which of the patient size and rotational aspects of CARE Dose 4D are used. Both CARE Dose and CARE Dose 4D operate in helical and axial scanning.

CARE Dose 4D is controlled using what Siemens term the ‘image quality reference mAs’, which may be adjusted by the user according to the needs of each clinical protocol. The scanner adjusts the tube current for each rotation, setting a value that is higher or lower than the reference mAs depending upon the patient attenuation at that z-axis position, relative to Siemens' reference patient size. The rotational AEC is controlled using feedback from the previous rotation to set the tube current according to the attenuation measured at each tube angle.

CARE Dose 4D is based around supplying what Siemens term 'adequate image noise', which varies depending upon the size of the patient. It uses a variable level of tube current adjustment. The degree to which the tube current is adjusted for patient size can be selected, using 'weak', 'average' or 'strong' compensation settings. These provide a relatively low, medium or high degree of mA adjustment, and are set separately for patient cross sections which are larger or smaller than the reference patient. For example, the scanner could be set up to provide 'average' reductions from the reference mAs for slim patients and 'strong' increases to the reference mAs for obese patients.

All three settings result in less tube current adjustment than would be necessary to keep image noise constant for all patient sizes. Siemens claim this approach allows the user to more closely match clinical requirements, with smaller patients requiring smoother images as it is more difficult to differentiate organs correctly when less fat is present.

Toshiba

Toshiba's AEC is known as SureExposure, and provides patient and z-axis based AEC. Rotational AEC is not available. SureExposure operates in helical scanning only.

The SureExposure AEC system is operated by selecting a target image standard deviation from a drop down list. The user can modify this list in protocol setup mode by choosing the target SD values and editing the text that describes them The AEC set-up allows tube current to be limited by maximum and minimum values.

SureExposure also has a variable length reference position selector that can be used to highlight a particular region within the scan range. The AEC uses the mean attenuation within this scan region to calculate the tube current through this region. The system uses the SPR view to obtain the attenuation information needed to set the tube current for each rotation. As with all scanners, each convolution kernel has an associated level of image noise. Because SureExposure is controlled using image SD, the mA that is used will therefore depend upon the kernel selected for primary image reconstruction.

Summary

The AEC systems from each manufacturer are shown in Table 2, which indicates the capabilities of each system to respond to changes. Note that SmartmA is currently available only on the GE LightSpeed Pro series of scanners, and that CARE Dose 4D is implemented on Siemens Emotion and Sensation scanners with software version VB10 upwards.

Table 2. Summary of AEC system capabilities

Manufacturer	Patient size AEC	Z-axis AEC	Rotational AEC
GE	AutomA	AutomA	SmartmA
Philips	DoseRight ACS	-	DoseRight DOM
Siemens	CARE Dose 4D	CARE Dose 4D	CARE Dose 4D / CARE Dose
Toshiba	SureExposure	SureExposure	-

Table 3 lists the methods that allow the operator to control the image quality when using the AEC system for each manufacturer

Table 3. Methods for setting AEC image quality level

Manufacturer	Method for setting exposure level
GE	'Noise Index' sets required image noise level for the 'Standard' kernel
Philips	System gives same level of image noise as a 'Reference Image' acquired earlier
Siemens	'Reference mAs' is set for standard sized patient
Toshiba	Target image standard deviation is set

The methods for setting AEC image quality level define to some extent the way the system adjusts when the scan protocol is changed. For example, a system based upon setting an image noise level will need to take account of more scan parameters than one that uses the reference mAs concept. Table 4 shows the AEC systems' response to the variation of relevant scan and reconstruction parameters, indicating where the tube current is automatically controlled. Where a cell is marked 'Yes', a change in that parameter will result in a change in the tube current by the AEC. None of these systems for tube current control is inherently better or worse than another, but it is particularly important for users involved in setting up scan protocols to be aware of these relationships on their scanner.

Table 4. Response of AEC systems to variation of scan and reconstruction parameters

Scan/Recon Parameter	Tube voltage	Rotation time	Helical Pitch	Slice Thickness	Convolution kernel
GE	Yes	Yes	Yes	Yes	No
Philips	Yes	Yes	Yes	Yes	No
Siemens	No	Yes	Yes	No	No
Toshiba	Yes	Yes	Yes	Yes	Yes

AEC system testing

ImPACT has tested the AEC system from each manufacturer using a custom built phantom designed to assess most aspects of AEC performance.

Phantom

The cone shaped AEC phantom is manufactured from acrylic, with an elliptical cross section that has major and minor axes in the ratio 3:2. The cross sectional area is the same as a 50 mm diameter circle at one end, and a circle of 350 mm at the other. The phantom is 300 mm long, and the increase in area along the z-axis is equivalent to a circular cone with sides angled at 45° to the scan plane. Major and minor diameters of the ellipse increase from 61.2 to 428.7 and 40.8 to 285.8 mm. The AEC phantom has a mounting lip that hooks over the edge of the wooden carrying case from a Catphan phantom (Phantom Laboratories, Salem, NY, USA), as shown in Figure 5 and Figure 6. The AEC phantom design was based on a similar phantom with a circular cross section, designed by Y Muramatsu [14].

The increase in cross sectional area along the length of the phantom is designed to test the patient size and z-axis aspects of AEC systems. The cross sectional shape is typical for an abdomen, and increases in size from smaller than a newborn child, to larger than an average adult [15], [16]. The elliptical shape of the cross section means that the attenuation through the phantom varies with projection angle, allowing rotational AEC systems to be tested.

Figure 5. The ImPACT AEC Phantom mounted on a Catphan box in a scanner gantry.

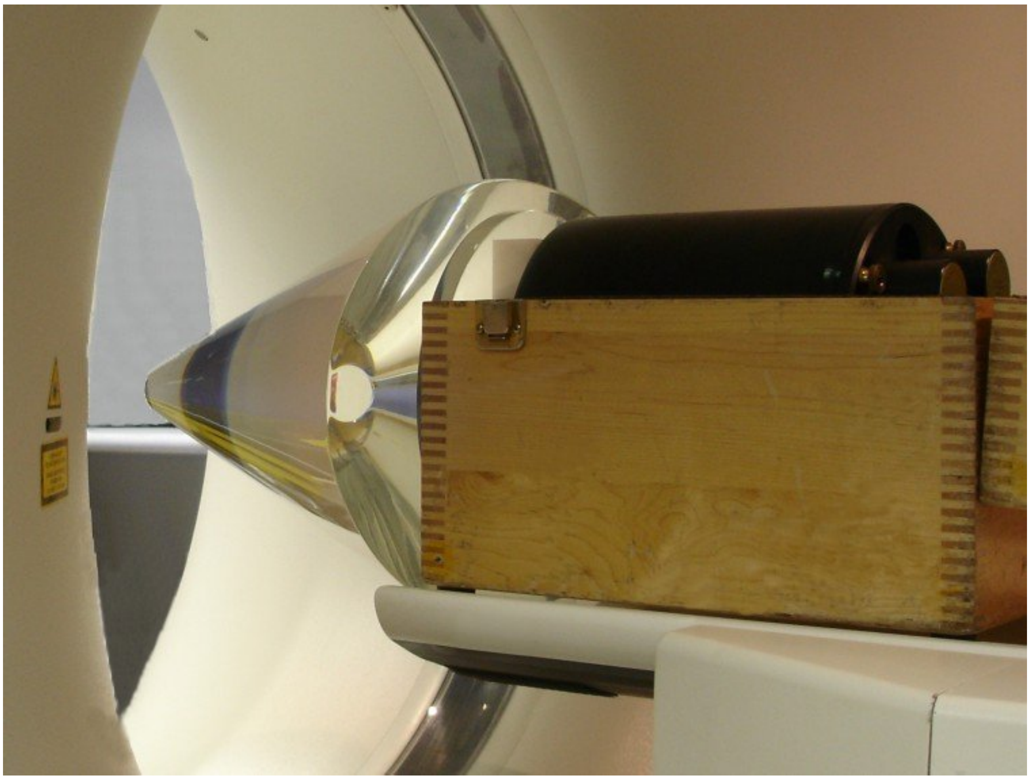
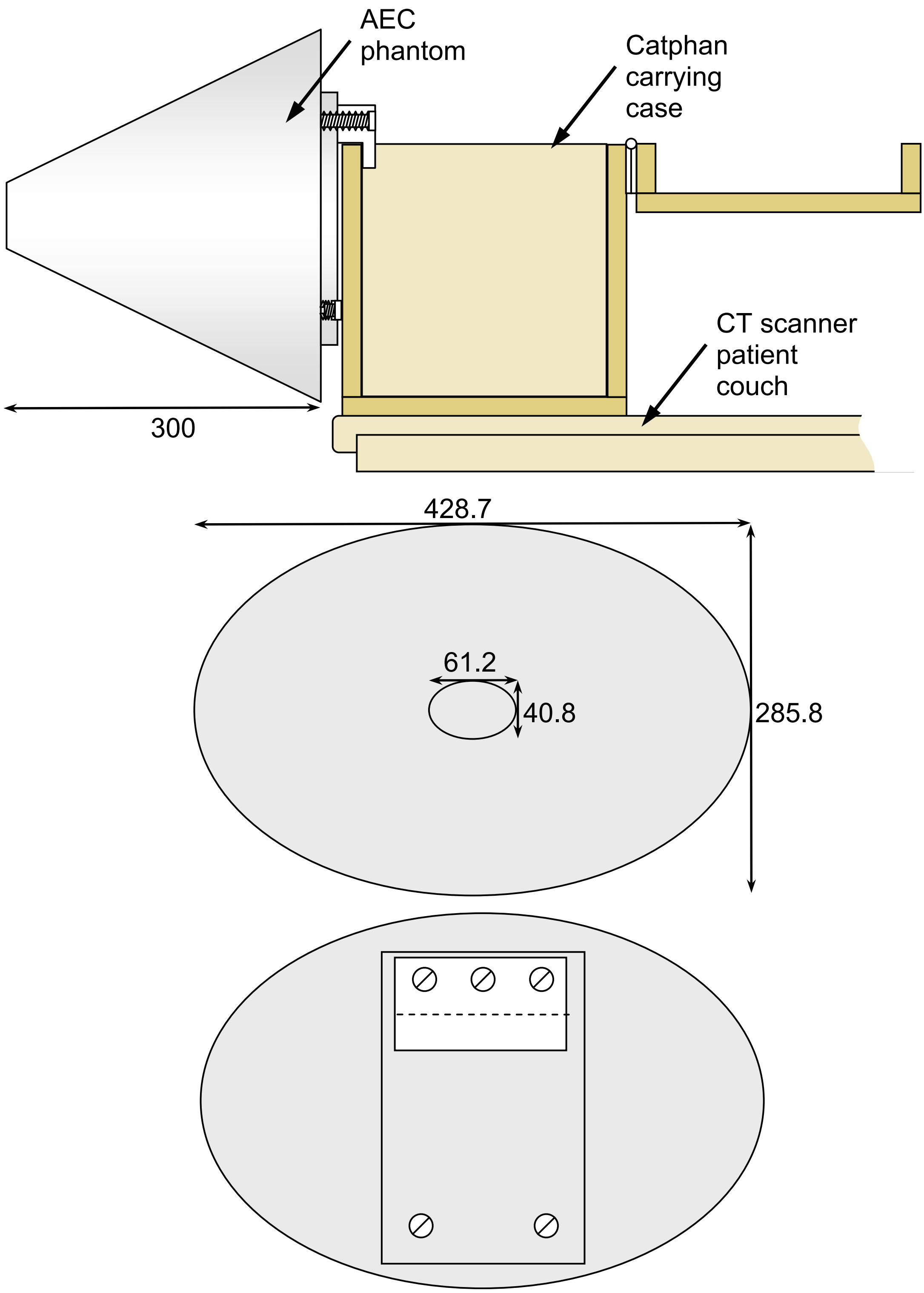


Figure 6. The ImPACT AEC phantom from the side, mounted on a Catphan box, and viewed from the 'front' and 'rear', dimensions in mm



Testing approach

The capabilities and methods for tube current control of each of the AEC systems being tested vary, so it was not practical to use a standard testing protocol. Instead, a general approach was laid out that aimed to explore the capabilities of each aspect of the systems tested.

The AEC capabilities of 16 slice scanners from each manufacturer were tested.

Standard testing conditions were 120 kV, 1 s (or less) rotation time, 200 set mA, although this depended upon clinical protocols, 5 mm slice width, 5 mm reconstruction increment, the scanner's standard reconstruction kernel, and a reconstruction field of view (RFOV) dictated by the widest part of the phantom (up to 430 mm, so 450 mm was generally used).

The phantom was carefully aligned so the cone's base was parallel to the scan plane, and centred in the field of view. Depending on the scanner, the systems were tested in axial and helical scanning modes, with the AEC switched on and off. If different levels of AEC could be selected separately, the effect of each element was individually assessed. A scan series was selected that covered the length of the phantom, although this length was restricted where necessary in order to keep tube cooling delays within reasonable limits.

For patient size AEC, a range of patient sizes were simulated by restricting SPR views to different sized portions of the phantom. For rotational modulation systems the modulation was usually greatest at the wide end of the phantom.

The effect of changing the AEC image quality level was assessed, and other scan parameters were separately varied to assess their effect upon the performance of the systems. Tube voltage, scan time, collimation, image slice width, convolution kernel, helical pitch and table travel direction were all investigated.

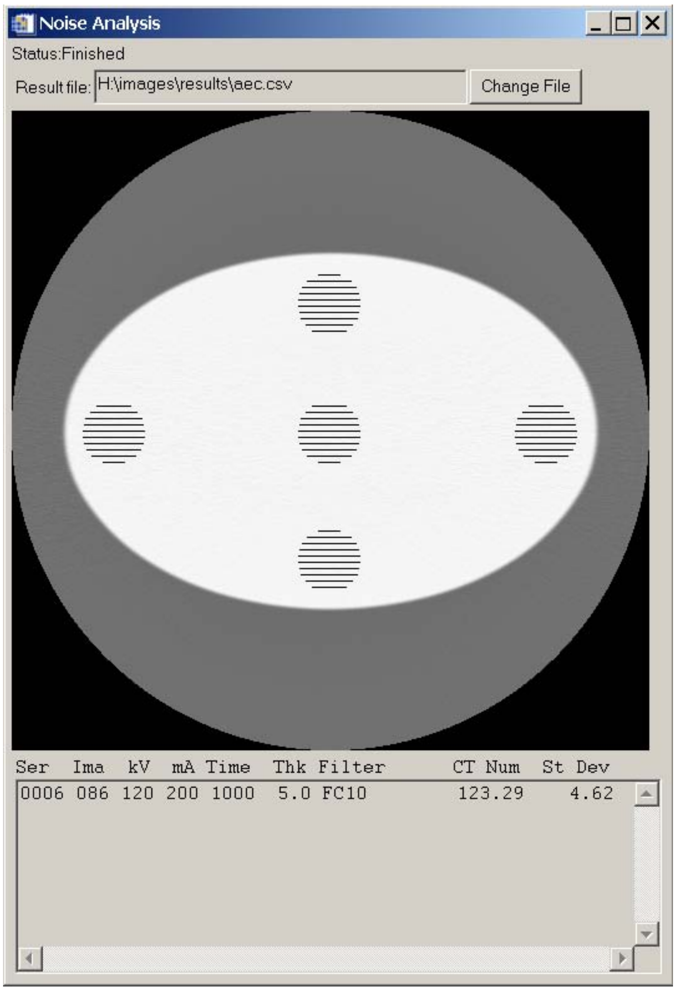
Image assessment

The images from the testing were written to CD in DICOM format, or sent to a laptop PC running the DCMTK DICOM storage SCP from the University of Oldenburg [17]. The images were analysed using ImPACT's DICOM image analysis tools, written in the IDL programming language (Research Systems, Inc, Boulder CO, USA, http://www.rsinc.com).

Each image was assessed with five 2000 mm² regions of interest (ROIs) placed automatically at the centre of the phantom, and 1 cm from the edge at each of the four cardinal points (see Figure 7). The ROIs at the periphery of the image in the upper, right, lower and left positions are referred to as the north, east, south and west ROIs in this report. 2000 mm² was chosen for the ROI size as it offered the best compromise between result repeatability and sensitivity to cross field variation of image noise. The mean CT number and image noise (standard deviation of CT number values) were calculated for each image, and the results listed in a comma separated .csv file that was read into Microsoft Excel for further analysis.

Information on radiation dose was inferred from the tube current values recorded in the DICOM image information. The dose to any one z-axis position along the phantom is directly proportional to the tube current used to acquire images at that position. Thus, if the tube current is doubled, and all other factors held constant, the dose to the phantom will also double. Note that if the same tube current is used at different z-axis positions along the phantom, the dose at each position will be different. Dose is generally defined as energy absorbed per unit mass of a body, and both these quantities will change when the phantom cross section increases or decreases.

Figure 7. Analysis of the noise in an AEC phantom image



Presentation of results

The bulk of the results in the following section are presented as graphs of the measured standard deviation in an image plotted against the AP thickness of the phantom in that image. Unless otherwise stated, the standard deviation is from the central region of interest only. The AP diameter displayed on the x-axis of the graph is the measured anterior-posterior (top to bottom of image) dimension of the phantom in each image. At any point along the phantom, the lateral diameter will be 1.5 times greater than the AP diameter, as the elliptical cross section has major and minor diameters in the ration 3:2.

With each standard deviation graph, a graph is plotted showing the tube current used for each image against the AP thickness of the phantom in that image. These graphs are shown with a logarithmic y-axis scale, due to the attenuation of the phantom increasing exponentially along its length. Systems that incorporate a z-axis AEC will result in a mA vs. AP thickness graph that appears linear on this logarithmic scale. The gradient of the line will depend on the AEC response; constant noise based systems will have a steeper gradient than systems that do not aim to keep noise constant for different patient cross sectional sizes. For the first graph for each AEC system, which shows the effect of varying the AEC control measure, the tube current against AP diameter graph is also shown on a linear scale. In all cases, images where the tube current is at the minimum or maximum value possible on that scanner are excluded from the graph and any further calculated results. In the clinical case, where the attenuation of the patient where the required mA is higher or lower than the range that the generator can provide, the image noise will be higher or lower respectively.

Standard deviation values from the peripheral regions of interest were used to examine rotational AEC systems, which have a non-uniform effect on the noise across the field of view.

Results

In order to limit the size of the results section, a selection of the results obtained from testing is presented for each scanner. The results presented have been chosen to show aspects of interest on each AEC system, and are discussed in the 'Result discussion' section. Note that the inclusion of data regarding particular aspects of an AEC's performance is not an indication of a perceived 'problem' with that AEC, and is generally included to show typical system responses to changes in scan parameters. See Table 4 on page 18 for more details of the response of each system to changes in scan and reconstruction parameters.

GE

A GE LightSpeed Pro 16 was tested. Standard axial parameters were 120 kV, 200 mA, 1 second rotation time, 10 mm beam width (16 x 0.625 mm detectors), 2 x 5 mm slice width, 'Standard' reconstruction kernel and 450 mm reconstruction field of view. For helical scanning, 20 mm beam width (16 x 1.25 mm detectors) and 5 mm slice width was used.

The GE AEC system has two elements, AutomA, which sets the tube current based upon a 'Noise Index' (NI) entered by the user, and SmartmA, which can be switched on or off and modulates the current during the course of a rotation. SmartmA is only available on the 'Pro' versions of LightSpeed scanners.

Figure 8 shows the effect of AutomA on the standard deviation measured along the length of the phantom using NI values of 5,10,15 and 20 compared to a constant 200 mA scan. The 'zig-zag' nature of the lines is due to two images being acquired per rotation, at different z-axis positions, through different thicknesses of the phantom.

Figure 8. Measured standard deviation along the AEC phantom for axial scanning on GE LightSpeed Pro 16

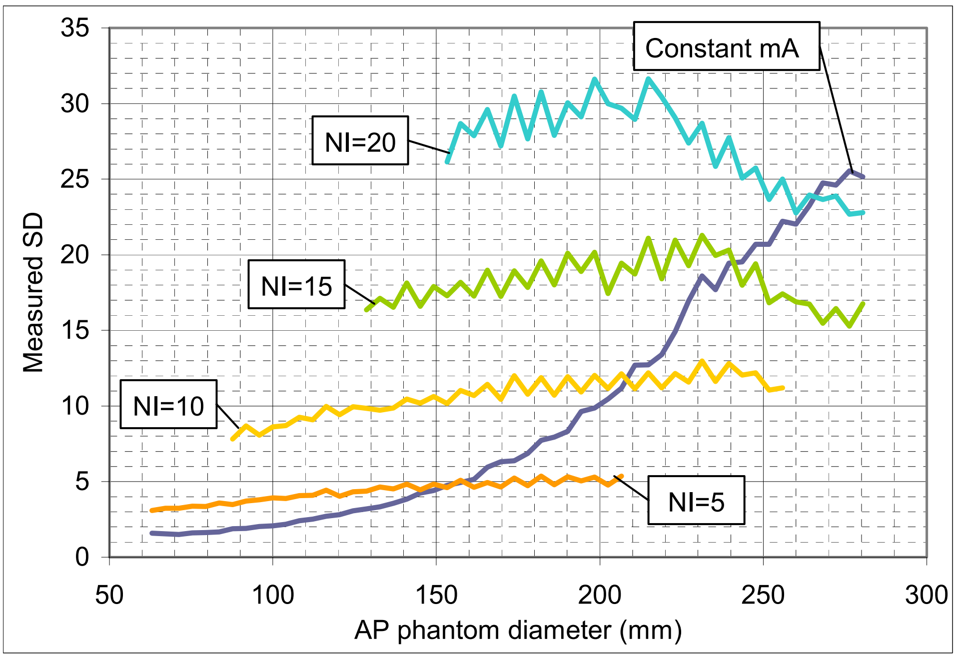


Figure 9 shows the tube current used by the scanner for each of the scan series in Figure 8.

Figure 9. Tube current along the AEC phantom for axial scanning on GE LightSpeed Pro 16 (logarithmic y axis)

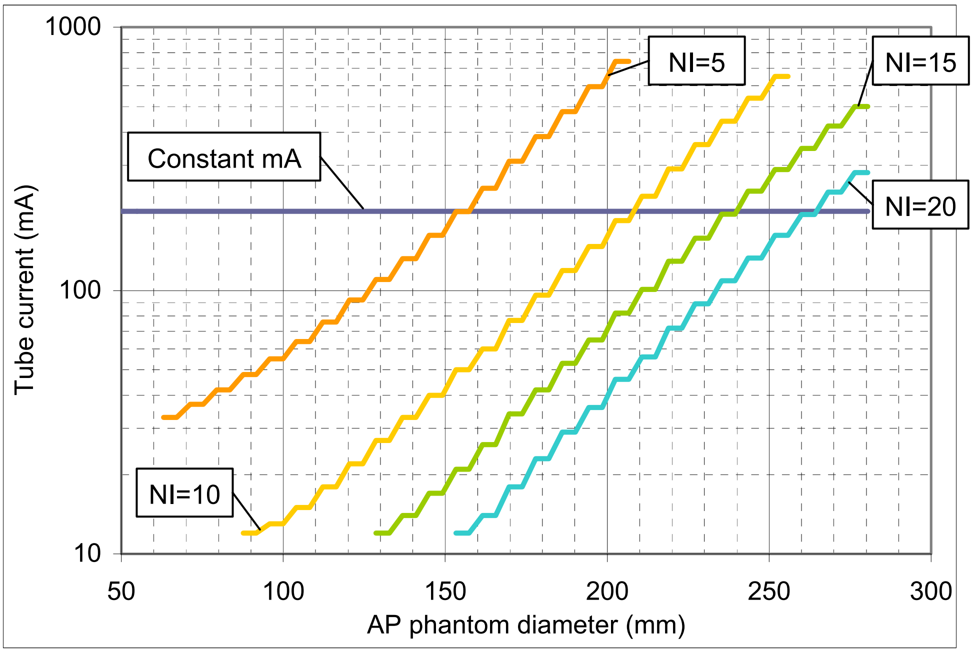
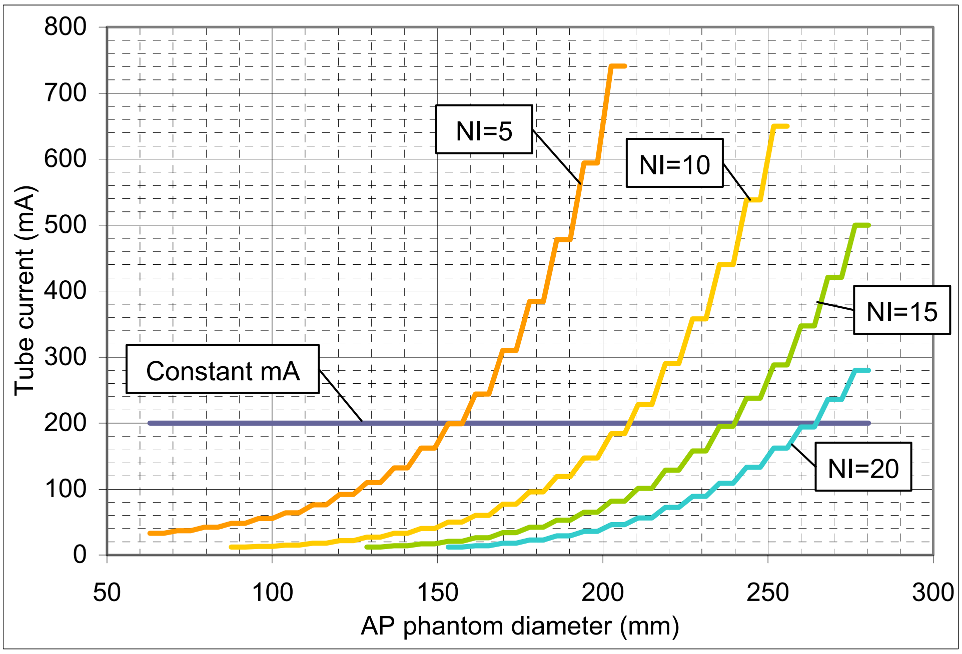


Figure 10. Tube current along the AEC phantom for axial scanning on GE LightSpeed Pro 16 (linear y axis)



The mean and range of measured standard deviation values for each axial scan series is shown in Table 5.

Table 5. Mean and range of SD for axial AEC images on GE LightSpeed Pro 16

AEC NI	Min mA	Max mA	Mean SD	SD range
OFF	200	200	9.6	1.5 - 25.5
5	10	783	4.4	3.1 - 5.4
10	10	783	10.7	7.8 - 13
15	10	500	18.2	15.3 - 21.3
20	10	280	27.4	22.7 - 31.6

Figure 11 and Figure 12 show the standard deviation and tube current for the AutomA system in helical scanning, at a range of pitches, using a NI of 10. The results presented are for table travel into the gantry (increasing phantom diameter). Similar results were obtained for the images acquired with the table moving out of the gantry (decreasing phantom diameter).

Figure 11 shows a cyclical pattern to the image noise, where the standard deviation rises and falls as the phantom diameter increases. The cycle length is the same distance as the table feed per helical rotation, due to a single tube current value being used for each rotation around the phantom.

Figure 11. Measured standard deviation along the AEC phantom for helical scanning (NI = 10) on GE LightSpeed Pro 16

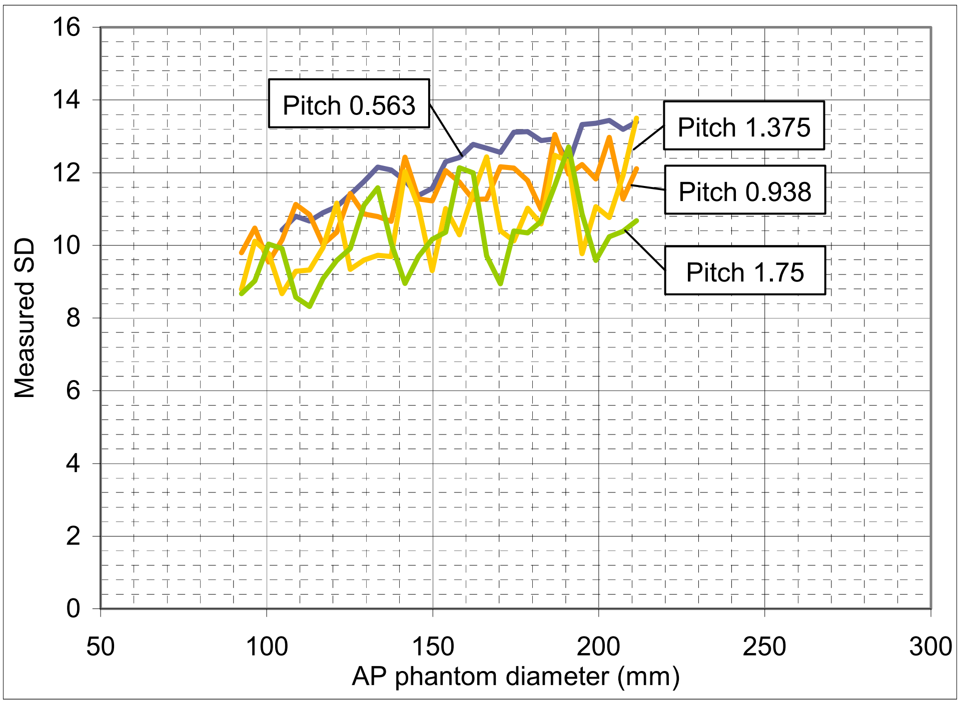


Figure 12. Tube current along the AEC phantom for helical scanning on GE LightSpeed Pro 16 (logarithmic y axis)

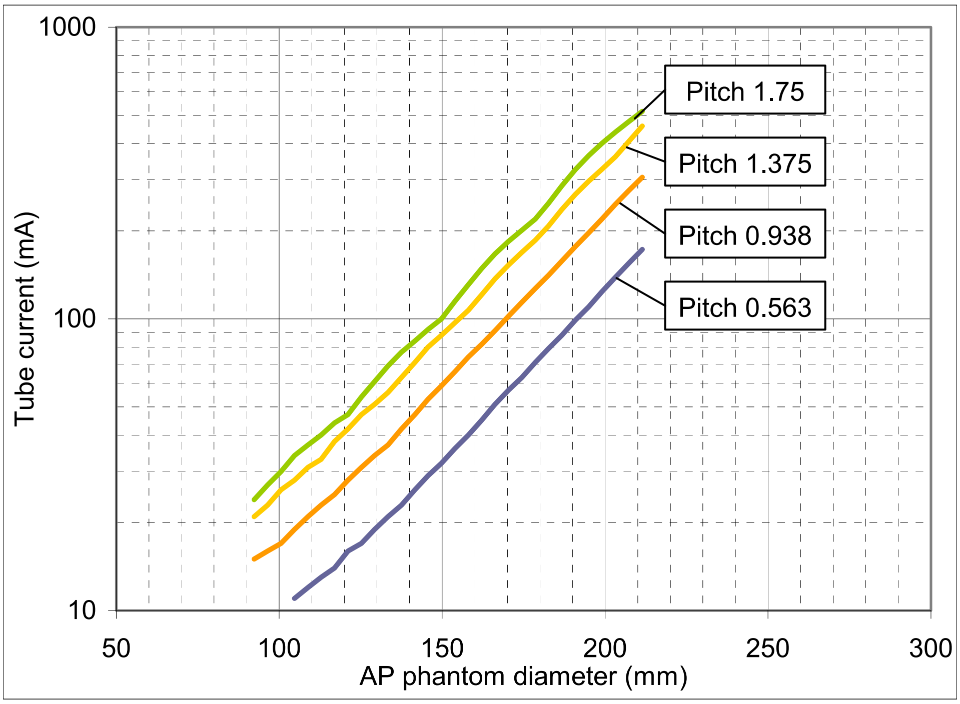


Table 6 shows the effect of SmartmA upon tube current and image noise. SmartmA was used in conjunction with AutomA, with a noise index of 10 at four locations, 50 mm apart, along the length of the phantom. For images acquired with SmartmA switched on, the table shows the tube current recorded in the image, as well as the maximum and minimum mA values indicated on screen.

The mean SD figure is an average from the five ROIs at centre and peripheral positions in each of 10 images. The 'Relative SD' column shows the noise in images with SmartmA on compared with SmartmA off, and the 'relative SD, mA adjusted' column shows the previous figure with an adjustment made for the different tube current values used to acquire each set of images:

Relative SD, mA adjusted = Relative SD x √(SmartmA on mA ÷ Smart mA off mA).

Table 6. Effect of SmartmA on tube current and image noise on GE LightSpeed Pro 16

SmartmA	AP x lateral diameters (mm)	Tube current (mA)	Mean SD	Relative SD	Relative SD, mA adjusted
Off	136 x 204	31	9.09	-	-
Off	177 x 265	87	9.28	-	-
Off	218 x 327	264	9.08	-	-
Off	255 x 382	666	8.32	-	-
On	136 x 204	29 (28-32)	9.38	1.03	1.00
On	177 x 265	82 (80-89)	9.43	1.02	0.99
On	218 x 327	252 (247-270)	9.25	1.02	1.00
On	255 x 382	646 (640-681)	8.31	1.00	0.98

Table 7 shows the effect of using AutomA on scans planned from AP and lateral scout (SPR) views. The phantom was scanned axially at the same z-axis position, using 120 kV, 1 x 5 mm slice and a NI of 6, at the point in the phantom where the AP and lateral diameters are 188 x 282 mm. This shows that the tube current, and the SD in the image can depend considerably on the SPR view angle. The lateral SPR view goes through the thickest cross section of the elliptical phantom. In order to minimise the effect of this dependency, the same SPR view angle needs to be used consistently for each exam type.

Table 7. Effect of SPR view angle on tube current and image noise

Scout view	mA	SD
AP	235	8.52
Lateral	353	6.70

Philips

Philips DoseRight system was tested on an Mx8000 IDT scanner, rather than one of Philips' more recent Brilliance systems. Standard axial parameters were 120 kV, 200 mA, 1 second rotation time, 24 mm collimation (16 x 1.5 mm detectors), 6 mm slice width, reconstruction kernel B and 400 mm field of view for axial scanning. For helical scanning, 200 mAs per slice, 5 mm slice width and pitch of 0.9 were standard.

The DoseRight AEC system comprises DoseRight ACS (Automatic Current Selector), where the patient AEC that sets the tube current for a scan based upon a previously stored reference image, and DoseRight DOM (Dose Modulation), which controls rotational tube current modulation. Philips has indicated that they will add DoseRight Z-DOM, providing z-axis AEC, to their AEC system's capabilities on version 2 of the Brilliance scanner software.

Figure 13 and Figure 14 show the standard deviation in images acquired with DoseRight ACS switched off and on respectively. Three SPR views, acquired at separate z-axis positions, were used to simulate different sized patients. A short helical scan series was acquired for each 'size' of patient with ACS off and on. The diamond in Figure 14 indicates the position and standard deviation of the image used to define the reference image. DoseRight ACS aims to set the tube current so that 90% of images in a scan series have equal or lower noise than the reference image, and the remaining 10% have equal or higher noise than the reference image. In this case, the reference image had measured SD of 9.0%, and the 90th percentile of the image noise distribution over the three series was 10.4%. Figure 15 shows the tube current used for each of the series, as well as the tube current for the reference image.

Figure 13. Measured standard deviation along the AEC phantom for three short helical series using 200 mA on Philips Mx8000 IDT (DoseRight ACS off)

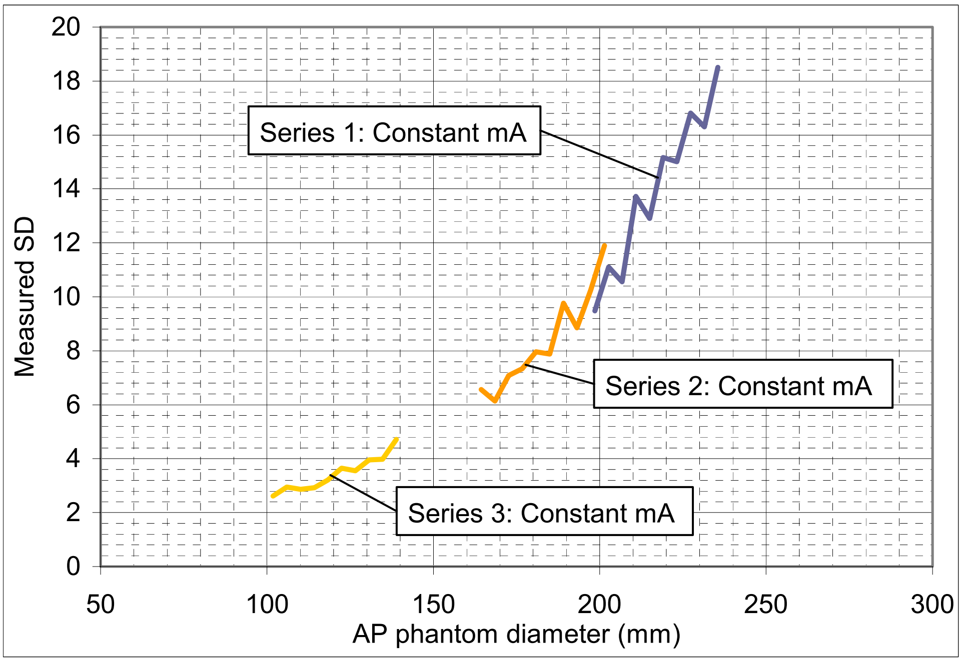


Figure 14. Measured standard deviation along the AEC phantom for three short helical series using DoseRight ACS on Philips Mx8000 IDT.

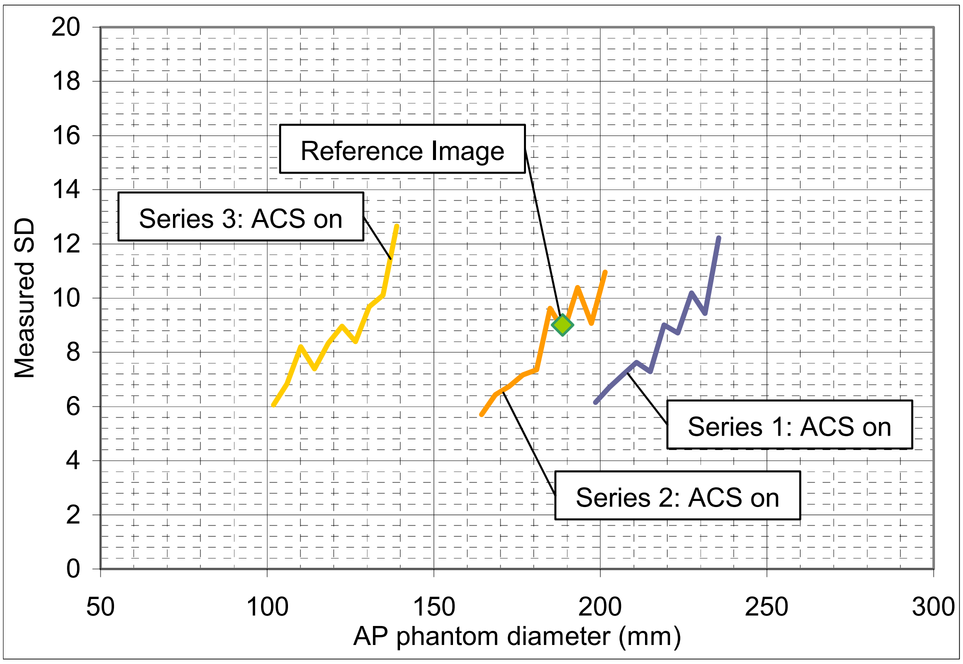


Figure 15.Tube current along the AEC phantom for three short helical series using DoseRight ACS on Philips Mx8000 IDT (logarithmic y axis)

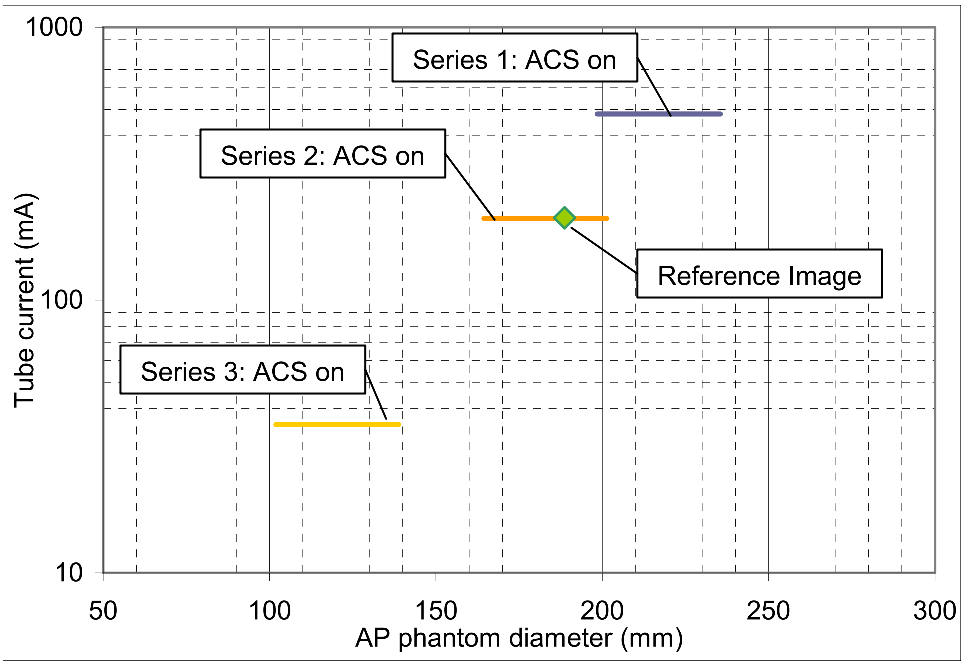


Figure 16. Tube current along the AEC phantom for three short helical series using DoseRight ACS on Philips Mx8000 IDT (linear y axis)

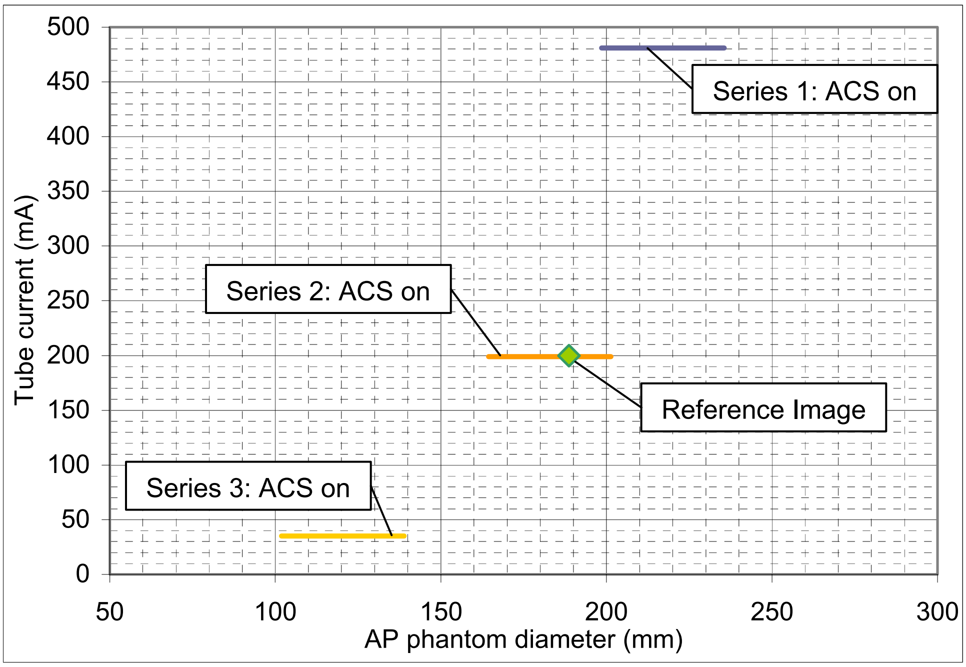


Table 8 summarises these results for the three scan series in Figure 14 and Figure 15. The '90% SD' column shows the SD in the image that represents the 90th percentile of the noise distribution. This is the image that the system aims to match to the reference image.

Table 8. Tube current and standard deviation for reference image and three short helical series using DoseRight ACS on the Philips Mx8000 IDT

Series	mA	SD range	Mean SD	90% SD
Reference	200	9.0	9.0	N/A
1	481	6.4 - 11.9	8.8	10.6
2	199	5.9 - 10.7	8.2	9.9
3	35	6.9 - 10.6	8.6	10.5

Table 9 shows the effect of DoseRight DOM upon tube current and image standard deviation for 10 axial images at each of three locations, shown in Figure 17, approximately 110 mm apart along the z-axis. The 'Mean SD' column shows the average for regions of interest at the centre and four peripheral positions. The relative SD with DOM on to DOM off is shown, as well as the relative SD with an adjustment made for the tube current used for each image: Relative SD, mA adjusted = Relative SD x √(DOM on mA ÷ DOM off mA)

Figure 17. Approximate locations of axial scans in Table 9

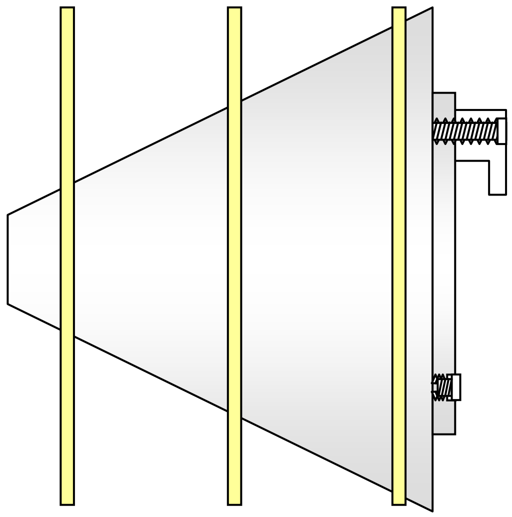


Table 9. Effect of DoseRight DOM on tube current and noise on Philips Mx8000 IDT

DoseRight DOM	AP x lateral diameters (mm)	Tube current (mA)	Mean SD	Relative SD	Relative SD, mA adjusted
Off	79 x 119	200	2.38	-	-
Off	175 x 263	200	7.04	-	-
Off	268 x 401	200	20.9	-	-
On	79 x 119	200	2.39	1.01	1.01
On	175 x 263	186	7.05	1.00	0.97
On	268 x 401	170	21.9	1.05	0.97

Table 10 shows the effect of DoseRight DOM on noise across the field of view for the images in Table 9 at the wide end of the phantom (268 x 401 mm AP diameter). The ratio of SD for DOM on to DOM off in each ROI is shown, with an adjustment made for the mean mA in each image (200 mA with DOM off, 170 mean mA with DOM on). When DOM is switched on, the noise in ROIs at the centre and the east and west of the image is reduced, whereas the noise in the north and south positions is very similar to the noise with DOM off. This shows the effect of rotational AEC techniques reducing the variation of image noise across the field of view.

Table 10. Effect of DoseRight DOM on SD in different ROIs on the Philips Mx8000 IDT

Position	SD (DOM OFF)	SD (DOM ON)	Relative SD	Relative SD, mA adjusted
Centre	42.2	42.8	1.01	0.94
North	16.5	18.0	1.09	1.01
East	22.1	22.3	1.01	0.93
South	15.7	17.3	1.11	1.02
West	22.7	22.1	0.97	0.90

Siemens

A Siemens Sensation 16, with Straton x-ray tube was tested. All of the results presented were acquired in helical mode, using a modified 'standard abdomen' protocol with settings of 120 kV, 400 effective mAs, 1 second rotation time, 24 mm beam width (16 x 1.5 mm detectors), 5 mm slice width, helical pitch 0.75 and B31 reconstruction kernel. Siemens uses the term 'effective mAs' to describe the actual scan mAs, corrected by the helical pitch of the scanner. For example, an effective mAs of 400, at a pitch of 0.75 and a scan time of 1 second uses a tube current of 300 mA (300 mA x 1 s / 0.75 = 400 mAs).

Siemens' Care DOSE 4D system is enabled by ticking the 'Care DOSE 4D' box in the scan setup page, and is controlled by entering a reference effective mAs value.

Figure 18 and Figure 19 show the image standard deviation and indicated tube current along the length of the AEC phantom, with CARE Dose 4D switched off, and on, using the 'Weak', 'Average' and 'Strong' settings that can be selected in the 'Examination Configuration' section of the scanner options. The actual mA used at each position depends upon the scanner's assessment of the attenuation relative to the standard patient. For the ImPACT AEC phantom, the mA was always lower than the reference value of 300 mA (reference effective mAs was 400, with helical pitch of 0.75).

Figure 18. Standard deviation along the AEC phantom on the Siemens Sensation 16 for helical scanning

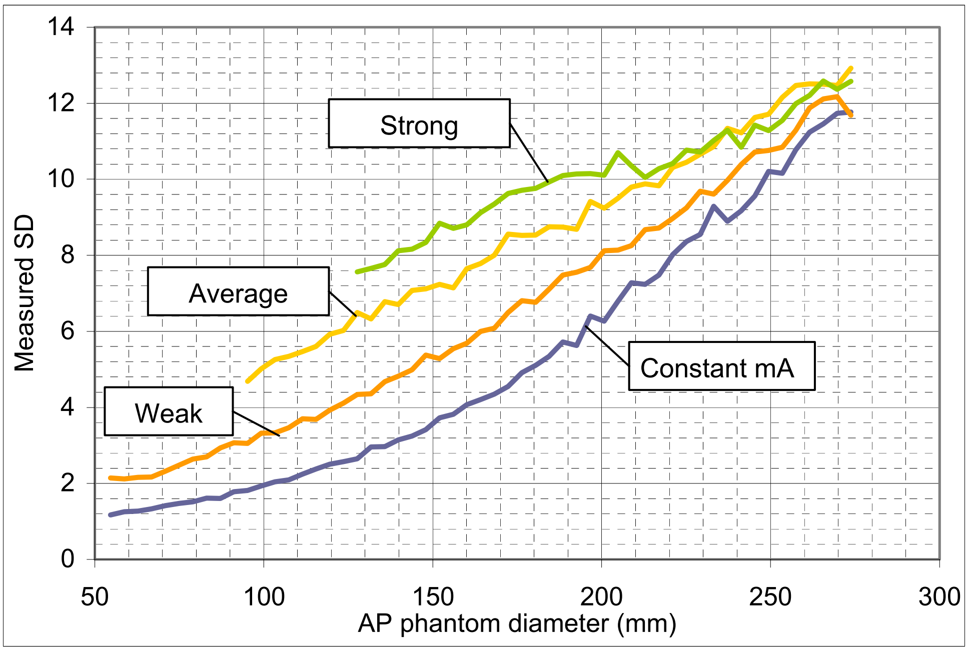


Figure 19. Tube current along the AEC phantom on the Siemens Sensation 16 for helical scanning (logarithmic y axis)

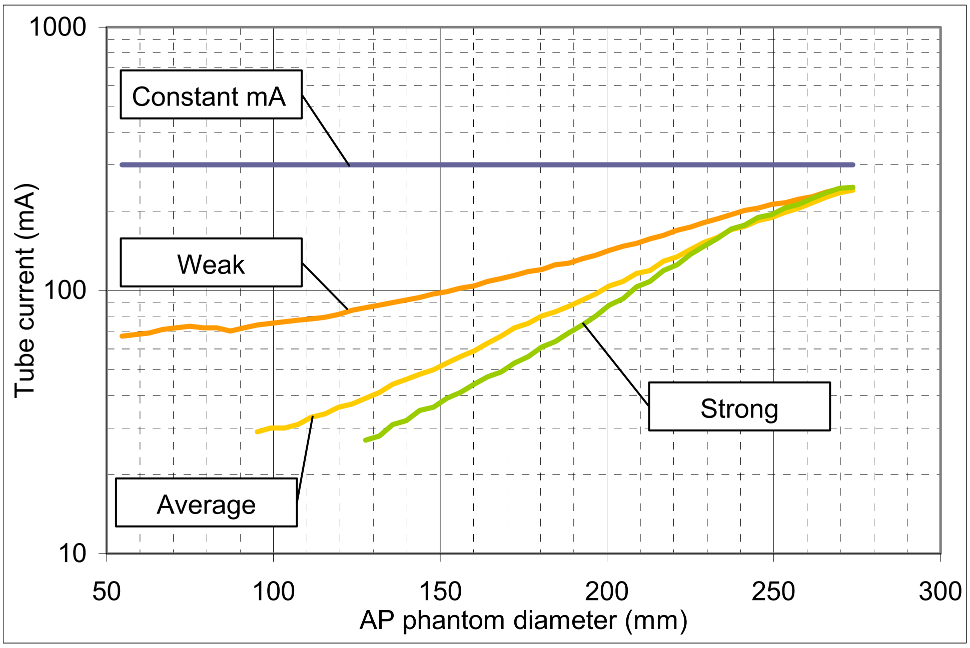


Figure 20. Tube current along the AEC phantom on the Siemens Sensation 16 for helical scanning (linear y axis)

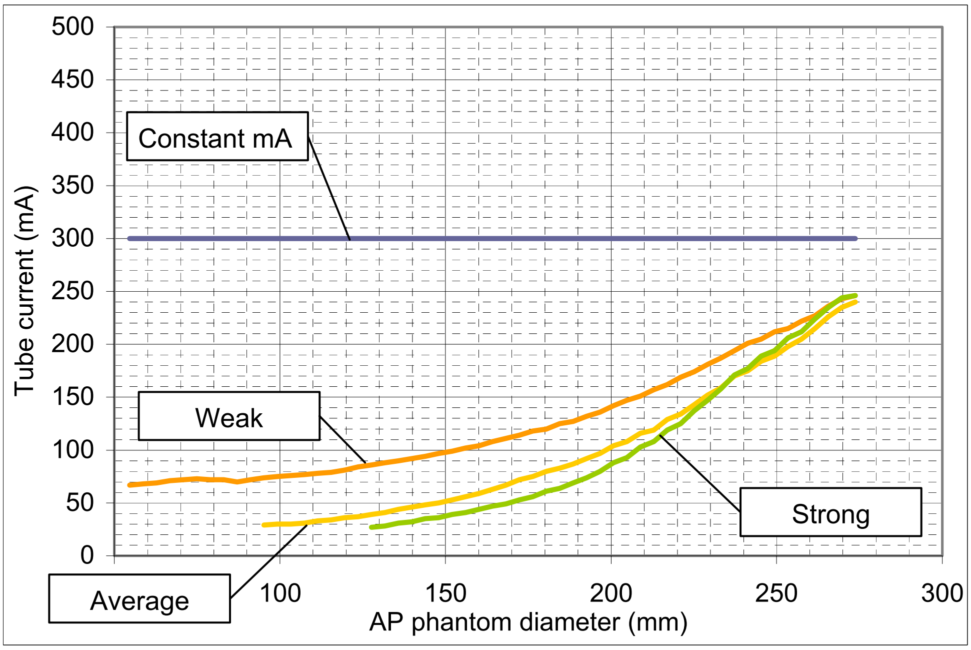


Figure 21 and Figure 22 show the standard deviation and tube current along the AEC phantom at different helical pitches. The same reference effective mAs was used at all pitches.

Figure 21. Standard deviation along the AEC phantom at different helical pitches on the Siemens Sensation 16

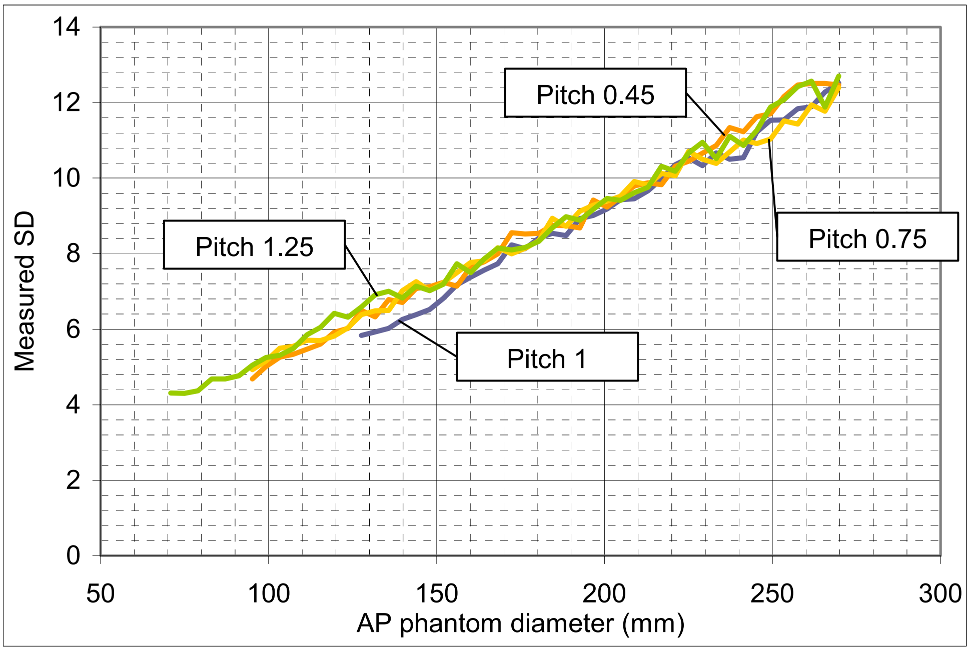


Figure 22. Tube current along the AEC phantom for different helical pitches on the Siemens Sensation 16 (logarithmic y axis)

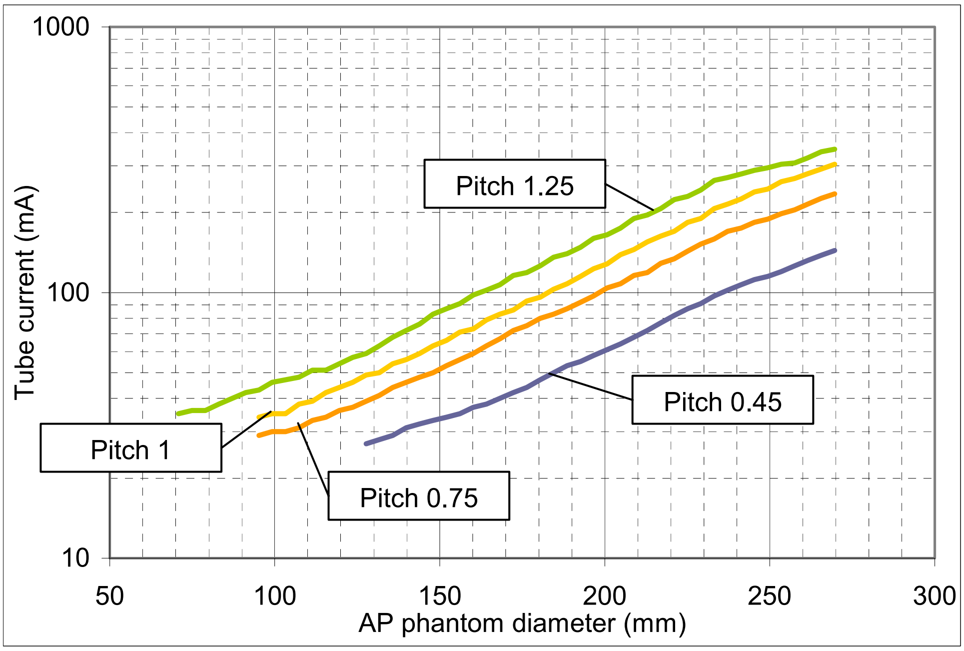


Figure 23 and Figure 24 show the measured standard deviation and tube current along the AEC phantom length at different kVs. These graphs show that the tube current adjustment does not depend upon tube voltage, except where the tube current is close to the minimum possible on the scanner.

Figure 23. Standard deviation along the AEC phantom for helical scanning at different kVs on the Siemens Sensation 16

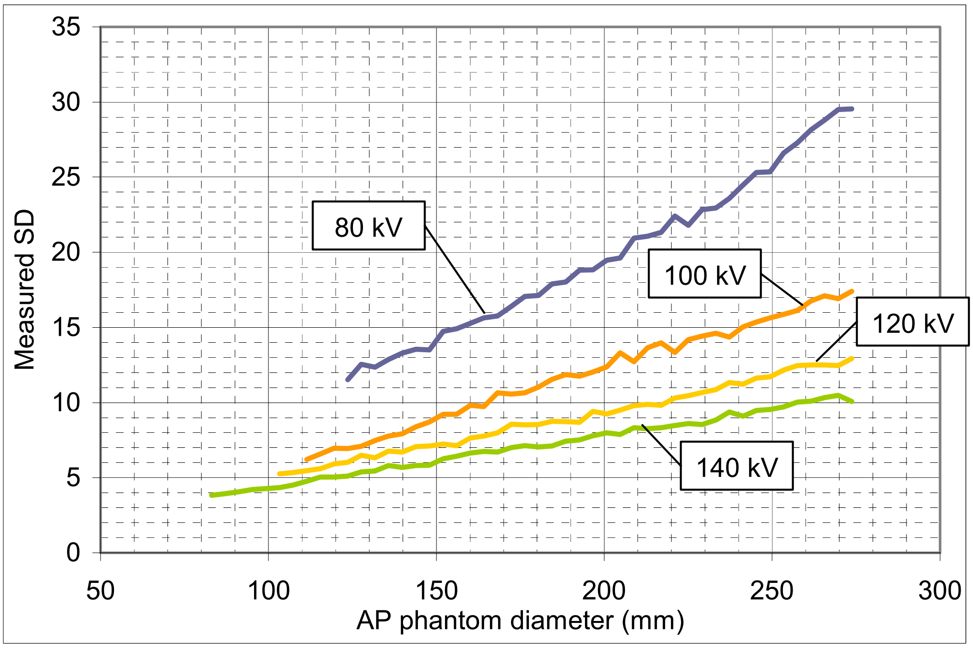


Figure 24. Tube current along the AEC phantom for helical scanning at different kVs on the Siemens Sensation 16 (logarithmic y axis)

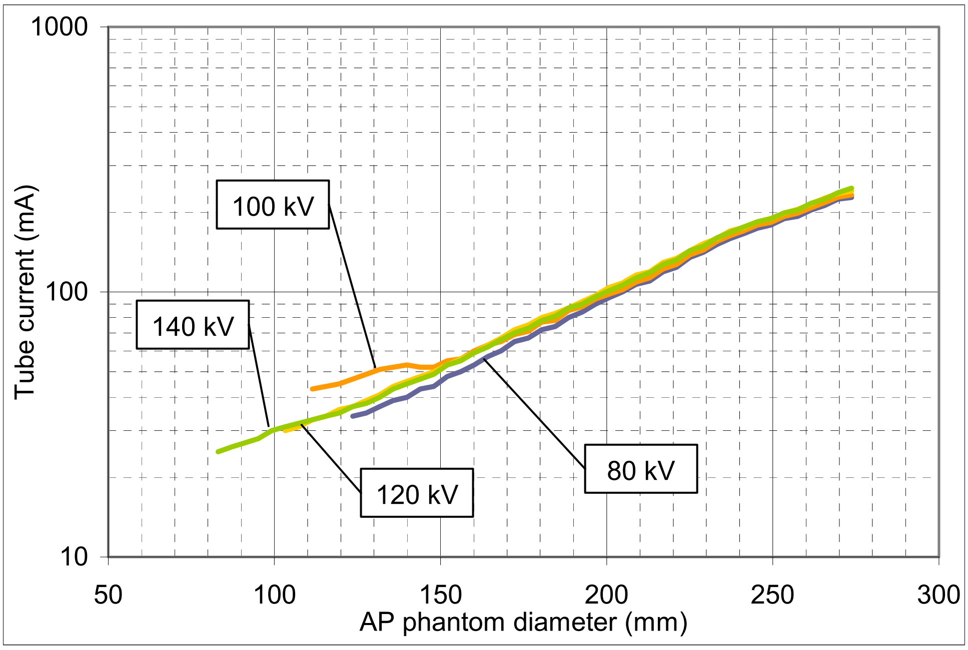


Table 11 shows the effect of planning scans using AP and lateral Topogram (SPR) views with CARE Dose 4D. The phantom was scanned at the point in the phantom where the AP and lateral diameters are 189 x 283 mm. These results show that higher tube currents are used when the lateral topogram is used. This is not unexpected, as the lateral view goes through the thickest cross section through the elliptical phantom.

Table 11. Effect of SPR view angle on tube current and image noise

Scout view	mA	SD
AP	69	10.10
Lateral	76	9.48

Toshiba

A Toshiba Aquilion 16 was tested. All images are acquired in helical mode (SureExposure does not operate in axial mode). Standard conditions were 120 kV, 200 mA, 1 second rotation time, 16 mm beam width (16 x 1 mm detectors), helical pitch of 0.9375, 5 mm slice width, FC10 reconstruction kernel and 450 mm reconstruction field of view.

The SureExposure AEC system is operated by selecting a target image standard deviation from a drop down list. The user can modify this list in protocol setup mode by choosing the target SD values and editing the text that describes them.

Figure 25 shows the effect of SureExposure upon the standard deviation in images along the AEC phantom, with constant tube current and using target standard deviation values of 5, 10, 12 and 17. Figure 26 shows the tube current used for each scan, and Table 12 summarises these results.

Figure 25. Measured standard deviation along the AEC phantom for helical scanning on Toshiba Aquilion 16

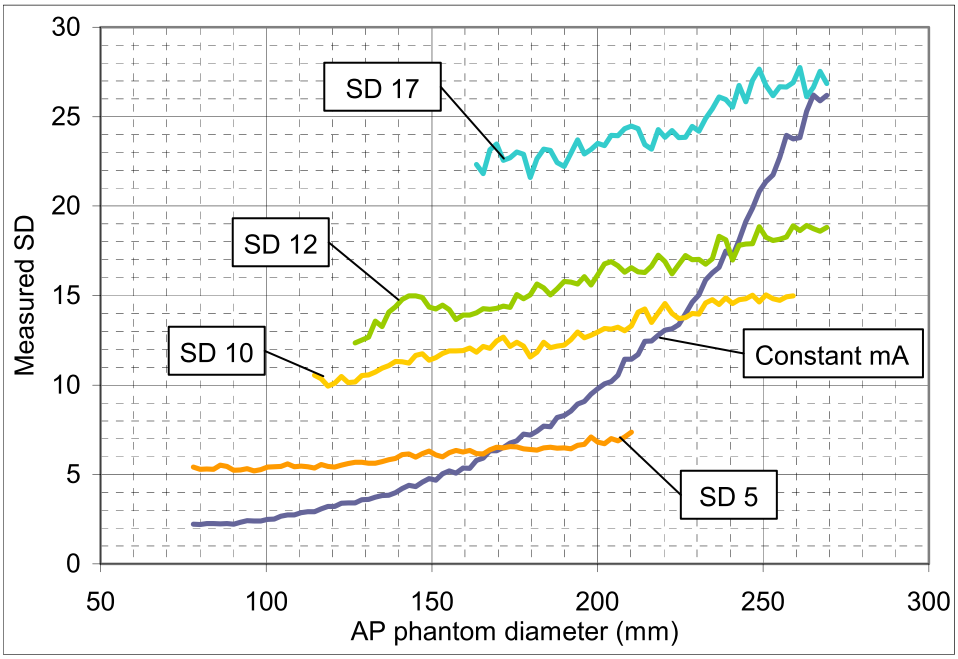


Figure 26. Tube current along the AEC phantom for helical scanning on Toshiba Aquilion 16 (logarithmic y axis)

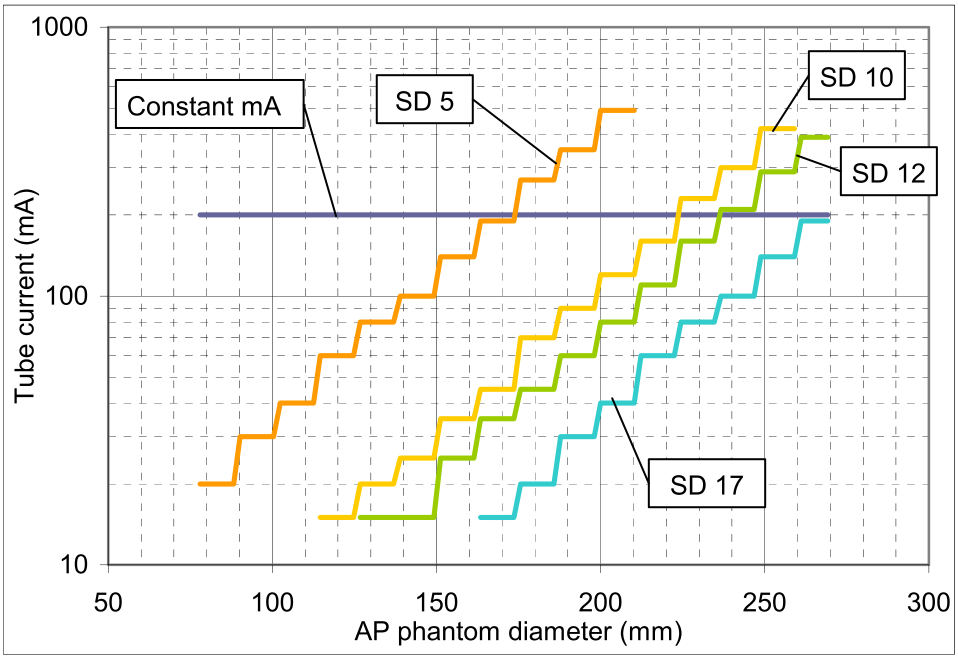


Figure 27. Tube current along the AEC phantom for helical scanning on Toshiba Aquilion 16 (linear y axis)

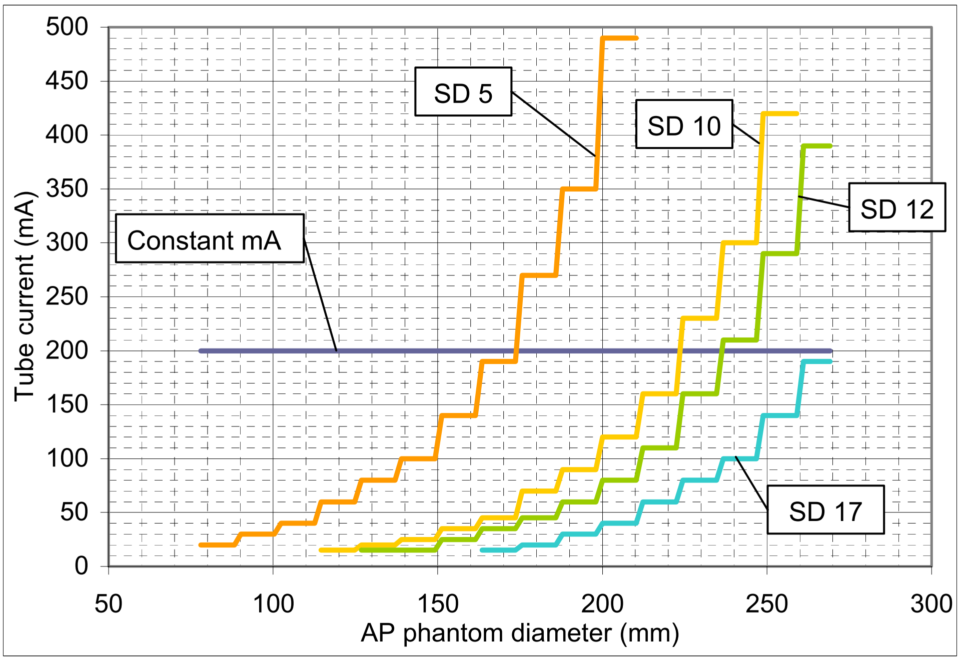


Table 12. Mean and range of SD for helical AEC images on Toshiba Aquilion 16

AEC SD	Min mA	Max mA	Mean SD	SD range
OFF	200	200	9.2	2.2 - 26.2
5	20	490	6.0	5.2 - 7.4
10	15	420	12.7	9.9 - 15
12	15	390	16.0	12.3 - 18.9
17	15	190	24.5	21.6 - 27.8

Figure 28 and Figure 29 show the effect of changing the tube voltage upon image standard deviation and tube current, for a target SD of 12. The AEC compensates for this change, as expected with an AEC controlled using image SD.

Figure 28. Measured standard deviation along the AEC phantom for helical scanning on Toshiba Aquilion 16 at 120 and 135 kV

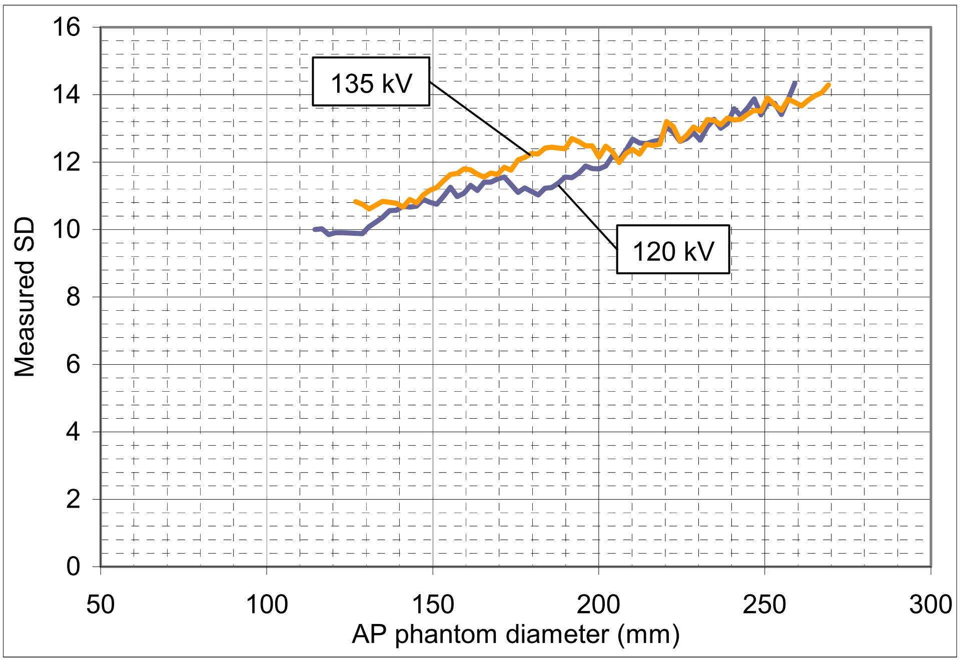


Figure 29. Tube current along the AEC phantom for helical scanning on Toshiba Aquilion 16 at 120 and 135 kV (logarithmic y axis)

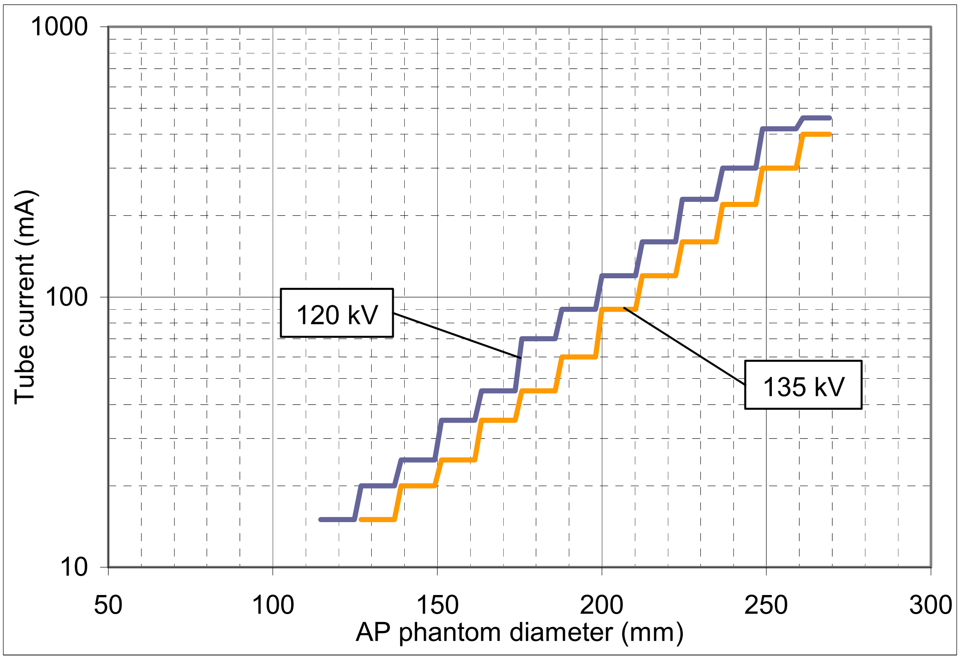


Figure 30 and Figure 31 show the effect of changing the reconstruction kernel upon image standard deviation and tube current. FC13 is the sharpest and FC10 is the smoothest of the three kernels used. As SureExposure is controlled using image SD, it adjusts the tube current for different levels of noise in each kernel.

Table 13 summarises these results, and shows the relative tube current used with each of the kernels in the 'Mean relative mA' column.

Figure 30. Measured standard deviation along the AEC phantom for helical scanning on Toshiba Aquilion 16 for different reconstruction kernels

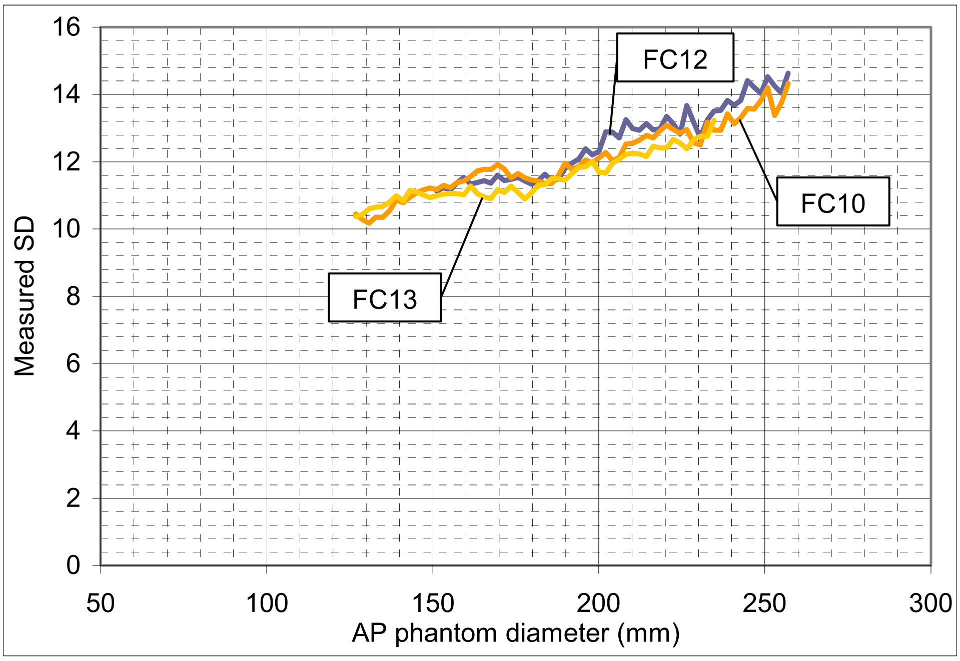


Figure 31. Tube current along the AEC phantom for helical scanning on Toshiba Aquilion 16 for different reconstruction kernels

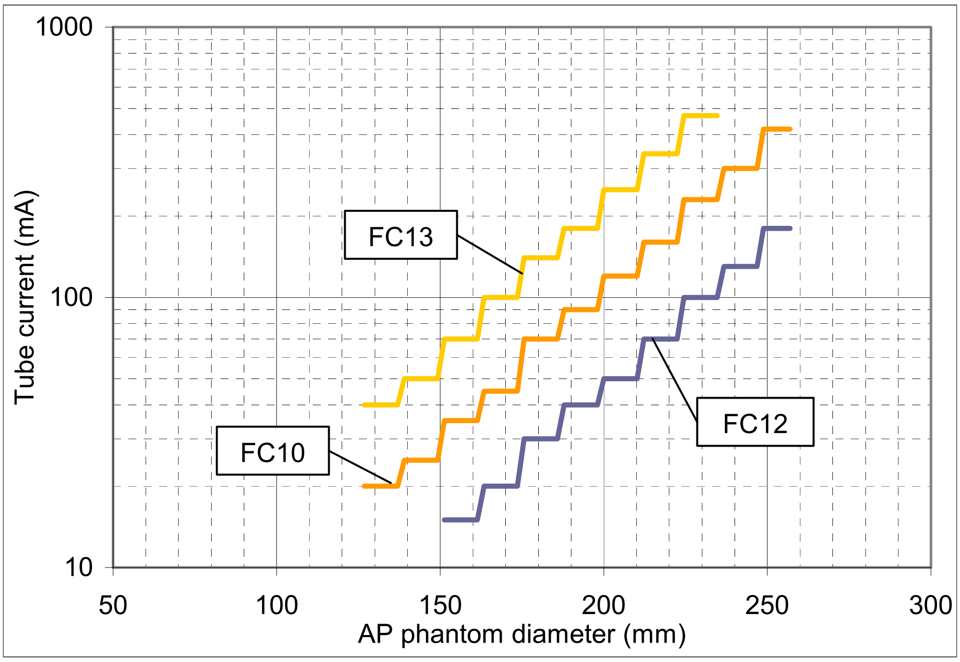


Table 13. Tube current range and relative values for helical AEC images on Toshiba Aquilion 16 for different reconstruction kernels

Kernel	Min mA	Max mA	Mean relative mA	Mean SD	SD range
FC12	15	180	0.4	12.6	11.1 - 14.6
FC10	20	420	1.0	12.1	10.2 - 14.3
FC13	40	470	2.1	11.5	10.4 - 13.2

Results discussion

The results from the current AEC systems from all manufacturers show that in broad terms they generally achieve their stated aims. The task required of the systems is inherently complex, so it is not surprising that requested image noise levels are not precisely achieved, and that image quality is not completely uniform along the scan length. All the systems assessed offer significant improvement over manually adjusted tube currents that are constant over a scan series.

Patient size and z-axis AEC

The GE and Toshiba systems were the most similar, using a standard deviation entered by the user to set the tube current for each rotation. GE's AutomA system allows the user to manually enter any Noise Index value. Toshiba's SureExposure allows selection of a target SD from a list of values with associated application descriptions in a drop down list. The list can be modified during protocol set up.

Both systems resulted in image noise that tended to be higher than the requested value; this may have been due to the fact that they were both planned from an AP SPR. An AP projection is the least attenuating angle through the phantom, so the overall size of the phantom may tend to be underestimated in this view. Table 7 shows that the SPR angle affects the results on the GE system; this might generally be expected for other AEC systems.

The main difference between the GE and Toshiba systems was the response of the AEC to changes in convolution kernel. Because the Toshiba's SureExposure is controlled by setting the required image standard deviation, the tube current values that is needed for the system to achieve this depends on the kernel. Toshiba users should be aware that changing the kernel can significantly affect the tube current, and with it the patient dose. For example, in order to achieve the SD required by the user, changing from FC10 to FC12 will approximately halve the tube current, changing to FC13 will double it.

The standard deviation of images acquired using AutomA reflect the noise index most closely when GE's 'Standard' reconstruction kernel is used. GE actually state that the noise index is an arbitrary scale, rather than being equivalent to the measured standard deviation, but there is obviously a correlation between the two. Images acquired using sharper or smoother kernels will result in images that have standard deviations which are higher or lower than the noise index. Toshiba's Real EC system aims to match the image noise to the target standard deviation value, and this is generally achieved, but as with GE's AutomA when using the Standard kernel, the images were noisier than requested for higher target SD values.

Philips' DoseRight ACS aims to set the tube current so that the level of noise within a scan series is distributed around the noise found in the reference image. This aim was achieved, although as with the Toshiba and GE AECs, the noise in the images was slightly higher than expected. Philips currently has no z-axis AEC, but this is planned for a software version 2.0 of their Brilliance CT scanners that will be available in the near future.

Siemens' CARE Dose 4D is based upon the principal that different sized patients require different levels of noise in order to obtain adequate image quality. CARE Dose 4D gives less noisy images for smaller patients or smaller patient cross sections, relative to the noise in an image of a 'reference' sized patient. It also gives noisier images for larger patients or larger cross sections within the same patient. Siemens' view that different sized patients require different levels of image quality has been repeated in clinical presentations such as [18]. CARE Dose 4D obtains variable levels of image noise along the length of a patient; Siemens claim that this reflects the clinical requirements of their customers. One effect of Siemens' approach to AEC is that it can deal with smaller or larger patients before reaching the minimum or maximum tube current that is achievable on the scanner.

The degree of mA adjustment with CARE Dose 4D can be selected by the user, as 'weak', 'average' or 'strong'. On the ImPACT AEC phantom, the tube current used was always lower than the reference value even at the widest end of the phantom, which goes up to a maximum of 429 by 286 mm. This meant that the testing was restricted to the effect of CARE Dose 4D on patients who are smaller than the reference patient size.

The results for both the GE and Siemens AEC systems showed that the mA that is used also depends on the SPR view angle, with higher tube currents being used for lateral SPRs that follow the thickest angular path through the phantom. For this reason, the SPR view angle that is used to set the image quality level for a particular examination protocol needs to be the same as the one that is used clinically.

Rotational AEC

Tube current modulation systems showed some changes to image noise across the field of view in an asymmetric phantom. The best improvements were seen through the widest sections of the phantom, as the difference between attenuation in the lateral and AP directions is greatest. This allowed a greater degree of modulation through the wider phantom cross sections, resulting in a lower mean mA than through small cross sections. As the mean mA is reduced, the image noise increases, but to a smaller degree than would be expected if the mA was constant at the lower mean value.

When rotational AEC is used without a z-axis AEC, one thing that can seem counter-intuitive is that lower mean tube currents are used when the patient is very asymmetric than when the patient is more circular in cross section. For instance, if a patient was scanned with their arms above their head out of the scan field of view, the mean mA would be higher than if they had their arms by their sides. In order to maintain image quality with the patient's arms down, an increase in the mA would be needed. With a rotational AEC system, the mean mA would be lower, as the cross section would now be more asymmetric than before. For this reason, rotational AEC systems are best used in conjunction with z-axis AEC in order to vary the maximum mA on a slice-by-slice basis. This would result in both a higher mean tube current and greater modulation of the tube current for the patient with their arms down.

The GE SmartmA system works in a different manner to the Philips and Siemens rotational AEC systems. SmartmA plans the modulation in advance from the SPR view, rather than from feedback from previous rotations. With the GE system, the degree of modulation is known in advance of the start of the scan which enables more accurate prospective display of the dose parameters of $CTDI_{vol}$ and DLP. Siemens use the SPR view to estimate the modulation in advance, but the actual modulation pattern is governed by scanner feedback.

AEC Phantom

The ImPACT AEC phantom provides a good method for assessing the capabilities of AEC systems. The variation in size along the z-axis, coupled with rotational asymmetry means it is capable of testing each aspect of AEC performance.

The performance of AEC systems may be different when they are used clinically rather than with a uniform phantom. The homogeneous nature of the ImPACT AEC phantom is different to a patient's cross section and the distribution of tissue types such as soft tissue, bone and air will affect the system's assessment of the attenuation. The phantom is manufactured from acrylic, which has a CT number of approximately 120 HU, and is denser than average patient anatomy. This may affect the performance of the AEC systems relative to a similar phantom made of a tissue equivalent material.

The AEC phantom is designed to mimic approximately the cross sectional shape of the abdomen, so rotational AEC systems only vary the tube current by a relatively small amount. A more asymmetric phantom would produce more modulation, and test the response of these systems to a greater degree. The shoulders typically have a lateral to AP dimension ratio of greater than 2 [16], compared to the ratio of 1.5 used in the AEC phantom, so a phantom constructed to simulate this region may be a better tool for rotational AEC assessment. If inserts were used to simulate the bones in the shoulders, it would more accurately reflect the real anatomy that is responsible for photon starvation artefacts.

Another challenge for the AEC systems is in coping with differences in attenuation that occur as a single step, rather than the gradual increase in diameter of the ImPACT phantom. In practice, changes in patient attenuation usually occur gradually (see Figure 4 on page 14), with the biggest step being at the level of the diaphragm or between the shoulders and the neck.

Clinical use of AEC systems

Clinical use of AEC systems requires careful consideration of the system set up in order to ensure that they are used with the best results. At a basic level, it is usually possible simply to switch the AEC on or off within any scan protocol, but it is important to consider the image quality and radiation dose implications.

Introducing AEC to clinical practice

AEC systems have been primarily developed as part of manufacturers' dose reduction or dose control programmes. Whilst the use of AEC should generally lead to reduced patient doses, it is also possible to operate an AEC system in a way that results in higher patient doses than would occur with a standard fixed mA system. AEC systems do not reduce patient dose per se, but enable scan protocols to be prescribed using measures related to image quality. If the required image quality is well specified by the user, and suited to the clinical task, then there is scope for reduction in patient dose.

Reductions in patient dose are accompanied by reductions in image quality, but this may be achieved without an impact upon the diagnostic usefulness of the examination. For example, when using z-axis AEC, a lower tube current is used for the less attenuating regions of the patient. This would mean a reduced patient dose relative to a scan made with a constant tube current. In the images from the less attenuating regions, the image quality will be lower than it would have been if a constant tube current was used, but is equivalent to the images in the rest of the scan series. For scanning a range of patient sizes, AEC systems can remove uncertainties involved in manually setting exposure levels. Because if this, there is less need to err on the side of caution by setting the tube current too high in order to avoid images that are too noisy.

The successful introduction of AEC into clinical practice requires input from both the equipment manufacturer and the user. The methods for control of AEC systems represent a change in the way that CT scanners are operated. New concepts such as noise index, reference mAs and reference images require understanding from the users in order for the AEC to be operated effectively. If the manufacturer supplies the scanner with good default scan protocols, and appropriate image quality values then the users' task is significantly easier. Staff training by applications specialists who understand the system is also vital.

Image quality and protocol optimisation

The optimisation of scan protocols involves a range of variables such as tube voltage and current, image thickness, collimation and helical pitch. In addition, selection of the appropriate image reconstruction parameters is also important.

Although setting the x-ray tube current is only part of the wider task of protocol optimisation, the use of AEC takes the guess work out of adjusting mA. If all other parameters are held constant, a reduction in image noise is accompanied by a rise in patient dose, and vice-versa. Therefore, the process of protocol optimisation is about ensuring that the image quality is always good enough for diagnosis, without creating an image that is 'too good', resulting in higher than necessary patient dose.

The key to ensuring the image quality is set correctly is to use an appropriate image noise level, reference mAs or reference image in the AEC setup. This is not a straightforward process.

One method of approaching this goal is to focus image quality assessment upon depiction of clinically relevant pathology, such as is outlined in the European Guidelines on Quality Criteria for Computed Tomography [19]. Another method that has potential promise for the future is the use of tools that allow users to simulate the effect of increasing image noise in a clinical image. This is done by taking the raw detector data, and using statistical methods to approximately reproduce the signals that would be obtained if a different tube current was used. The simulated data can then be reconstructed as normal, and the results compared. At present, end-user tools of this type are not available on 16 slice scanners, but manufacturers have indicated that they are involved in the development of this type of software. A similar facility is available on Toshiba's 32 and 64 slice Aquilion scanners.

Understanding the effect of scan protocol modification

One of the challenges facing users involved in modifying clinical scan protocols is gaining knowledge of the way that the various scan and reconstruction parameters affect image quality and patient dose. Although the use of CT AEC is generally quite straightforward, there are significant differences from one system to another. For example, changing the tube voltage will not affect the tube current on Siemens CARE Dose systems, but it will do on other AECs. Changing the reconstruction kernel will alter the tube current used by the Toshiba SureExposure system, but not by others. There is no obvious correct or incorrect way for manufacturers to deal with these situations, but it is important that users are aware of the behaviour of their system, and the effect that varying scan and reconstruction parameters has upon the AEC.

Monitoring patient dose

The best method to check the effect of the use of an AEC system upon patient doses is to monitor the scanner's own dose indicators. All modern scanners now display routinely the $CTDI_{vol}$ (Volume CT Dose Index) and DLP (Dose Length Product) for each examination. $CTDI_{vol}$ is a measure of the intensity of the local radiation exposure within the scanned volume, whereas DLP is the product of the $CTDI_{vol}$ and the scan length, which gives an approximate measure of overall radiation risk. By monitoring these parameters before and after the introduction of an AEC, its effect upon radiation dose for different exam types can be assessed. AEC exposure levels can be modified if necessary with reference to the figures, although the effect of changing any other parameters such as beam collimation or kV should also be accounted for.

Conclusions and future

Recent implementations of automatic exposure control systems for CT represent a significant advance in the operation of CT scanners, and allow for a more logical and reproducible method of scanner control. Each of the systems in this report operates in slightly different ways, with separate control methods and subtleties of operation.

The results of the testing on the ImPACT AEC phantom show that each system broadly achieves its aims, bearing in mind the difficulties of predicting image quality before that image has been acquired.

Introduction of the AEC systems into clinical practice should be approached carefully, and there is a need for education of users by the manufacturer as to how they can be best used. Users need to understand how the systems work, and have a good idea of the dependence of both image quality and radiation dose upon scan and reconstruction parameters. Scanners need to have default AEC protocols that draw a sensible compromise between the demands of image quality and radiation dose.

At the moment, CT AEC systems control tube current, but there is potential for future development to be able to control tube voltage and rotation time.

The use of CT AEC may enable better communication between imaging centres about imaging practice, particularly when scanners from more than one manufacturer are being used. In order to do this, a common method for operation of the system would be of great help. This could lead to greater protocol uniformity and consistently good quality diagnostic images acquired at optimal radiation dose.

References

1 Catalogue of Diagnostic X-Ray Spectra and other Data, IPEM Report No 78. K Cranley, BJ Gilmore, GWA Fogarty, L Desponds (Electronic Version prepared by D Sutton). pub IPEM, 1997

2 http://chronicle.uchicago.edu/970206/hodges.shtml

3 Survey of CT practice in the UK Part 1: Aspects of examination frequency and quality assurance. PC Shrimpton, D Hart, MC Hillier, BF Wall, K Faulkner. NRPB report 248, National Radiological Protection Board, Chilton, England, 1991.

4 Survey of CT practice in the UK Part 2: Dosimetric Aspects. PC Shrimpton, DG Jones, MC Hillier, BF Wall, JC Le Heron, K Faulkner. NRPB report 249, National Radiological Protection Board, Chilton, England, 1991.

5 Diagnostic medical exposures in the UK. PC Shrimpton, BF Wall, D Hart. Appl Radiat Isotopes 1999; 50:261-269

6 Estimated Risks of Radiation-Induced Fatal Cancer from Pediatric CT. DJ Brenner, CD Elliston, EJ Hall, WE Berdon. AJR 2001;176:289–296

7 CT scans in children linked to cancer. USA Today, January 22 2001

8 Influence of patient age on normalized effective doses calculated for CT examinations. A Khursheed, MC Hillier, PC Shrimpton, BF Wall. Br J Radiol 2002; 75:819-830

9 1990 recommendations of the International Commission on Radiological Protection. International Commission on Radiological Protection, Oxford, England: Pergamon, 1991. ICRP publication 60

10 Dose optimization in CT: creation, implementation and clinical acceptance of size-based technique charts. CH McCollough, FE Zink, JM Kofler, JS Matsumoto, KB Thomas. RSNA 2002 Scientific Program, Supplement to Radiology, November 2002, 225(p):591

11 Dose optimization in pediatric CT: current technology and future innovations. C Suess, X Chen. Pediatr Radiol 2002; 32: 729–734.

12 Dose reduction in CT by anatomically adapted tube current modulation. I. Simulation studies. M Gies, WA Kalender, H Wolf, C Suess. Medical Physics 1999; 26 (11): 2235-47

13 Dose reduction in CT by anatomically adapted tube current modulation. II. Phantom measurements. WA Kalender, H Wolf, C Suess. Medical Physics 1999; 26 (11): 2248-53

14 'Apollo' AEC phantom, Y Muramatsu, National Cancer Centre, Tokyo. Personal communication, November 2002.

15 CHILDATA, The Handbook of Child Measurements and Capabilities – Data for Design Safety. B Norris, JR Wilson. Department of Trade and Industry 1995

16 ADULTDATA, The Handbook of Adult Anthropometric and Strength Measurements – Data for Design Safety. L Peebles, B Norris. Department of Trade and Industry 1998

17 DCMTK from the OFFIS group at the University of Oldenburg. Available from http://dicom.offis.de/dcmtk.php.en

18 Pediatric CT: Radiation Dosimetry, Biological Risks and Procedure Optimization. DP Frush. RSNA 2002 Scientific Program, Supplement to Radiology, November 2002, 225(p):91

19 EUR 16262 - European guidelines on quality criteria for computed tomography. K Jessen, W Panzer, P Shrimpton, G Bongartzm J Geleijns, SJ Golding, AG Jurik, M Leonardi, G Tosi. Luxembourg: Office for Official Publications of the European Communities, 2000. ISBN 92-828-7478-8. Also available from http://www.drs.dk/guidelines/ct/quality/

Appendix 1: Manufacturers' comments

A preliminary draft version of this report was submitted to the equipment manufacturers to enable them to comment on the contents. These responses were considered when producing the final draft version of this report. Further formal replies were received from Siemens and Toshiba on receiving the final draft; these are included over the following pages.

SIEMENS

Medical Solutions

Siemens Medical Solutions
MED CTE PA
P.O. box 12 66
D-91301 Forchheim
Germany

Mr. Nicholas Keat
ImPACT Group,
St George's Hospital,
London SW17 0QT

11 February 2005

Dear Sir,

re: **ImPACT evaluation report - CT Scanner automatic exposure control systems**

On behalf of Siemens Medical Solutions, I wish to thank you for the opportunity to review this ImPACT evaluation report. We are of the opinion that any dose modulation system designed for daily clinical use should not only reduce the dose relative to the patient anatomy in any projection plane but also improve image quality with the minimum requirement of user intervention. We appreciate that the graphs of mA values are displayed in a linear scale which we believe is an accurate method to reliably determine the mAs and dose data.

In response to the evaluation of the CARE Dose 4D and CARE Dose Automatic Exposure Controls, Siemens wish to present the following additional points for consideration:

- For the dimensions of the test phantom utilised in this study, Siemens would have advised a paediatric clinical protocol would have been a more suitable selection to optimise the results of CARE Dose 4D.
  - The actual scan parameters selected (kV, mAs, rotation time) were more typical of an obese adult abdomen protocol.
  - Although the phantom size is indicative of a paediatric patient, as can be seen from the graphs, the selected protocol proved to achieve for most parts of the phantom the desired dose reduction and acceptable noise levels.
  - The alternative paediatric protocol with 0.5 s rotation time would have delivered even lower mAs for the slim section of the phantom.
- The mA curves presented in the report do not reflect the actual mA of the tube used with Siemens CARE Dose 4D. i.e. in angular projections of high attenuation, (typical lateral) , the mA will exceed the presented average tube current up to a factor of 2. Likewise the mA will be reduced to 10% of the lateral mA for the low attenuation projections (a.p. views) as can be seen in cited literature [References 11,12,13].
- The Siemens approach, CARE Dose 4D, **always utilises** a single SPR view for the purpose of estimating the required mA for all z and angular views.
- It should be noted that in our standard protocols, CARE Dose 4D uses real-time z and angular mA modulation.

Division
Computed Tomography

Head:
Dr. Bernd Montag

Medical Solutions
Group Executive Management:
Prof. Dr. E. R. Reinhardt, Group President
Dr. Hermann Requardt
Götz Steinhardt

Postal address:
Siemens AG

P.O. box 12 66

D-91301 Forchheim Erlangen

Office address:
Henkestr. 127
D-91052 Erlangen

Telephone (09131) 84-0

Page 1 of 1

SIEMENS

- Our approach of adequate image noise was selected from the basis of extensive clinical trials with several clinical sites.
- The required diagnostic image quality should be achieved at low or at least, reasonable dose levels, even for obese patients.
- It is our opinion that a constant noise approach may deliver both undesirable mAs and dose levels.

We agree with your comment that the use of dedicated clinical protocols is of utmost importance in the day to day use. In fact , with CARE Dose we believe that we have introduced not only a safe and effective dose modulation method that improves image quality, whilst using dedicated protocols (even with inexperienced users), but that it also covers a wide range of patients with a minimum of alteration of existing scan parameters,

We thank you for the opportunity to respond to this report, but should you have any questions, please do not hesitate to contact us accordingly.

Yours sincerely,

Dr Christoph Suess Ph.D
CT Research & Development
Siemens Medical Solutions
Forchheim, Germany

Page 2 of 2

TOSHIBA

TOSHIBA MEDICAL SYSTEMS EUROPE BV
HEAD OFFICE

Zilverstraat 1, 2718 RP ZOETERMEER, THE NETHERLANDS
Phone: +31 (0)79 368 92 22, Fax: +31 (0)79 368 94 44
E-mail: info@tmse.nl, www.toshiba-europe.com/medical

Nicholas Keat
ImPACT
Medical Physics Department
Knightsbridge Wing
St. George's Hospital
London SW17 OQT
United Kingdom

Subject: Evaluation CT scanner automatic exposure control systems

Dear Mr. Nicholas Keat

Thank you for giving us the opportunity to react on the ImPACT Evaluation of "CT scanner automatic exposure control systems".

We are very happy with the outcome of this evaluation however we feel the need to react on following statement made at page 44:
"Toshiba users should be aware that changing the kernel can significantly affect tube current, and with it the patient dose. For example, in order to achieve the SD required by the user, changing form FC10 to FC12 will approximately halve the tube current, changing to FC13 will double it."

First of all we appreciate that ImPACT makes users aware of the fact that using an AEC program a change in scan and/or reconstruction parameters directly influences the tube current and as such the patient dose. It is correct that for example changing the reconstruction kernel from an "average image (FC10)" to a "smoother image (FC12)" will reduce the dose and that changing from an "average image (FC10)" to a "sharper image (FC13)" will increase the dose. This dose increase/decrease however will only take place when the user keeps the requested SD (image noise) at the same level for all examinations.

Although we know it is not ImPACT's intention the statement at page 44 might give the impression that our CT scanner shows unexpected demeanour while in fact it is the only scanner correctly taking all reconstruction effects into account. A correctness confirmed by following statements in your report:

- *It is important that medical imaging systems can be consistently relied upon to provide images of sufficient quality to provide accurate diagnosis. In order to achieve this in CT, scan acquisition and reconstruction variables have to be set correctly and with reference to each other.*
  (see "Benefits of CT scanner AEC" page 11)
  As stated by ImPACT, both scan and reconstruction variables have effect on the image quality; as such reconstruction variables, including the kernel, should be implemented in an AEC.

TOSHIBA

- *The scanner aims to set the tube current that is required to achieve the requested standard deviation on an image by image basis. (see "Standard Deviation based AEC control" at page 13)* As the user is the only one who knows which image quality is required he is responsible for selecting the correct combination of reconstruction type and SD. In case the user request is wrong the system output (image quality and/or patient dose) will be "incorrect", this not because of wrong system behaviour but because of wrong user request.
- *Table 4 shows the AEC systems' response to the variation of relevant scan and reconstruction parameters, indicating where the tube current is automatically controlled. (see page 18)* Changing one of the listed parameters directly effects the image quality, therefore each change in one of these parameters should automatically lead to a change in tube current maintaining the requested SD. This table shows that the Toshiba AEC, called ^{SURE}Exposure, as only AEC program takes into account all image quality relevant parameters.

From all above, confirming that our ^{SURE}Exposure is the only AEC program taking all parameters into account, we conclude it to be incorrect that specifically Toshiba users are warned that they "should be aware that changing the kernel can significantly affect tube current". To our opinion it would be more correct to state there were needed that some of the used AEC programs do not take into account some of the image quality influencing parameters.

Your sincerely

Henk de Vries
Product Manager
Computed Tomography / Nuclear Medicine Business Group
Toshiba Medical Systems Europe BV
Head Office

Appendix 2: ImPACT and the MHRA

Background

One of the roles of the Medicines and Healthcare products Regulatory Agency (MHRA) is to fund evaluation programmes for medical devices and equipment. The programme includes evaluation of x-ray Computed Tomography Equipment currently available on the UK market.

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

ImPACT

ImPACT (Imaging Performance Assessment of Computed Tomography) is the MHRA's CT evaluation facility. It is based at St George's Hospital, London, part of St George's Healthcare NHS Trust.

ImPACT have developed test objects and measurement procedures suitable for inter-comparing CT scanner performance. For each CT evaluation hundreds of images are obtained from the system under test and subsequently analysed using custom written software. Dose measurements are made using ion chambers, and x-ray film is used to obtain additional x-ray dose information.

MHRA support to purchasers and users

The ImPACT team is available to answer any queries with regard to the details of this report, and also to offer general technical and user advice on CT purchasing, acceptance testing and quality assurance.

ImPACT
Bence Jones Offices
St George’s Hospital
London SW17 0QT

T: +44 (0) 20 8725 3366
F: +44 (0) 20 8725 3969
E: impact@impactscan.org
W: www.impactscan.org

MHRA contact point for general information on the CT evaluation programme:

Device Evaluation Service
MHRA
Elephant and Castle
London SE1 6TQ

T: +44 (0) 20 7972 8181
F: +44 (0) 20 7972 8105
E: des@mhra.gsi.gov.uk
W: www.mhra.gov.uk