International Journal of Engineering Research & Management Technology
ISSN: 2348-4039
December-2019 Volume 6, Issue-6
Email: editor@ijermt.org www.ijermt.org

A STUDY ON RADIATION EXPOSURE AND REDUCING RADIATION DOSE IN COMPUTED TOMOGRAPHY (CT) SCAN

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ABSTRACT

The number of CT scans performed has significantly increased as a result of the development of single- and multi-detector row helical computed tomographic (CT) scanners using slip ring technology. Much technical advancement has been made as a result of worries about the accompanying radiation dose increase. The automated modulation of tube current is one such breakthrough. Automatic tube current modulation is used to ensure continuous image quality regardless of the attenuation characteristics of the patient, hence lowering the radiation dose to the patient. The principles, clinical applications, and restrictions of various automatic tube current modulation approaches are covered in this paper.

KEY WORDS: Radiation, Dose, Computed, Tomography, Scan.

1. INTRODUCTION

Equipment used: At the Shri Atal Bihari Vajpayee Medical College and Research Institution (Formerly Called as Bowring and Lady Curzon Medical College and Research Institute) in Bangalore, Karnataka, computed tomography acquisitions were carried out on a cutting-edge 32-detector Fujifilm/Hitachi - Supria 32 CT scanner.

CT scanner Fujifilm / Hitachi – Supria 32 slice

Power: 51 kW **Gantry bore:** 75 cm **Scan range:** 180 cm

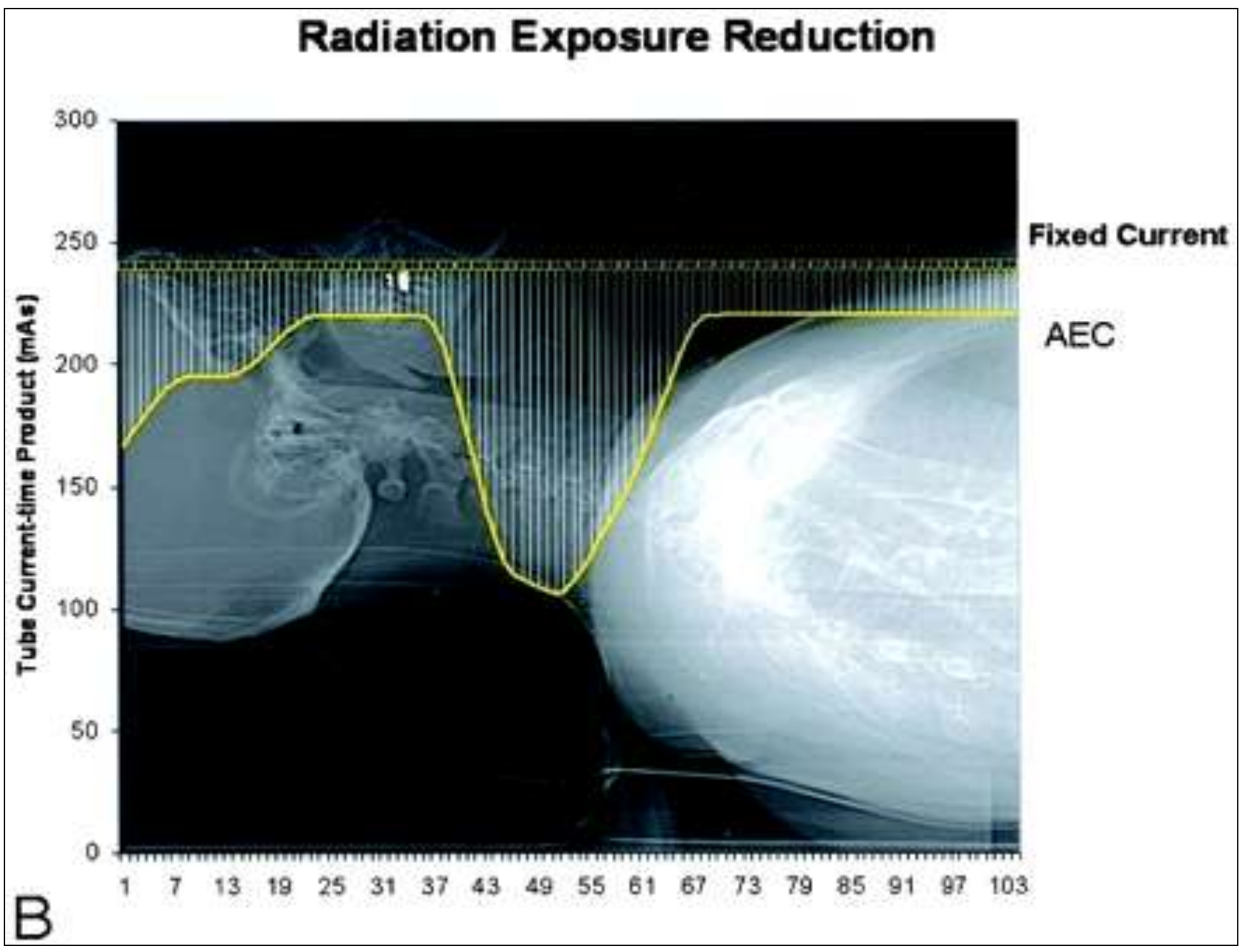
1.1 HIGHLIGHTS

- 5 MHU X Ray tube
- Sub second scan time for all examinations

- 0.625 mm minimum slice thickness
- 75 cm wide gantry bore for improved patient experience
- The compact footprint needs small installation space
- Iterative reconstruction algorithm for low dose examinations: Intelli IP Advanced
- Intuitive GUI design with 24-inch wide monitor
- Slices per rotation: 16 / 32
- Field of view: 500 mm

An essential factor in determining radiation dose and picture quality in x-ray-based exams is tube current, which is expressed in milliamperes. Recent developments in computed tomography (CT) technology, such as the use of automatic tube current modulation (ATCM), enable a reduction in the radiation exposure during CT exams.

FIGURE-1 RADIATION EXPOSURE



Graphs provide graphic representations of the automatic exposure control (AEC) technique's modulation of tube current (A) and, therefore, tube current-time product (mA) (B) with body region along z-axis (continuous lines in A and B). For contrast, the fixed current technique's tube current and tube current-time products are displayed as dotted lines in A and B, respectively.

FIGURE-2 AEC IS SMALLER THAN THAT WITH FIXED CURRENT

Table 1. AEC techniques currently available from different vendors

AEC Technique	GE Healthcare	Siemens	Philips	Toshiba
x-y axis/angular	Smart mA	CARE Dose	D-DOM	—
z axis/longitudinal	Auto mA	ZEC	Z-DOM	SureExposure
x-y-z/combined	Auto mA 3D	CARE Dose 4D	—	SureExposure3D

Note: AEC = automatic exposure control.

AEC techniques currently AEC Technique GE Hea x-y axis/angular Smart z axis/longitudinal Auto m x-y-z/combined Auto m. The term "ATCM" refers to a group of methods that allow the tube current to be automatically adjusted in the x-y plane (angular modulation) or along the z-axis (z-axis modulation) depending on the size and attenuation properties of the body part being scanned in order to maintain constant CT image quality while exposing less radiation to the patient.

The automatic exposure-control or photograph-timing techniques employed in traditional radiography are therefore comparable to ATCM approaches. The use of ATCM techniques should enable a general reduction in radiation exposure during CT exams, which is important given the rising worries about CT radiation exposure. Unfortunately, due to the quick development of technology, many manufacturers have created unique ATCM methods and employ proprietary terminology.

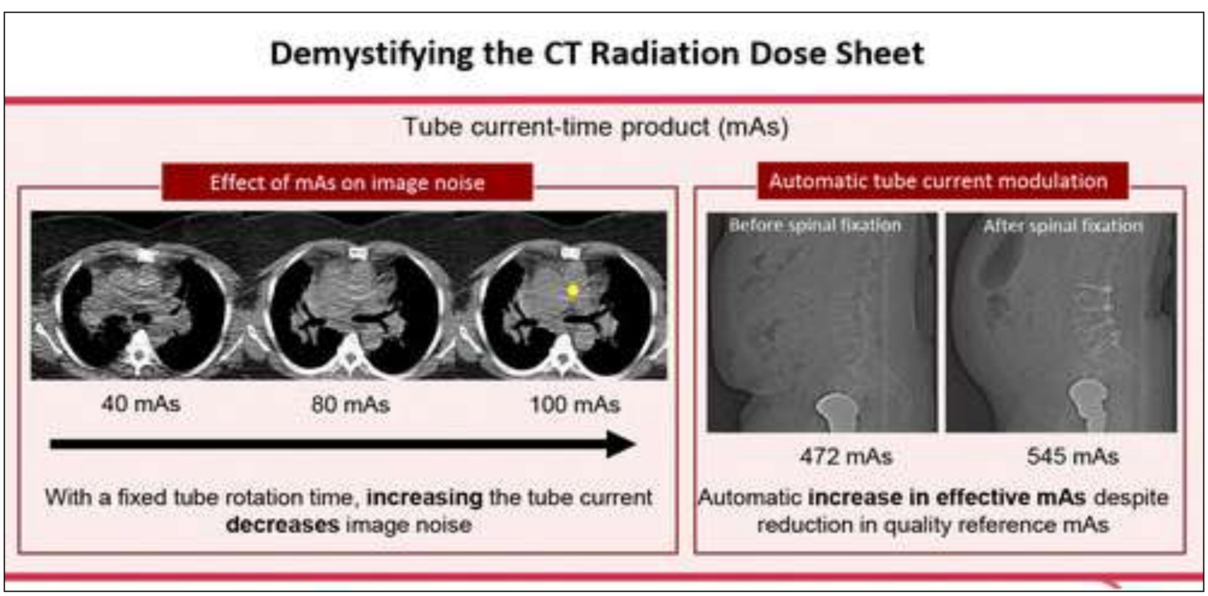
Table 2. Different methods chosen by manufacturers to optimize mA by setting exposure level

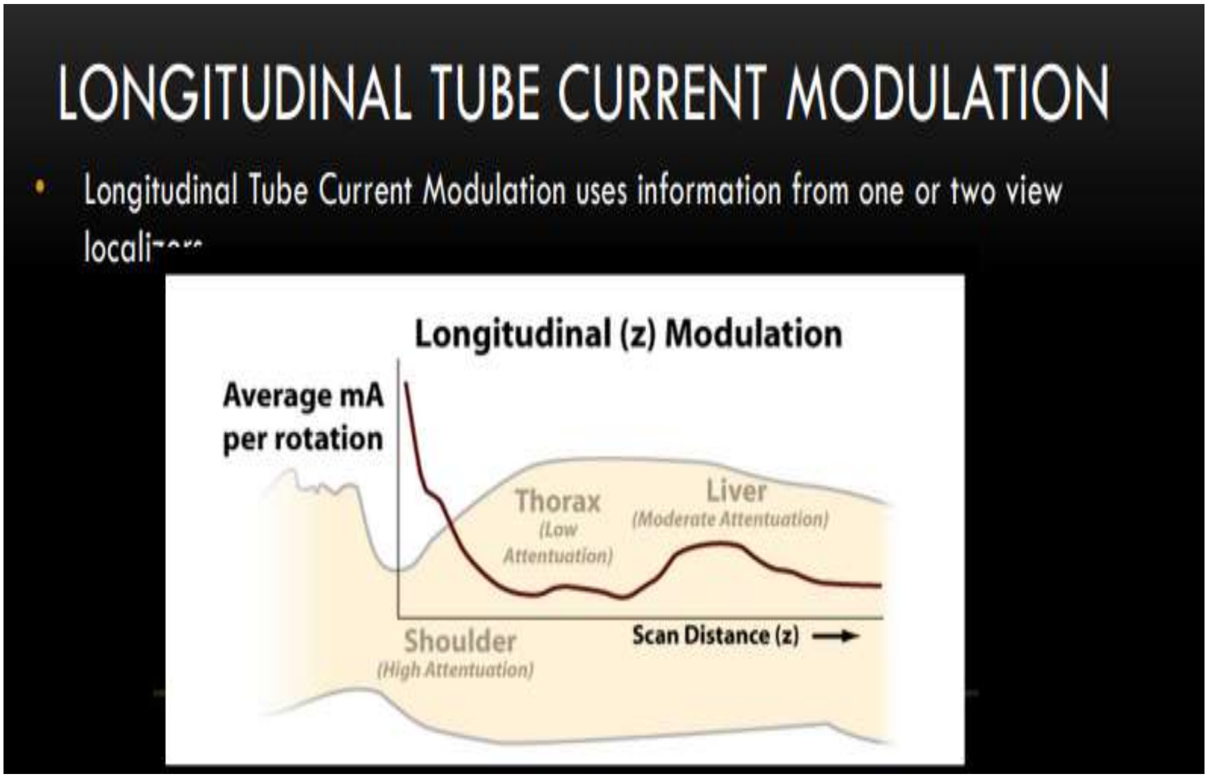
Technique	Specified Parameter	Implications
Auto mA 3D (GE Healthcare)	Noise index	Implies user-desired noise in the entire image
	Minimum and maximum mA	Range of allowed mA to achieve desired noise index
	Smart mA	Selection of this optional function adds x-y modulation to z modulation (Auto mA)
CARE Dose 4D (Siemens)	Reference mAs	Implies need for image quality equal to that obtained with the use of specified reference mAs in a standard adult (70-80 kg) or child (20 kg)
Z-DOM (Philips)	Baseline mAs	"Baseline mAs" is used as a reference to obtain constant image noise along the z axis
SureExposure 3D (Toshiba)	Standard deviation	Implies need for obtaining images at specified image noise (standard deviation)

Note: mA = tube current; mAs = tube current-time product.

FIGURE-3 DIFFERENT METHODS CHOSEN TECHNIQUE SPEC AUTO MA 3D (GE HEALTHCARE) NOISE IN MINIMU

However, the inclusion of ATCM procedures in contemporary CT scanners marks a significant step in the direction of standardising tube existing protocols by doing away with the practise of radiologists and techs making arbitrary selections. The fundamentals of ATCM techniques, their clinical applications, and the benefits and drawbacks of their use in diagnostic CT scanning will all be covered in this article.





1.2 PRINCIPLES OF ATCM TECHNIQUES

Patient dosages can be significantly decreased using CT scanners with automatic tube current modulation (ATCM). The goal of modulation is to maintain the same degree of image quality throughout a scan. It is based on measurements of x-ray beam attenuation in body tissues taken from scan projection radiographs (SPRs).

Contrary to conventional radiography, when performing a CT scan, the x-ray tube continually revolves around the patient, generating x-rays that pass-through a cross section of the body and produce attenuation profiles (raw image data) at the detectors. To rebuild each cross-sectional CT picture, x-ray beam incident projections from various angles around the region of interest are used. Image quality and radiation dose are impacted by scanning factors like photon fluence and incident beam energy, which are determined by scanning parameters like tube current and tube potential. A decrease in tube current reduces radiation exposure while increasing image noise or mottle, a key factor in determining image quality, when all other scanning parameters remain constant. The same is true for tube current, which increases radiation exposure while decreasing image noise. Picture noise and radiation exposure are opposing criteria on a diagnostic acceptability scale, and placing undue emphasis on either one could have negative effects on image quality or radiation dose.

Pictures with higher noise may mask lesions that may have been seen in lower-noise photos, whereas images with lower noise (less radiation exposure) may not disclose further diagnostic information. Previous clinical and experimental studies' authors have said that, based on the weight and cross-sectional dimensions of the patients having CT scanning, sufficient picture quality can be attained with a reduction of tube current.

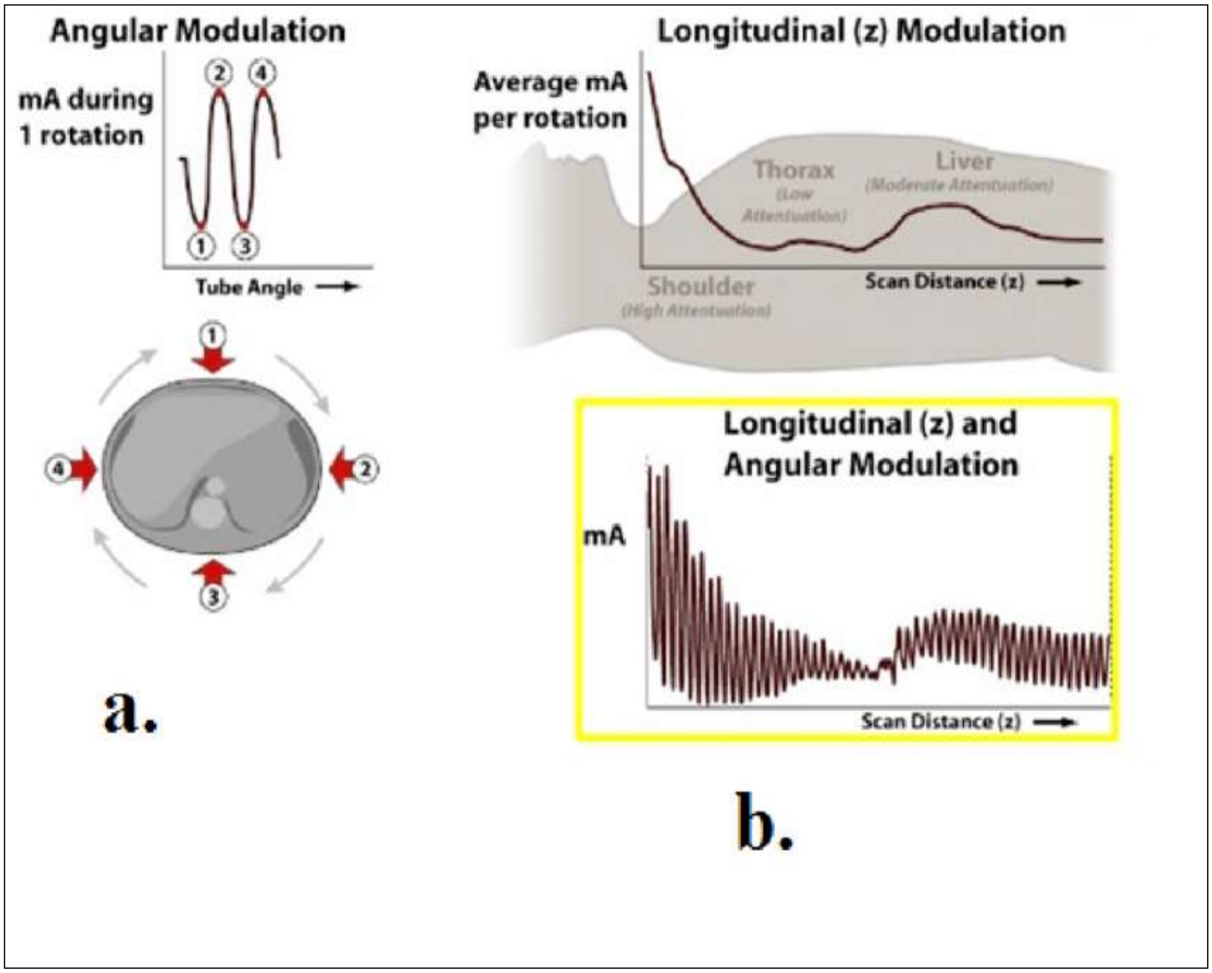


FIGURE-4 DEMONSTRATION OF ATCM TECHNIQUE; (A) ANGULAR OR X-Y MODULATION AND (B) LONGITUDINAL Z-AXIS MODULATION

These investigations' findings can be explained by the incident beam's varying attenuation as it passes through a specific cross-sectional dimension at a specific projection angle (1,2). Image noise is determined by the resulting attenuation, which is influenced by scanning settings, particularly tube current. For instance, a beam that experiences more attenuation in one dimension or projection will produce more noise and requires more tube current than a beam that experiences less attenuation in another projection. Depending on the patient's weight or measurements, tube current can be manually adjusted to strike a good balance between radiation dose and image noise. These modifications do not, however, ensure consistent image quality throughout the course of the examination. The selection of a fixed tube current, for instance, does not take into consideration variations in beam attenuation between the shoulder area and midchest region or between anteroposterior and lateral cross-sectional dimensions in chest CT scanning.

Because ATCM reacts quickly to significant variations in beam attenuation, ATCM approaches allow for the maintenance of continuous image quality at a required radiation exposure level. The basis for ATCM is the observation that x-ray quantum noise in transmitted beam projections determines image noise. This method intends to regulate x-ray quantum noise on the basis of local body anatomy in order to maintain constant image noise with increased dosage effectiveness. At the moment, ATCM can be implemented using either angular (x-y) modulation or z-axis modulation. Both methods vary the tube current in an effort to maintain steady image quality at the lowest dose while also decreasing streak artefacts brought on by a small number of photons and reducing tube loading (heating). Modulation by Angle For a single-detector row helical CT scanner, the angular-modulation approach was introduced in 1994 (. Based on the measured density of local structures and the object of interest's absorption values, this software-based technique modified tube current.

On two localizer radiographs, local x-ray beam absorption in 100 central channels was measured to acquire this information (lateral and anteroposterior views). In order to equalise regional variations in beam absorption, a preprogrammed sinusoidal modulation of tube current was performed during 360° rotation. This resulted in relatively consistent noise content and decreased radiation exposure. According on patient geometry (asymmetry), a reduction in radiation dose of up to 20% has been documented. An online, real-time, anatomy-adapted tube current modulation strategy that does not require the knowledge of radiography localizer images to produce ATCM is a recent improvement to the angular-modulation approach. Angular-modulation techniques decrease x-rays in projection angles that are unimportant in terms of lowering the total noise content by automatically adjusting the tube current for each projection angle to the patient's attenuation. When an online angular modulation technique was compared to the previous angular-modulation technique in phantom investigations, the online modulation technique showed a significantly better radiation reduction (up to 50%). Beam attenuation in patients with circular cross-sectional geometry is constant in all perspectives (x-ray beam projection angles). However, attenuation varies significantly between projections in a noncircular cross sectional geometry, often by more than three orders of magnitude.

The overall image noise content in these circumstances is determined by image noise at high attenuation projection angles. Thus, the tube current can be significantly decreased at angular projections with a small patient diameter or body region being scanned, resulting to a relatively lower attenuation, without a discernible rise in the overall noise content of the image. Regardless of the patient's attenuation profile, tube current is maintained constant without rotational modulation during a 360° rotation. For low-attenuation projections, the angular-modulation approach lowers tube current as a function of projection angles (anteroposterior vs lateral projections). With this method, the patient's online attenuation profile is used to determine the modulation function, an objective measure of image quality. With a 180° delay from the x-ray generating angle, the modulation function data are processed and delivered to the generator control for tube current modulation. Using the angular-modulation technique, a reduction in radiation dose of up to 90% can be achieved in the anteroposterior or posteroanterior direction in regions with marked asymmetry, such as the shoulders in CT scanning of the chest, where attenuation is significantly less in the anteroposterior direction than in the lateral direction. In conclusion, the angle modulation technique helps to increase dose effectiveness in the x-y axis by lowering radiation exposure in a certain scanning plane. The tube current in asymmetric anatomical regions is also adjusted using the dose-right dose modulation, or DOM, approach of angular modulation.

Based on the idea that tube current should be regulated in accordance with the square root of the measured attenuation for that projection, this method was developed. In a 360° tube rotation, the DOM approach adjusts the current in accordance with the square root of the attenuation determined from the analogous and prior 180° or 360° views. Alternatively put, the attenuation determined at an angular view

Later, to achieve a similar angular view, the tube current is optimised using (projection angle). Modulating the Z-axis When compared to angular modulation, the z-axis-modulation (AutomA, GE Medical Systems; Real E.C., Toshiba Medical) approach works considerably differently. The Auto mA approach regulates the tube current automatically to keep the quantum noise level in the image data at a user-specified level.

It offers a noise index so that users can choose how much x-ray noise appears in the reconstructed images. The scanner calculates the tube current required to produce images with a chosen noise level using a localizer radiograph. Therefore, z-axis modulation makes an effort to have all images, regardless of patient size and anatomy, have a same amount of noise. In the centre portion of an image of a uniform phantom, the noise index value is almost equal to the image noise (standard deviation). Using the patient's localizer radiograph projection data and a set of empirically derived noise prediction coefficients by utilising the reference technique, the system in the z-axis-modulation technique calculates the tube current. The reference method uses a transverse reconstruction algorithm with a standard 2.5 mm thick section that was acquired at the chosen peak voltage and 100 mAs. One localizer radiograph's projection information can be utilised to calculate the patient's density, size, and shape. The patient's density and size information about the projection area are contained in the total projection attenuation data of a single localizer radiograph, whereas the patient's shape information is contained in the projection's amplitude and area, providing an estimate of the patient's elliptic asymmetry expressed as an oval ratio at a specific z-axis position. The ratio of an ellipse's a and b parameters—the lengths of its long and short axes—is known as the oval ratio. Using the equation for an ellipse's area, the patient's ellipse parameters can be found. The localizer radiograph's measured projection area and amplitude provide the area and length of one axis, axis a, of the ellipse, allowing the calculation of the length of the second axis, axis b.

2. CONCLUSION

The picture standard deviation due to x-ray noise for a certain reconstruction procedure is determined by these properties of the localizer radiograph, which predict the quantity of x-rays that will reach the detector for a given approach. The projection area and oval ratio from the localizer radiograph are used to calculate the predicted x-ray noise at a specific z-axis position for the reference technique (i.e., reference noise), using the polynomial coefficients that were established from the noise measurements in a set of phantoms representing a variety of patient sizes and shapes. The tube current needed to achieve a specified noise index can be easily calculated using well-known x-ray physics equations that state that noise is inversely related to the square root of the number of photons and that the number of pho tonnes is proportional to section thickness, section acquisition time, and tube current. These equations require knowledge of the reference noise and the difference between the reference technique and the selected technique data. In order to account for noise variations between helical selections and the transverse reference technique data, an adjustment factor for various helical pitches is included in the computation in the present version of z-axis modulation of tube current. Since the tube current is automatically reduced for smaller patients and specific anatomical regions, recent data from clinical studies of z-axis modulation imply that the radiation dose reduction with this technique is anticipated to be greater than that with fixed-tube current methods. It happens frequently that the noise index chosen for scanning and the actual noise assessed on the image by sketching a region of interest will be different. This is because noise index settings only change the tube current, whereas the standard deviation also depends on the reconstruction algorithm, the thickness of the reconstructed section (if it differs from the prospective thickness), the use of image space filters, variations in the anatomy and motion of the patient, and the presence of beam-hardening artefacts. Due to insufficient signal strength at the detector and the superimposition of electronic noise, significant disparities between the chosen noise index and the standard deviation can also arise in very large patients. These differences can be reduced by utilising a larger peak voltage. The bow-tie filter's poor beam attenuation can also cause pictures to be noisier if patients are not properly centred in the scan field of vision as attenuation from a bow-tie filter

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