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Principles of Image Quality and Dosimetry in computed tomography

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PRINCIPLES OF IMAGE QUALITY AND DOSIMETRY IN COMPUTED TOMOGRAPHY

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INTRODUCTION

Since its introduction in 1972, Computed Tomography (CT) has revolutionized radiology. Among x-ray imaging techniques, CT has the unique ability to clearly image individual slices of the body with no interference from tissues outside of the slice of interest. Though conventional tomography can emphasize the structures within a particular plane, it cannot eliminate the effect of out of plane structures - they still appear as blurs in the image. The ability of CT to image only the plane of interest and the high degree of scatter rejection inherent in the CT imaging process allow CT to visualize low contrast lesions that would be invisible by any other x-ray imaging method.

A CT scanner is a highly sophisticated device demanding careful calibration for proper operation. Periodic quality assurance testing and routine calibration and preventative maintainance are essential. In this paper we will discuss some of the properties of CT scanners and the essential features of quality assurance including evaluations of image quality and dosimetry.

THE CT IMAGE

The CT image is derived by computer from the measured x-ray attenuations at all positions and angles through the slice of interest. This image production by computer processing requires that the image be broken up into a finite number of elements called pixels. A 256x256 image matrix indicates that the image is made up of a rectangular matrix of pixels that is 256 pixels high and 256 pixels wide yielding a total of 65,536 pixels. Other common image matrices are 160x160, 320x320, and 512x512.

Each pixel in a particular image is assigned a number as a result of the imaging process. Typically this number, which is called the CT number, is between -1000 and +3000 and represents the x-ray linear attenuation coefficient (μ) at the corresponding position in the body. More precisely, it represents the average of the attenuation coefficient within a volume defined by the pixel in two dimensions and by the selected slice thickness in the third dimension (fig.1). This volume element is referred to as a voxel. For a 256x256 image of an 256mm wide field and a 8mm slice thickness, the pixel size is 1mm x 1mm while the voxel size is 1mm x 1mm x 8mm.

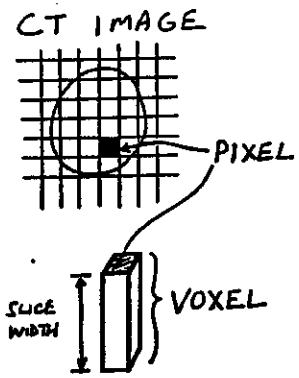

FIGURE 1.

The CT number assigned to each pixel is related to μ by a linear offset function. At present, the most common method is to assign a value of -1000 to Air and 0 to Water. In this method, a tissue which has a μ equal to two times that of water would have a CT number of +1000. When this method of assigning CT numbers is used, the CT numbers are sometimes reffered to as Hounsfield Numbers and are said to be expressed in Hounsfield Units (HU). The following equation shows how to mathematically convert the attenuation coefficient into a CT Number expressed in HU:

$$CT\#_x\ (HU) = \frac{\mu_x - \mu_v}{\mu_v} \times 1000 \qquad (1)$$

In this equation $CT\#_x$ (HU) is the CT number of the tissue in Hounsfield Units, μ_x is the linear attenuation coefficient of the tissue, and μ_v is the linear attenuation coefficient of water. Since the CT number changes by 1000 as μ_x changes from 0 to μ_v, a change in CT number of 10 HU represents a change in attenuation coefficient of 1% of μ_v. Other definitions of CT numbers have been used, particularly on older scanners, but these are not referred to as Hounsfield Units: For example, CT scanners produced in the past by EMI used a definition of CT number in which the 1000 in equation (1) was replaced by the number 500.

To be diagnostically useful, this matrix of CT numbers which is the source of the CT image must be converted into a gray scale image that can be viewed. A very useful property of the CT modality, which is shared with other digital imaging methods, is the myriad ways in which the image can be manipulated. The density and contrast of the final visual image can be freely changed - after the x-ray exposure - by the adjustment of two controls on the imaging console: the window width and the window level. This is in contrast to

conventional film radiography in which the density and contrast of the image are determined prior to the x-ray exposure by adjusting the kVp and mAs - if these are incorrectly adjusted, the result is an image of improper density or contrast, which must be retaken. This problem cannot occur in CT scanning. Here, if the mAs is set too low, the image may appear excessively noisy but it can still be displayed with the proper density. The free adjustment of the contrast and density of the CT image after the x-ray exposure also allows the visualization of structures over a very great range of densities. Conventional film radiography may have problems discerning detail in both the lung field and the mediastenum on the same film, but such problems are easily handled by digital imaging. If needed, two separate film records can be made of the same CT scan using different settings of the window level and window width to optimally show, for example, the lung field in one image and the mediastenum in the other.

In producing a grey scale image from a matrix of CT numbers the computer must be told how each number is to be turned into a shade of grey. This is accomplished by selecting the window width and window level. The window width specifies the range of CT numbers that will be converted to shades of grey in the image. The window level specifies the CT number which lies at the midpoint of this range. All pixels with CT numbers below this range will be reproduced in the image as black; all pixels with CT numbers above this range will be reproduced as white. All pixels with CT numbers within this range will be reproduced as shades of grey from very dark grey for those pixels with CT numbers close to the bottom of this range to very light grey for pixels with CT numbers close to the top of this range. The method by which the window width and window level controls manipulate the image in a particular case is shown below in figure 2:

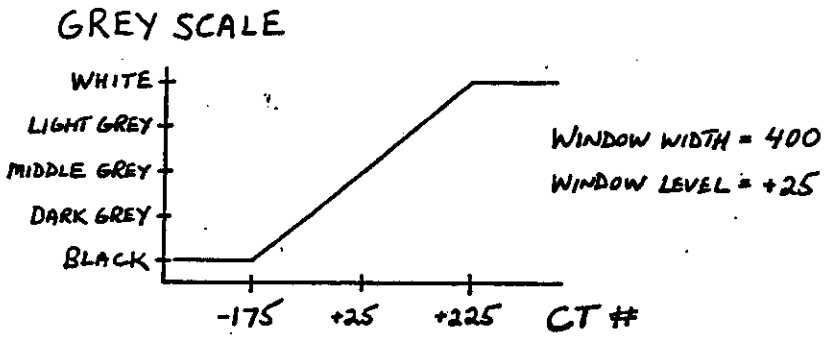

FIGURE 2.

In this case the window width is 400 and the window level is +25. All CT numbers below -175 transform into black in the image while all CT numbers above +225 transform into white. The pixels with CT numbers between -175 and +225 will display as shades of grey in the image.

CT SCANNER COMPONENTS

It is convenient to divide the CT scanner system into three separate components each of which often occupies a separate room:

1. The Scanner Gantry.

   This includes the x-ray source, the x-ray detectors with their associated electronics for data squisition, the mechanical mechanism which moves the x-ray source and (sometimes) the detectors around the patient, and a mechanism for precisely correlating the exact position of the gantry with the data coming from the detectors during the actual scanning. Each of these components is controlled by the computer and/or feeds data to it. The gantry of course contains an aperture into which the patient is placed to be scanned. The above components are contained in a ring shaped structure around this aperture. The generator which provides the high voltage and filament control for the x-ray tube is located outside of this ring structure but nearby. The patient table, its positioning controls, and the devices for slice localization may also involve computer control.

2. The Computer.

   In addition to controlling the functions of the scanner gantry, the computer must receive and store the data from the x-ray detectors. These data are then processed into a matrix of CT numbers and stored in the computer, ready for transmission to the display console. Usually, data for more than a single image cannot be stored in the limited computer memory. As images are produced they must be transferred to some other memory device. For short term storage which allows rapid access to the image, a hard disk storage device can be used. For long term economical archival storage (with much slover access to the image) magnetic tape can be used. A brand new storage medium which combines the rapid access of hard disk with the large economical storage capacity of magnetic tape is optical disk storage. While optical disk storage is more expensive than magnetic tape, it occupies significantly less physical space to store the same number of images.

3. The Control and Display Console.

   This console contains the operator controls of the CT scanner and the vieving monitor which displays the CT image. When the operator calls an image from the computer, the matrix of CT numbers for that image is sent to the console which then translates this matrix into an image on the vieving monitor under the guidance of the window width and window level controls. CT scanning systems can be configured with a separate, additional display console so that previous images can be viewed while scans are being performed. It is most convenient if this additional display console has its own separate computer so that it is truly independant and is not sloved down when the main computer is occupied with scanning and image reconstruction. While the physician can view the image and make diagnosis on the display monitor, it is almost always desireable to produce a "hard copy" film image for later vieving independant of the vieving console. This is accomplished by using an image documentation instrument which contains a separate monitor and a camera to photograph the image on the monitor. Newly developed laser documentation systems can yield exceptional image quality by bypassing the conventional monitor-camera link and forming the film image more directly.

TYPES OF CT SCANNERS

CT scanners can be divided into 5 basic types or generations based on the configuration and motion of the x-ray source and detectors during the scan. The numbering from 1st to 5th generation is derived from the chronological order in which these scanner types were developed. These scanner types vary greatly in their clinically significant properties. Scan times have ranged from under 1 second to 4 minutes; thus patient throughput has varied greatly. There are also significant variations in image quality and the degree of freedom from image artifacts.

The original first generation scanners, developed by EMI, consisted of an x-ray tube whose output was collimated to a narrow pencil beam and a single detector. To obtain the image data during a scan, the tube and detector executed a common linear motion so that the x-ray pencil beam scanned across the patient. The tube and detector then rotated by 1 degree around the center of the patient and scanned back across the patient in another linear motion. This translate-rotate motion was repeated 180 times so that all views of the patient could be measured. The highly collimated beam and single detector allowed a high degree of scatter rejection, but was a very inefficient way to collect the image data. Thus the scan times were very long, about 4 minutes to obtain data for a single slice. To help improve patient throughput, EMI added a second detector which was used to obtain data for a second slice simultaneously with the first slice. Since the pencil beams used for the two slices were not parallel (they came from the same x-ray source), artifacts could easily be produced by high density objects in the outer parts of the field. Also the long scan time limited this scanner to neuro work where motion is limited without the need for breath holding. Still, some patient motion was common during the long scan time and image artifacts and blurring due to motion were often seen.

Second generation scanners continued to use the translate-rotate form of motion, but significantly reduced the scan time by obtaining the data for several angles in a single translation. As an example of this scanner type, the EMI 5005 scanner collected data over 10 degrees in one translational sweep. In this way the number of translation motions was reduced from 180 to 18, and the scan time was reduced to a selection of 20 or 60 seconds. This was achieved by using an x-ray fan beam of 10 degrees and a total of 30 detectors over an arc of 10 degrees, instead of a single detector, to collect the data. With scan times of 20 seconds, body scanning could now be performed using breath holding to reduce motion with many patients. Motion artifacts were also reduced in neuro work, and patient throughput was greatly increased.

A breakthru in CT scanner design occured with the development of third generation scanners. In this design there is only a single motion type for the tube and detectors: a 360 degree rotation. The separate translation motion is eliminated. In this way very fast scan times are possible down to about 1 second, with typical scan times of 3 to 5 seconds for most clinical studies. At this scan speed it becomes much easier for patients to successfully breath hold during the scan. Motion artifacts from both internal motion and unintentional patient motion are greatly reduced.

In order to collect a complete set of attenuation data using only rotational motion, the x-ray beam and bank of detectors must be wide enough that the entire patient is covered at a single position. This requires an x-ray fan beam and detector arc of about 30 to 45 degrees for whole body scanning. To cover this large arc 200 to 1000 separate detectors are needed. Obviously, with this large number of detectors, the cost and calibration effort is

significantly increased over earlier scanner types. To make full use of the rapid scan times possible with these scanners it is desireable to increase the computing power in order to reduce the image reconstruction time. Newer scanners have reconstruction times of a few seconds per image. This allows each slice to be viewed as the scans are performed rather than taking a series of scans and performing the image processing later.

The third generation design presented some difficult engineering problems related to detector stablity. To produce an artifact free image, the detectors must be precisely calibrated and then remain very stable between calibrations. If this is not properly accomplished, the result will be the formation of ring artifacts in the image: rings of high or low density. The size of the ring depends on the location of the defective or miscalibrated detector. If the detector is near the outside of the detector array, the ring diameter will be large; if the detector is near the center of the detector array, the ring diameter will be small. If the bad detector is very near or at the center of the detector array, a rather unfortunate artifact occurs: the ring here has shrunk to a small area of low or high density that can easily be mistaken for a lesion. Thus it is very important to reduce these ring artifacts to the minimum possible level. The maintenance of detector calibration is complicated by the fact that the patient remains between the x-ray tube and the detectors during the entire scan; thus the detectors cannot be calibrated during the scan by exposure to the unattenuated x-ray beam. The solution to this problem is two-fold: (1) The selection and design of highly stable x-ray detectors and electronics, and frequent calibration of the detectors (when a patient is not in the scanner). (2) The development of computer software techniques to correct for any residual miscalibration of the detectors in the image processing stage. In general, the results have been rather successful, and a properly functioning, well calibrated third generation scanner should have no clinically significant problems with ring artifacts.

Several manufacturers of CT scanners took a different approach to solving this problem of ring artifacts and developed a fourth generation CT scanner. This scanner, like the 3rd generation scanner, utilized a rotate-only motion. With the 4th generation scanners, however, only the tube rotates; the detectors remain stationary in a ring that completely surrounds the patient. This method has two advantages over the 3rd generation type: Every detector can be calibrated during the scan simply by making the x-ray fan beam somewhat larger than that needed to cover the patient. The unattenuated beams at either side of the patient can then be used to twice calibrate each detector as it passes under these beams. Secondly, the nature of the data collection does not actually demand precise detector calibration to avoid artifacts. Useful scans can even be made with several totally dead detectors.

The 4th generation scanner design also has several disadvantages compared to 3rd generation scanners. Since the detectors are distributed over an entire 360 degree circle around the patient, a rather large number of detectors would be required using detectors of optimum size which were packed closely together to absorb the largest possible fraction of the radiation exiting the patient. This large number or detectors can significantly increase the cost of producing the CT scanner. To reduce the number of detectors, it is necessary to either increase the detector size which degrades the image resolution and/or increase the spacing between the detectors which then decreases the geometrical efficiency of the detectors and reduces the total fraction of radiation which is absorbed by the detectors. In this later case a higher dose would have to be delivered to the patient to obtain equivalent image quality. A practical 4th generation scanner design typically must

compromise between these items of detector number, detector size, and detector spacing.

Another significant disadvantage of the 4th generation design compared to the 3rd generation concerns scatter rejection. In 3rd generation scanners the detectors are placed along the arc of a circle whose center is the x-ray source. Since the detectors and x-ray tube move together around the patient, their relative geometrical relationship remains fixed. Thus each detector can be fully collimated to the x-ray source, greatly reducing the amount of scattered radiation which reaches the detectors. In 4th generation scanners this collimation to remove scattered radiation from within the scan plane is not possible. As the tube moves around the patient each stationary detector sees the position of the x-ray source moving in an arc, thus there is no fixed position to which the collimation can focus. This lack of collimation septa between the detectors increases the fraction of scatter in the detected beam. The image reconstruction process must compensate for this extra scattered radiation or artifact will result. Even when this is done successfully, the extra scatter will either result in a somewhat noisier, degraded image or, to maintain image quality, a higher dose will be required. Given the preceeding analysis one might ask whether 3rd or 4th generation designs are better among the currently available scanners. The answer must be that the skill of the manufacturer in implementing its particular design are at least as important to consider as are the specific advantages and disadvantages inherent in the different designs.

Two types of detectors are used in modern CT scanners. Gas-filled chambers (Xenon gas under pressure) can be used as x-ray detectors. The ionization of the gas in the chamber by radiation produces a signal proportional to the radiation intensity. These detectors have the advantage of a high level of stability and thus are particularly suitable for use in 3rd generation scanners. The disadvantage of Xenon gas detectors is their low absorption of radiation: typically only about 50% of the radiation incident on the detector is absorbed.

The other type of detector uses a solid scintillation material as the x-ray absorber. These materials emit light proportional to the amount of radiation absorbed. This light is then detected by a photomultiplier tube or a photodiode which converts the light into an electrical signal. These detectors are better absorbers of radiation than gas filled detectors: they can absorb close to 100% of the radiation incident on them. They have the disadvantages of (1) lesser stability and (2) afterglow, which is the emission of light for a short time after the actual x-ray exposure. Afterglow can be a particular problem with faster scan times that approach 1 second. These problems can be minimized by the proper selection and fabrication of scintillator materials, but the use of these detectors still poses a significant engineering task not encountered with gas filled detectors. If these solid scintillation detectors are used in third generation scanners without careful design and calibration procedures, ring artifacts can easily result.

There are several other evolutionary improvements in CT scanner performance that are worth mentioning here, in addition to reduced scan time and reduced image reconstruction time. Scanner designs incorporating an increased number of detectors and improved reconstruction algorithms have significantly improved the resolution limits of these scanners down to 0.4 mm. The implementation of variable selectable slice thicknesses down to 1 mm permit the fullest realization of these improved resolution limits by reducing blur due to partial volume effects. Larger gantry apertures and the ability to tilt the gantry allow better access to patient pathologies. Larger tube heat

capacities allow for faster scan repetition rates and higher patient throughput. The use of digital plane radiographic images, produced by the CT scanner as scout films, allows for the precise adjustment of scan location to suit the clinical need. After producing the scout film the CT operator can interact with the computer by drawing the precise angle and the starting and ending positions at which he wishes to scan. The patient table and gantry can then move to these specifications under computer control. Often the scanning process for the entire examination can be put under computer control after the required scan techniques are selected.

The most recent major development in CT scanner technology was made by Imatron, Inc. This latest design is sometimes refered to as the fifth generation scanner. With this instrument scan times down to 1/20 sec (50 msec) can be achieved. At this speed, effective freezing of cardiac and respiratory motion is possible. This incredible scanning speed is accomplished by effectively placing the patient within the x-ray tube! When positioned for a scan, the patient is surrounded both by two detector rings and by several rings of tungsten which act as the x-ray source. To scan the patient, an electron beam is accelerated towards the patient along the scan axis and then deflected so that it hits one of the tungsten rings. The electron beam is then quickly scanned over the length of the tungsten ring to complete the scan. Since this process involves no moving parts, a very rapid scan time is possible.

QUALITY ASSURANCE OF CT SCANNERS

As mentioned before, peroidic calibration and preventative maintenance by the manufacturer are essential to the proper operation of a CT scanner. This alone, however, is not sufficient to assure the optimum daily functioning of the CT scanner. To accomplish this it is necessary to have an independant program of quality assurance that can daily monitor the most sensitive aspects of scanner performance and periodically perform more complete measurements of scanner performance. These quality assurance (QA) tests are needed (1) to monitor the day to day performance of the scanner between routine calibrations; (2) to check that scanner calibrations, maintenance, and repairs are in fact being done correctly; (3) to detect problems that may not be found and corrected in the routine calibrations and maintenance; and (4) to monitor the scanner performance over time so that any degradations due to scanner aging or modifications can be detected and addressed.

Most of the test procedures for CT quality assurance that will shortly be discussed are also useful in the process of buying a CT scanner and setting it up for clinical use: (1) Scanner comparisons are often performed as part of the selection process of a CT scanner. This testing of various CT scanners to compare their levels of technical performance can be done by the physicist who is assisting in the process of purchasing a CT scanner. This same physicist should construct a list of performance specifications that the CT scanner must meet after it is installed and calibrated and before it is finally accepted by the purchaser. (2) To verify that the scanner does indeed meet these specifications, the physicist should perform acceptance testing which involves all of the QA tests that will be discussed in addition to verification of the delivery and proper operation of all equipment and software specified in the purchase agreement. The acceptance tests additionally provide a baseline to which all future quality assurance test results can be compared.

In CT scanning, as with any diagnostic x-ray procedure, the objective is to obtain an image of good diagnostic quality at a minimum risk to the patient. Minimizing risk involves minimizing radiation exposure as one element. To help accomplish these goals, the tests involved with CT scanner quality assurance must include 3 general areas of performance evaluation:

1. Image Quality
2. Dosimetry
3. Performance of "Non-imaging" Components

CT QUALITY ASSURANCE PHANTOMS

To perform the CT quality assurance tests it is neccessary to use an object which generally simulates the x-ray attenuation and scattering properties of the body and which contains test objects with which scanner performance can be evaluated. Such an object is referred to as a phantom. To simulate the attenuation and scattering properties of the body, the phantom must have the correct size and be composed of materials similar to tissue in their x-ray interactions. Commercially available phantoms and those provided by the CT scanner manufacturers generally use either water or a water equivalent solid as their principle component. Phantoms are often available in different sizes suitable for simulating both head and body sections. The head phantoms may contain a water-equivalent main section which is surrounded by a bone equivalent material to simulate the skull. This "bone ring" can be very useful in demonstrating the presence of artifacts due to the skull.

Phantoms which utilize the solid water equivalent materials have several advantages over water-filled phantoms. Being solid, there is no problem with water leakage. The solid phantom can be composed of a main body with several cavities in it, and inserts containing test objects can be placed into these cavities. Thus only a single, relatively thin main phantom body is required. Any number of test objects can be incorporated into this phantom body by using the appropriate small test inserts. This leads to a relatively compact and lightweight phantom system which is easy to use. Water filled phantoms generally must be significantly larger and heavier if they are to perform the full range of QA tests. Since it is very inconvenient and time consuming to open a water filled phantom to replace test objects during a QA test, the phantoms are usually designed to contain all the the needed test objects within the phantom at the same time. This requires several sections for the different tests and can make the phantom quite long. With water filled phantoms one must also take some care to avoid air bubbles and residue from impure water and growth of organisms in the water. One important advantage of water filled phantoms is their uniformity: with a water filled phantom you have a priori knowledge of phantom's uniformity and radiographic properties. With solid phantoms you must trust to the skill and reputation of the phantom manufacturer. A high degree of uniformity and accuracy in the phantom material is usually only necessary in the QA tests which examine the CT number accuracy of water, the image noise, and the uniformity in the image of a uniform phantom. (These are the tests 1a, 1b, 2a, and 3a to be discussed in the next section.) If there is any question of the suitablity of a particular solid phantom for these tests it is usually a simple matter to substitute the uniform water filled phantom that is normally supplied by the CT scanner manufacturer for these particular tests.

IMAGE QUALITY TESTS

The following is a list of image quality tests which should be performed at the periodic comprehensive QA test. Tests listed in parenthesis, while they give highly useful information, may be quite difficult to perform accurately and are not essential for the QA program.

1. CT NUMBER ACCURACY
   a. The CT number accuracy of water.
   b. Uniformity of the CT number of water over a uniform phantom image.
   c. Contrast scale accuracy and stablity: CT number measurements of a few "standard" materials.

2. IMAGE ARTIFACTS
   a. Artifacts present in the image of a uniform phantom, such as ring artifacts.
   b. Artifacts due to the presence of high density objects in the scan such as the skull.

3. LOW CONTRAST PERFORMANCE
   a. Image noise - obtained from the image of uniform phantom.
   (b. Low contrast detectablity - Contrast-Detail curves.)

4. IMAGE SHARPNESS
   a. High contrast resolution
   (b. Modulation transfer function (MTF), derived from an edge response or MTF phantom.)

5. IMAGED SLICE THICKNESS AND SLICE SENSITIVITY PROFILE

6. IMAGE DISPLAY QUALITY

7. IMAGE DOCUMENTATION SYSTEM QUALITY
   a. Performance of image documentation camera.
   b. Performance of film processor.

Tests 1a, 1b, 2a, and 3a all make use of the image of a uniform, water equivalent phantom. Separate scans should be taken at all normally used scan modes, including scans at the different selectable slice thicknesses. Dosimetry tests can often be performed at the same time, thus improving test efficiency. These scans are then analyzed using the CT scanner's built-in statistical software. A region of interest is selected in the central region of the phantom image; the average CT number and the standard deviation of the noise (pixel noise) are then determined (tests 1a and 3a). The values should be within the limits given by the CT scanner manufacturer. Probably the most important of all the image quality tests is the noise measurement, since the noise directly affects the detectablity of low contrast lesions. Any significant increase in the noise measurement should be immediately addressed. Measurements of CT number and noise can also be made in other regions of the image, or the image can simply be inspected for uniformity from center to edge (test 1b). Finally, the images should be inspected for the presence of any artifacts - such as ring artifacts (test 2a). It is useful to keep the test films on file for comparison with later tests.

The following tests are performed with CT phantoms containing the required test objects. To test the contrast scale accuracy and stability, rods of different substances, such as plexiglas, polyethylene, and teflon can be scanned (test 1c). The average CT numbers within the rods are then

determined. These CT numbers should be stable over time and should correspond to the manufacturer's specifications. Specific CT numbers cannot be given for all CT scanners, since the CT numbers will vary somewhat depending on the kVp and filtration used in the CT scanner and on the precise densities of the test materials.

A phantom scan using a bone equivalent ring can be inspected for image artifacts (test 2b). The most common artifact is skull bleed-in; here, the part of the image just inside the bone ring has an artificially raised density. Special phantom inserts, such as one containing an aluminum pin can be scanned to observe the resultant artifacts (test 2b). Any worsening in artifacts from previous tests should be noted.

One of the most important charateristics of CT imaging is its ability to perceive low contrast lesions. Thus the low contrast detectablity of a scanner is an important performance characteristic (test 3b). The low contrast detectablity specifies either the smallest objects that can be perceived at given contrast levels or, equivalently, the lowest contrast levels at which objects of specified sizes can be detected. In either case, the results can be expressed on a two dimensional graph of object size vs. contrast level in which a plot of the threshold points divides the graph into two regions of detectable vs. non-detectable objects. This graph is called a contrast-detail curve. For a particular CT scanner the actual shape of the curve will vary if either the reconstruction algorithm or the exposure factors (kV, mAs) are changed.

The low contrast detectablity test is performed by imaging test objects of varying low densities and sizes and then determining the threshold of perception of these objects. In general this is a rather difficult test from which to get reliable, reproducable results. First, the test is highly subjective and is greatly affected by viewing conditions, viewing distance, and an individual's own visual perception. Second, since noise is by nature a random, changing process, several scans of the same low constrast objects under the same conditions in the same scanner will very often give different test results. Thus the only useful way to perform this test is to have a precise protocol for evaluating the images by the same individual or group of people. In addition, the evaluation must either utilize several images of the same phantom or must make use of a phantom in which there are multiple objects for each particular size and contrast. The test results are then an average of the results from several images of the same or similar objects.

Because of the difficulties of performing and evaluating the low contrast detectablity test, many individuals may find it preferable to monitor a scanner's low contrast performance simply by testing for image noise. A specification of the standard deviation of the pixel noise is not, by itself, a sufficient indicator of low contrast performance when comparing different scanners. However when testing a specific scanner at specific techniques, a measurement of pixel noise is a very good indicator of any changes in low contrast performance. For these purposes of Quality Assurance the additional testing of low contrast detectability does not normally add a significant amount of useful information.

Image sharpness can be tested by scanning one of several types of resolution inserts (test 4a). One common type contains a pattern of rows of rods, another contains a pattern of parallel planes. The former yields an image of rows of dots, the latter an image of series of parallel lines. The resolution, like the image noise, can vary greatly depending on the reconstruction mode used. Special high resolution modes will enhance image sharpness but will also increase noise and image artifacts. Thus these modes

are used for imaging bone detail but are not good for soft tissue. The modulation transfer funtion (test 4b) is a more complete description of image sharpness than resolution. However it usually requires additional computer processing of the CT image of an edge response or MTF phantom. The software programs necessary for this processing are not normally available to the CT operator. For simple QA testing, the high contrast resolution test is sufficient. Any significant degradation in image sharpness occurring over time in a particular scanner should show up in the resolution test.

The imaged slice thickness and slice sensitivity profile (test 5) can be measured using an inclined aluminum ramp. With a 45 degree ramp, the imaged slice thickness is equal to the width of the ramp image. With a 26.6 degree ramp, the imaged slice thickness is one-half the width of the ramp image. Thus the 26.6 degree ramp is a more sensitive test object and is recommended. The easiest way to measure the imaged slice thickness - defined as the full width half maximum (FWHM) of the slice sensitivity profile - is to determine the mean CT number of the densest part of the ramp image and the mean CT number of the image outside the ramp. Then take the average of these two CT numbers. Set the window level to this average CT number and set the window width to its minimum allowed value. The measured thickness of the ramp image is then exactly the FWHM of the slice sensitivity profile (remember to multiply by 1/2 for the 26.6 degree ramp). The complete slice sensitivity profile can be obtained from the CT number profile across the ramp.

The image display quality (test 6) involves the correct setting of the monitor controls for contrast and brightness. Once the proper settings have been determined and set, they should never be changed except to adjust for monitor aging: When the contrast and brightness of individual CT images need to be changed, only the window width and window level controls should be adjusted. As part of this test, the monitor should be observed for proper focus across the screen. Distortion can be determined by measuring the displayed height and width of a phantom image: they should be within 5% of each other. If not, the vertical or horizontal gain of the monitor should be adjusted. The screen should be also observed for any sign of flicker or jumpiness.

The monitor in the image documentation system can be inspected in the same way as the display monitor (test 7a). In addition, any dust on the monitor should be removed. The simplest way to check the overall performance of the monitor and camera is to reproduce an image of one of the uniform phantoms at all of the image positions on a film. This film can then be inspected at each of its image positions for proper focus, image density, image contrast and artifacts.

Finally the film processor should be monitored on a frequent basis, daily if possible (test 7b). This involves monitoring the base + fog, speed index, and contrast of test films exposed in a sensitometer. Since the single emulsion films used in the image documentation systems are more sensitive to processor variation than the films used in conventional radiography, processor QA becomes particularly important in CT. Also special care is needed if the same processer is used for both the single emulsion CT films and regular double emulsion film since the required replenishment rate may be quite different. The processor performance may vary significantly as the mix of films is varied. Finally, the images produced in the other tests should be inspected for any artifacts that may be due to the processor: uneven developing, drip marks, scratch marks, roller marks, etc.

DOSIMETRY TESTS

The dose figures quoted by the manufacuters for their CT scanners can be very confusing because of the many different ways to specify dose. There is the additional problem that a dose figure is rather meaningless unless the exact techniques of kVp, mAs, and slice thickness are specified at the same time. The following discussion will try to eliminate much of the confusion regarding CT dose by indicating the different ways in which dose can be specified, and the best ways to measure it in a QA program.

The first consideration in dose specification is the location of the dose measurement. The dose may be specified as "surface dose" or "central dose". The dosimeter may be placed on the surface of the phantom, at its center, or in fact at any other intermediate position. If the dosimeter is placed at the center of the phantom (or any place within it), the phantom size and composition must be specified. The size of the phantom will also affect the measurement of surface dose, but to a lesser extent (mostly by affecting the distance between the dosimeter and the x-ray source). Surface dose can vary greatly around the surface of the phantom if the scan angle is not exactly 360 degrees. This will occur with the normal 180 degree scans of 1st and 2nd generation scanners, and also with 3rd and 4th generation scanners operated in limited angle, fast scan modes, and in overscan modes (to reduce motion artifacts).

At whatever position the dose is measured, there exists a dose profile across the slice thickness irradiated, which is produced by the radiation from a single scan. This dose profile is a function of dose (rad or gray) vs. the the distance (mm) along the scan axis. Two particularly useful parameters of this dose profile are its maximum dose and dose profile width (which is ususally taken as the full width at half maximum of the dose profile). The dose profile at the center of the phantom will be much wider than that at the surface due to the presence of a large amount of scattered radiation. To measure the dose profile at the center of the phantom, measurements must be made over a distance 10 to 20 times the selected slice thickness and the phantom must of course be wider than this distance. Partly because of the technical difficulties of measuring the central dose profile, we recommended measurement of the surface dose profile for QA tests. An additional advantage of the surface dose profile is that it directly reflects the state of collimation of the x-ray source, while this effect is masked in the central dose profile by the overwhelming presence of scattered radiation.

The total amount of radiation absorbed (at a specified depth) by the patient during a CT scan of a single slice is proportional to the area under the single slice dose profile (the mathematical integral of the dose profile) measured at the specified depth. Thus this integral of the single slice dose profile is a very useful quantity. Its units are rad-cm. If this integral is divided by the selected slice width, the resultant quantity has been termed the CT Dose Index or CTDI. The CTDI, which has units of dose (rad or gray), can be specified at any point in the phantom from its surface to center. We find that the surface CTDI is a particularly useful quantity for CT QA tests that is also easy to measure.

A specification of the surface dose alone is not the best way to evaluate the total amount of radiation absorbed by the patient, and thus his rediation risk. Different models of CT scanners with similar surface doses may produce significantly different central doses and therefore different radiation risks to the patient. However, a measurement of surface dose is a very good way to monitor any changes in the performance of a specific CT scanner, which is the precise purpose of the QA program.

Since all actual clinical CT studies involve multiple slices, the dose or dose profile is sometimes specified for a multiple slice series of scans. Normally the multiple slice dose profile is specified for each selectable slice width with the condition that the slice incrementation distance is equal to the slice width (contiguous slice condition). Two parameters of this profile are the maximum dose and the average dose. The average dose for contiguous multiple slices is clearly a particularly useful quantity. An important relationship that we note here is that the average dose for contiguous multiple slices is equal to the CTDI.

CT scanner dosimetry can be performed using TLD's or an ion chamber. A series of TLD's can be stacked together such that a dose profile can be measured. An ion chamber can be used to measure the integral of the dose profile and thus the CTDI. When the proper equipment is available, ion chamber measurements are simpler than those using TLD's, and can be useful when the exact dose profile is not needed. For CT dosimetry the ion chamber should be cylindrically shaped (pencil chamber), and its active length should be significantly larger than the bulk of the dose profile. For surface dose measurements of a single slice, a 4 cm active length is sufficient; while for central dose measurements, the chamber may need to be 10 to 15 cm long. To derive the integral of the dose profile (IDP) or the CTDI from the ion chamber measurement, use the following equations:

$$IDP\ (rad\text{-}cm) = 0.94 \times E \times A \qquad (2)$$

$$CTDI\ (rad) = 0.94 \times E \times \frac{A}{W} \qquad (3)$$

Here, E is the ion chamber reading (in Roentgen units); the .94 factor converts the ion chamber measurement of "exposure" in units of roentgen (R) into "dose" in units of rad; W is the selected slice width (in cm); A is the active length of the ion chamber (in cm).

If TLD's are not used and an idea of the width of the slice profile is still desired, film dosimetry can be used. In this case a sheet of film is placed over the surface of the phantom and a single exposure is made at each selectable slice thickness. The phantom is moved between each exposure so that the exposures do not overlap on the film. The film must not be overexposed, otherwise the apparent slice thickness will be too large - an optical density of about 1.0 to 2.0 is about right. Thus a suitable film and x-ray technique must be used.

We consider it important to do a very thorough check of dosimetry during the acceptance testing of a CT scanner. This should include the use of TLD's to obtain the single slice dose profiles for all selectable slice thicknesses. Subsequent QA tests may continue to use TLD's to test dosimetry. However it is also acceptable to use a suitable ion chamber placed on the surface of the phantom to measure the surface CTDI for comparison to previous values. If either the tube output or the width of the dose profile should increase between tests, it would be detected as an increase in the CTDI. Additional film dosimetry, as described above is then a useful adjunct to confirm the width of the dose profile.

It is very useful to compare the dose profile measured at the phantom surface with the slice sensitivity profile obtained from the CT image. When doing this comparison it is important to mathematically transform the dose profile from the position of the scanner surface to the scanner isocenter. This is done by multiplying the distance measurements in the dose profile by the fraction: D / (D - r) , where D is the distance between the focal spot of the x-ray tube and the scanner isocenter, and r is the radius of the phantom. This assumes the phantom is centered in the scanner aperture. The result of this transformation will be a slight increase in the width of the dose profile. Ideally, this transformed dose profile should closely match the slice sensitivity profile. Also, both the measured width of the slice sensitivity profile and the width of the dose profile should be close to the selected slice width.

PERFORMANCE OF NON-IMAGE COMPONENTS

Here we are including essential tests that do not fall within the categories of image quality or dosimetry testing. These tests include evaluations of the accuracy of the patient positioning devices:

1. Patient positioning using the digital scout images available with many current scanners. This test can be performed as part of the positioning of the test phantom in preparation for other tests.
2. The patient positioning lights that mark the location of the scan plane.
3. Manual and computer controlled motion of the patient table. Incremental changes in position should be accurate and reproducible in both directions.

TECHNIQUES FOR DAILY QUALITY ASSURANCE

Daily quality assurance is very important to maintain consistant image quality. The tests that we recommend can generally be done quite quickly using the uniform water phantom supplied by the CT scanner manufacturer. These tests include the following image quality tests that have been previously discussed: (1a) the CT number accuracy of water; (2a) artifacts present in the image of uniform phantom, such as ring artifacts; (3a) image noise - obtained from the image of uniform phantom; (7b) performance of the film processer.

These tests require the taking of a single scan at each of 2 to 4 of the most commonly used scan modes. Each image is then evaluated using the CT scanner's built in statistical analysis software: The average CT number and the standard deviation or pixel noise is determined for a central region in each image. These values are tabulated in a log where daily deviations are easily spotted. The most significant potential problem is an increase in the noise, since this will affect the low contrast detectablity of the scanner. These images are then viewed for the presence of artifacts. Finally, as mentioned previously, the processor is an essential link in the image quality of the final image and should not be neglected. Daily checks of base + fog, speed index, and contrast are recommended.

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