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College on Medical Physics
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Quality Assurance for Your CT Scanner

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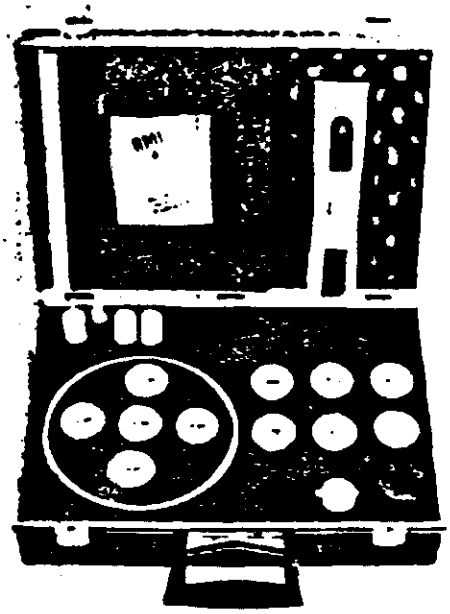
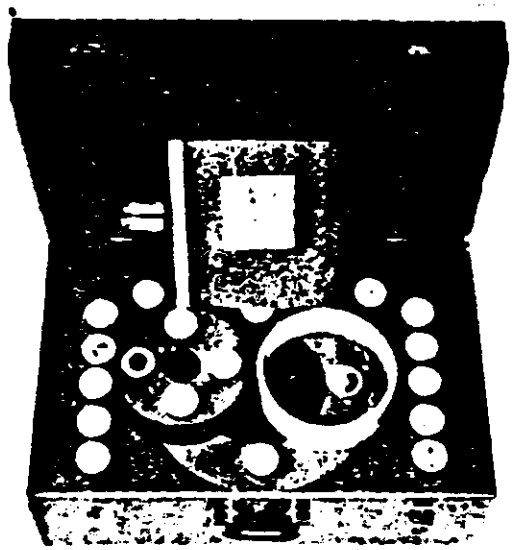
QUALITY ASSURANCE

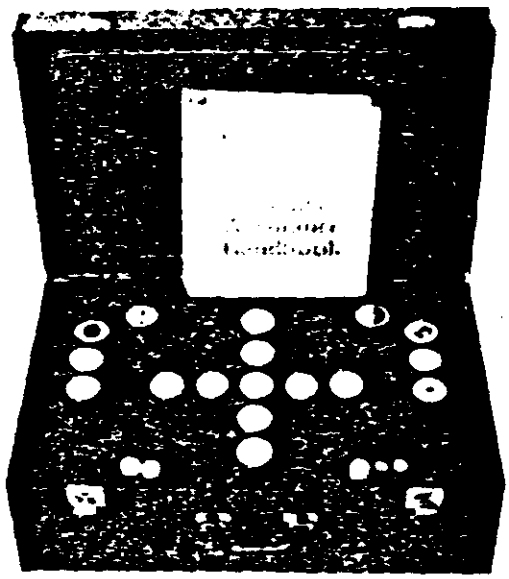
FOR YOUR CT SCANNER



Head Phantom
Model 460A



Combined Head/Body Phantom
Model 461A



Body Phantom
Model 462A

INTRODUCTION:

This Quality Assurance (QA) Manual describes a series of simple, effective tests using RMI's CT Phantom Kits to check CT scanner performance. These phantoms are designed for a complete CT quality assurance program, including daily quality control tests and periodic, comprehensive QC testing. Standard test inserts make possible basic QC tests, while optional inserts allow for more sophisticated tests that may be useful for physicists and engineers. These inserts are designed for maximum

flexibility with a minimum amount of time, effort, and cost. The RMI CT Phantom Kits have marked advantages over water bath phantoms in convenience and performance. Custom tests or new developments can be easily accomodated into the phantom.

STANDARD INSERTS

Part Number		Measures
460-010	- Plain (5, or 9 with Body and Head/Body Phantom)	Noise, CT Number, and Spatial Uniformity
460-011	- Edge/Contrast Scale (1)	Edge/Contrast Scale Response
460-012B	- High Contrast Resolution (1)	Spatial Resolution (1.50 to 0.4 mm at 100% contrast)
460-013A	- Low Contrast Resolution (1)	Low Contrast Detectability(0.6%)
460-014	- Alignment Pin (1)	Alignment Artifact
460-015	- Helix (4) U.S. Patent No. 4,344,183	Alignment, Slice Thickness, and Phantom Position
460-017A	- Aluminum Ramp (2)	Slice Thickness and Sensitivity Profile (26.6 degree slope)
460-020H	- Compact "Bone" Rod (2)	Beam Hardening Artifact
460-020	- Linearity (2) with Acrylic (1), Teflon (1), and Polyethylene (1) Rods	Linearity of CT Numbers

OPTIONAL TEST INSERTS

Part Number		Measures
460-012C	- High Contrast Resolution	Spatial Resolution (2.0 to 0.75 mm at 100% contrast)
460-013B	- Low Contrast Resolution	Low Contrast Detectability (0.3%)
460-018	- TLD Holder	TLD Dose Dose Profile
460-019	- Dose	Integral Dose (14.5 mm I.D.; use with TLD Holder, Pen Dosimeter, or Ion Chamber)
460-020D	- Clear Resin Rod	
460-020E	- Nylon Rod	
460-020F	- Lexan Rod	
460-020G	- Polypropylene Rod	
460-020I	- Polystyrene (clear) Rod	
460-021	- Impulse Response	Impulse Response
460-022	- Wedge	Modulation Transfer Function(MTF)
460-023	- keV Liquid	keV Effective and Linearity (29 mm, I.D.)
460-023A	- Tube for keV Liquids	29 mm Diameter Angled Tube

The Head Phantom Model 460A, consists of a 19 cm disc of "Solid Water" plastic with removable 4 mm layer of "compact bone." The combined Head/Body Phantom, Model 461A, has a head module with a slip on and off bone ring and body annulus. The Body Phantom, Model 462A, is 33 cm in diameter. The standard inserts, a carrying case, and instruction manual are included with each phantom kit.

PREFACE

Since its clinical introduction in 1972, the CT scanner has proven to be an invaluable tool in diagnostic radiology. The high level of electronic and mechanical complexity inherent in a CT scanner, as compared to the more common diagnostic instruments, make the implementation of a CT Quality Assurance (QA) program essential to the maintenance of good image quality and low patient dose. Experience has shown* that careful quality control (QC) measurements are needed to monitor scanner performance and to detect any degradation of image quality before significant diagnostic information is lost. Water-filled phantoms and phantoms supplied by CT machine manufacturers often have limited applications or are cumbersome and time consuming to use. The RMI CT Phantom kits make possible convenient, accurate, and comprehensive QC testing.

Every CT scanner system should be incorporated into a comprehensive QA program consisting of initial planning, QC measurements, careful record keeping, timely evaluation of test results, and follow-through on corrective action. Management of each CT scanner should include maintaining a notebook, or manual, which includes records on QC testing schedules and results, preventive maintenance and repair, equipment modifications or updating, and other pertinent information affecting the operation of the scanner.

The average CT number and standard deviation (σ) of regions of interest in several positions in the water equivalent phantom should be checked daily to monitor noise properties and CT number accuracy (spatial uniformity-temporal stability) of the scanner. Contrast scale - CT number linearity, high contrast resolution, low contrast detectability, alignment artifact, slice thickness, and localizing light accuracy should be checked weekly or bi-weekly. In addition, the image display and documentation systems should also be checked for correct contrast, brightness, focus, and to determine the presence of any image distortion, flicker, or jumpiness. Of course, processor QC is also an important consideration for maintaining the image documentation quality. Complete testing of all scanning modes in clinical use, including dosimetry tests, should be done about every three months. Frequent testing permits you to detect gradual changes in scanner performance, and to become aware of image degradation, before diagnostic performance is significantly affected.

*White, D.R., R.D. Speller, and P.M. Taylor, "Evaluating Performance Characteristics in Computerized Tomography", British Journal of Radiology, 54, 221-231 (March 1981).

Speller, R.D., D.R. White, et al., "A Survey of 29 EMI CT Machines in Britain", British Journal of Radiology, 54, 232-240 (March 1981).

Speller, R.D., D.R. White, et al., "An Evaluation of CT Systems from Ten Manufacturers", British Journal of Radiology, 54, 1053-1061 (December 1981).

TABLE OF CONTENTS

RMI extends special thanks to Frank Ranallo, Ph.D., University of Wisconsin and Keith Nelson, Product Development Manager, RMI for their work in producing this manual.

PHANTOM DESCRIPTION*

ıe RMI CT Head Phantom (Model 460A) is a disc of water equivalent plastic cm thick and 19 cm in diameter. The disc is surrounded by a cylinder of .mulated skull bone 4 mm thick. The CT Body Phantom (Model 462A) is a ıter equivalent disc 6 cm thick and 33 cm in diameter. The Head/Body ıantom (Model 461A) consists of three main parts, all 6 cm thick. It ıcludes a water equivalent disc 19 cm in diameter with a removable bone ıuivalent ring 4 mm thick, plus a water equivalent body annulus of 33 cm ıter diameter. This model has the advantage of being usable for both ıad and body scanning modes, with minimum weight and bulk. RMI's water ıuivalent CT phantom kits (Models 460A, 461A, and 462A) are compatible in ısign and have similar features which are described in the following ıragraphs.

ırious tapered inserts, of uniform water-equivalent composition or with ·-·. objects in them, can be inserted into tapered cavities molded into ı. discs. The conical inserts, which are easy to insert and remove, are 7 n high and 4.5 cm in diameter at the position of the scan slice. Once a ıantom is properly positioned with respect to the CT x-ray beam, a testing ıogram can be accomplished simply by interchanging the test inserts. The ıantom itself need not be moved. This greatly speeds the testing process.

ıe phantoms are made from a water equivalent plastic (Solid Water) which ıs the same x-ray properties as liquid water in the energy range of 10 keV ı 100 MeV. The simulated skull bone is a bone equivalent plastic ıterial. Properties of various tissue equivalent materials are given by ıite (Ref. 41, 42). The gel provided with the phantom is a water-ıuivalent gel that can be applied to the outside of the inserts before lacing them in the phantom cavities. It will reduce the effect of the nall amount of trapped air that may cause a low density ring in the image t the insert-cavity interface. As this low density ring will not affect ıcuracy for most of the usual tests, use of the gel need not be routine. ıe magnitude of this low density ring can be sufficiently reduced by ıshing the inserts firmly into the cavities with a slight twisting motion. case where the use of the gel is recommended is when the average CT ımber of the entire phantom area (including insert-cavity interfaces) is alculated. If CT number measurements are made away from insert cavity nterfaces, they will not be affected by a low density ring.

vailable tests using the RMI CT Phantoms with standard and optional nserts are listed on page 2. Each test is described in the remainder of hese instructions and a bibliography is included at the end for further nformation.

The formulation of tissue and water substitute plastics is due to Dr. avid R. White and Dr. Chris Constantinou of St. Bartholomew's Hospital, ondon, England and the conception of the phantoms is due to the ollaborative work of Dr. White and Dr. Robert D. Speller of Middlesex ospital Medical School, London, England.

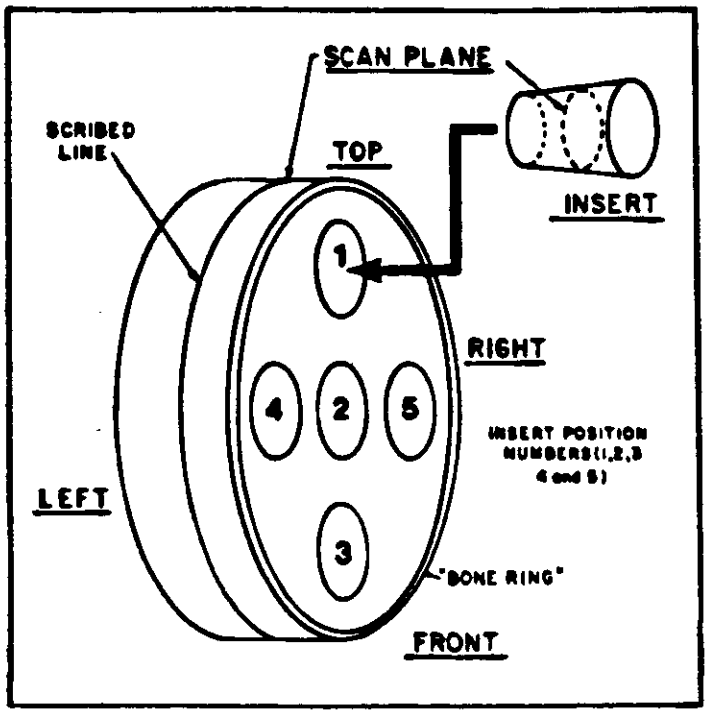


Fig. 4. Head Phantom configuration.

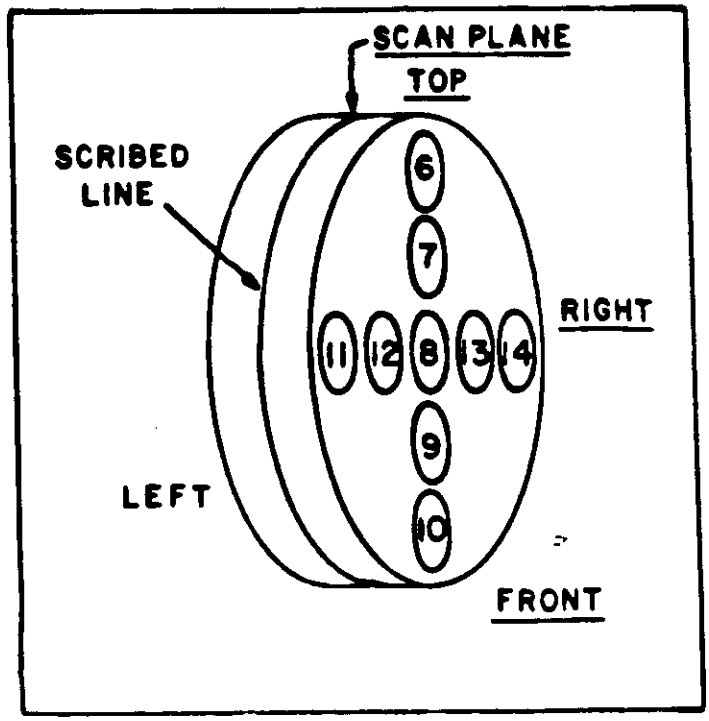


Fig. 5. Body Phantom Configuration

DEFINING "CT NUMBER"

An important note on the meaning of CT numbers:

Various systems of CT numbers have been used in the past, so to avoid confusion a quick review will be presented here. For all CT scanners, the CT number of some point in the image is ideally an expression of the x-ray linear attenuation coefficient (μ) possessed by that point in the subject being scanned. The two most common systems reference the linear attenuation coefficient to that of water ($\mu(H2O) \approx .19$ cm(-1)), so that water has a CT number of zero. The formulas for relating CT number (#) to μ for these two systems can be expressed in the following way:

System 1: CT # of substance x $= \dfrac{\mu(x) - \mu(H2O)}{\mu(H2O)} \times 500$

CT # of Air	=	-500
CT # of Water	=	0
CT # of Bone	=	+500

A change in $\mu(x)$ equal to 1% of $\mu(H2O)$ is equivalent to a change in CT number of 5 units.

System 2: CT # of substance x = $\frac{\mu(x) - \mu(H2O)}{\mu(H2O)}$ x 1000

CT # of Air = -1000

CT # of Water = 0

CT # of Bone = +1000 [μ(bone) $\approx$ 2μ(H2O)]

A change of μ(x) equal to 1% of μ(H2O) is equivalent to a change in CT number of 10 units.

The formula for expressing CT#'s can be written in a more general fashion for both System 1 and System 2:

CT# of substance x = $\frac{\mu(x) - \mu(H2O)}{\mu(H2O)}$ x [CT Scale Factor]

Where [CT Scale Factor], nominally, is 500 for System 1 and 1000 for System 2. This definition of CT scale factor will be used throughout the instructions.

NOTE: Though the nominal, stated, value for the "CT Scale Factor" may be 500 or 1000, the actual value may deviate somewhat from these numbers. The actual "CT Scale Factor" of a particular scanner can be determined during QC testing (See Test 5, P. 20).

TEST DESCRIPTIONS

1. LOCALIZING LIGHT ACCURACY (POSITIONING THE PHANTOM)

In addition to determining the accuracy of the patient positioning system, this test procedure also properly positions the phantom for subsequent tests. For all tests 2 through 15, the phantom is normally positioned such that the scribed line, on its circumference, is in the center of the scan plane. (See Fig. 6 for the geometry of the phantom and inserts relative to this scribed line.) The phantom can be taped or strapped to the patient table or, if it is a head phantom, secured to the head holder (Fig. 7). Optionally it can be fitted to a custom mounting bracket utilizing the threaded holes in the back of the phantom or RMI's CT Phantom Holder (Model 46-01 can be used for GE/Siemens scanners . The level supplied with the phantom should be useful in eliminating tilt in initially orienting the phantom and rechecking it as needed.

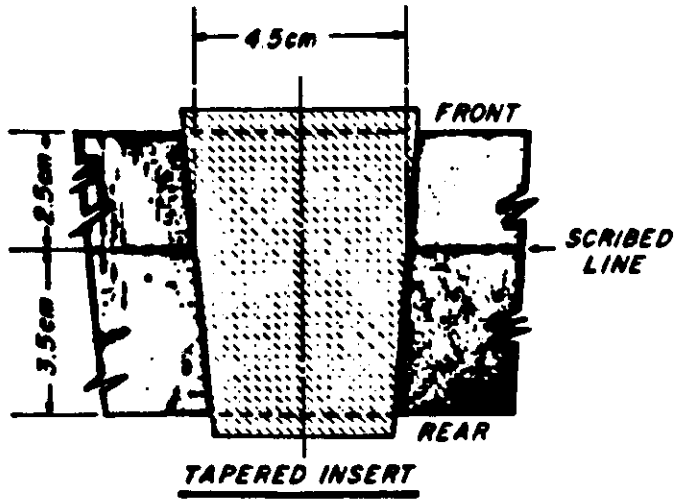


Fig. 6. Schematic diagram of insert position in the CT Phantom.

The position and angulation of the phantom with respect to the scan plane can be determined by using the four HELIX Inserts (Part #460-015)*. A HELIX Insert (Fig. 8) contains a radiopaque helix with a pitch of 2 cm per 360 degrees. The point on the helix that is in the plane of the phantom's exterior scribed line is marked with a 1/16" steel ball. The four HELIX Inserts should be placed in the four outermost phantom positions, inserted firmly with a slight twisting motion to assure proper seating of the insert (Fig. 7). The word "HELIX" should be horizontal with the position of the steel ball (as indicated on the label) on the right side. The CT image of the HELIX Insert will show the radiopaque helix as the arc of a circle (Fig. 9). The fraction of the circle that is imaged is determined by the slice width: 18 degree of arc is displayed for each mm of slice width. (Examples: A 180 degree displayed arc, or 1/2 of a full circle, indicates a 10 mm width; a 90 degree arc, or 1/4 of a full circle, indicates a 5 mm slice width.)

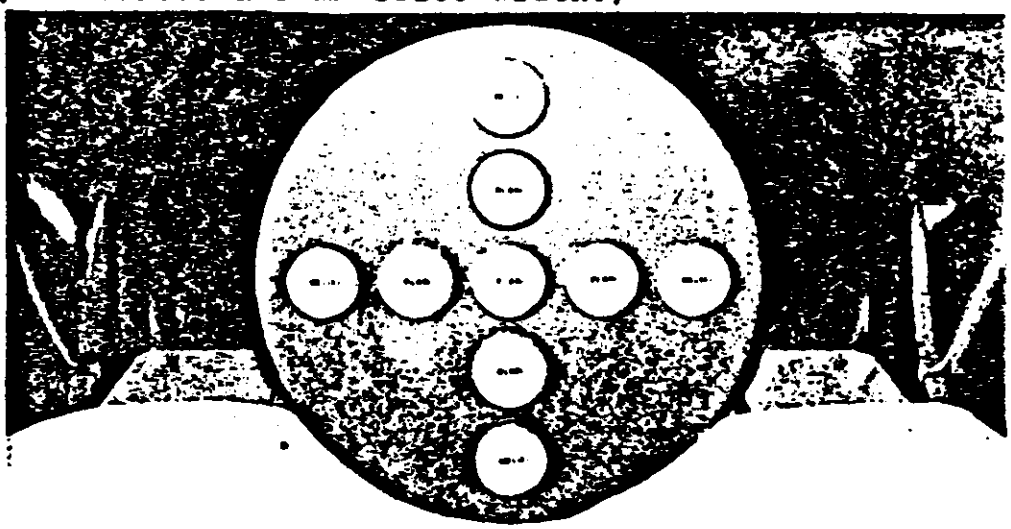


Fig. 7. Placement of CT Body Phantom on patient couch.

U.S. Patent No. 4,344,183

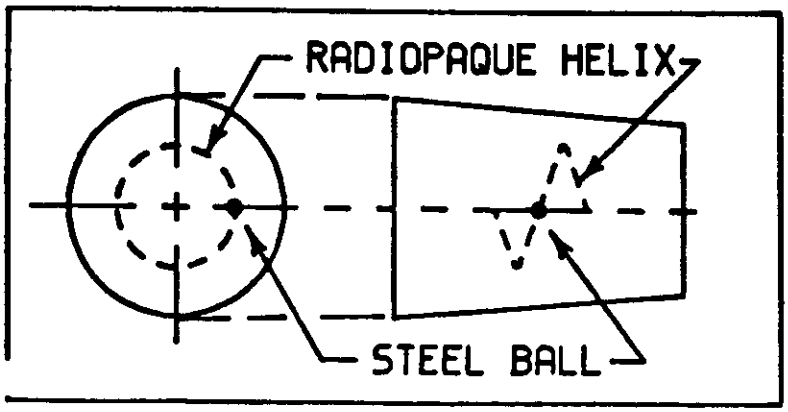


Fig. 8. Schematic diagram of HELIX Insert.

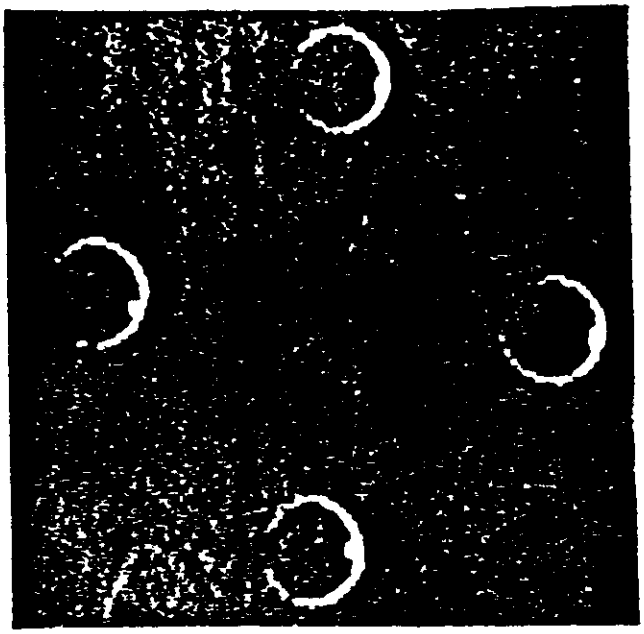


Fig. 9. Scan of Head Phantom with HELIX Inserts in the four outside positions.

When the phantom is properly positioned with the scribed line in the center of the scan plane, the steel ball will be in the center of the scan slice width and will be imaged at the midpoint of the displayed arc for all four inserts. If the images of the steel balls are not centered on the displayed arcs this indicates that the phantom is not properly positioned to the scan plane. The procedure for determining the amount of positioning error and correcting these errors is described below:

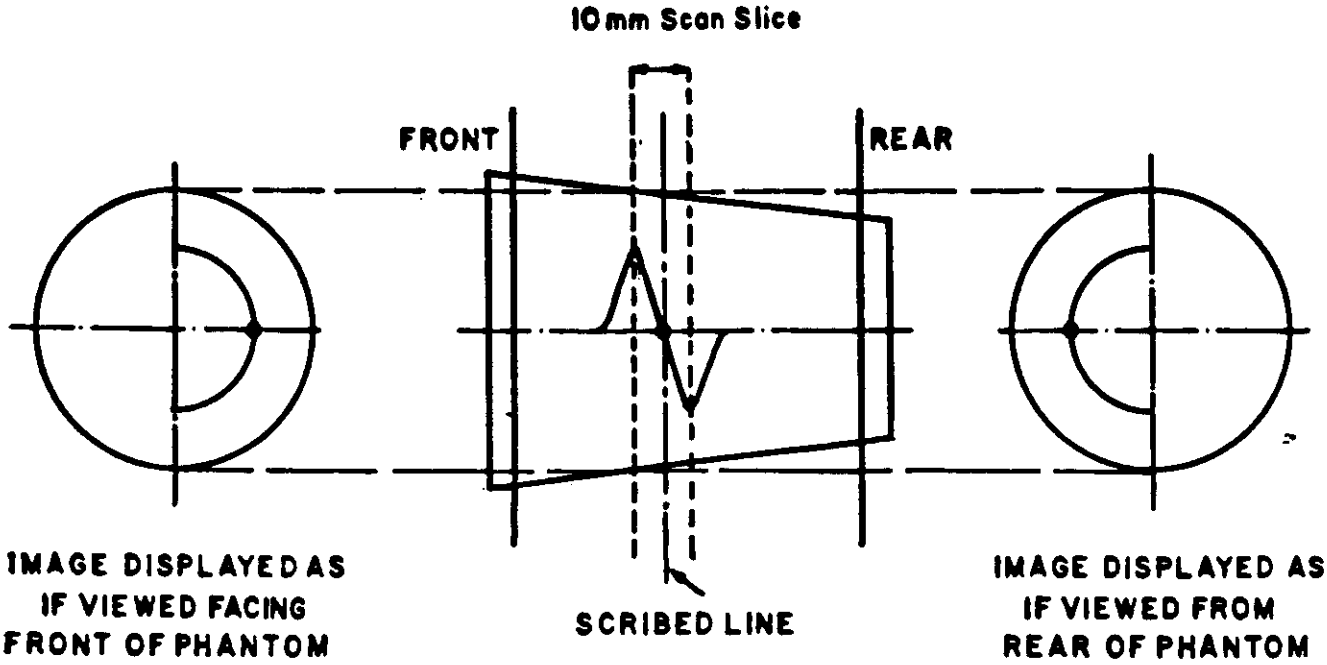
1) The angular error is defined as the angle between the center of the displayed helix arc and the position of the steel ball displayed on the arc (or the position the ball would have if it were displayed). Visually estimate or use a protractor (transparent plastic polar graph) to determine the angular error for each helix.

2) Divide the estimated angular error by 18 degrees to determine the error of phantom position in mm. This displacement should be determined at each helix position.

3) If the displayed arc must move clockwise for proper centering of the image of the steel ball, the phantom must be moved out of the gantry to correct the positioning (Fig. 10B).

4) If the displayed arc must move counter-clockwise for proper centering of the image of the steel ball, the phantom must be moved into the gantry to correct the positioning (Fig. 10C & D).

Fig. 10. Relationship of the displayed arc and the steel ball image for various slice widths and positions of the phantom.



10A. 1 cm slice width properly centered. Notice that the steel ball is in the center of the 180 degree displayed arc.

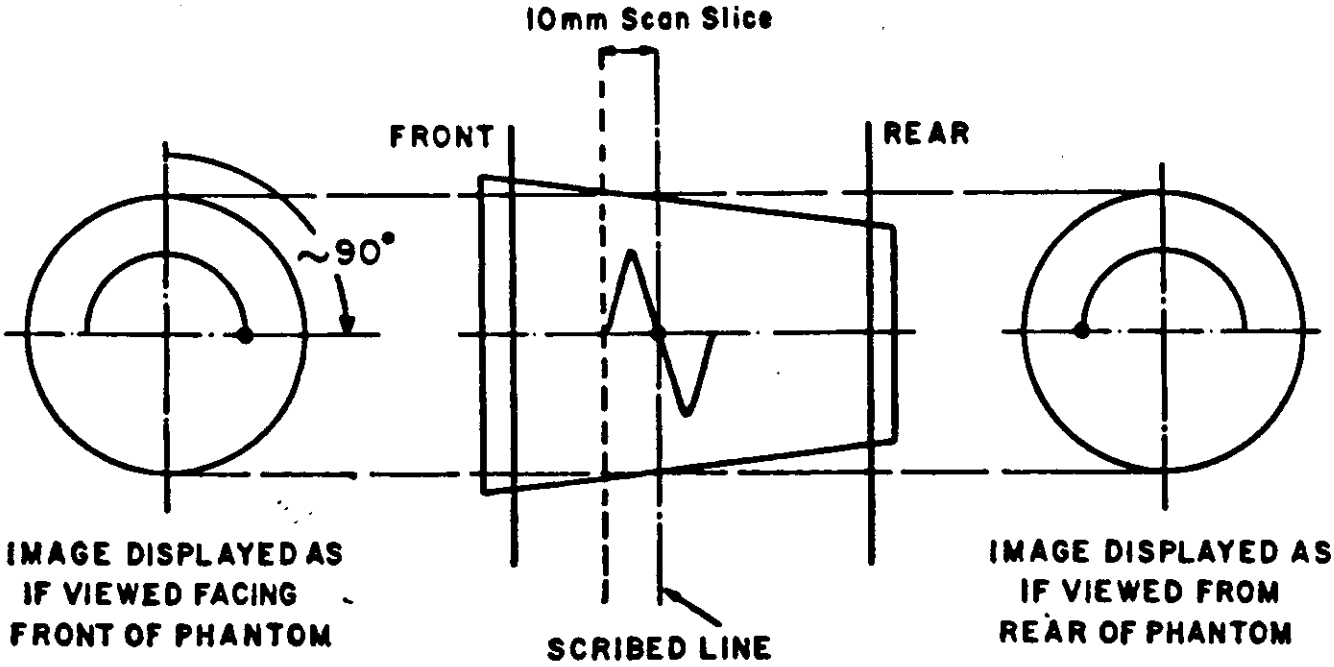
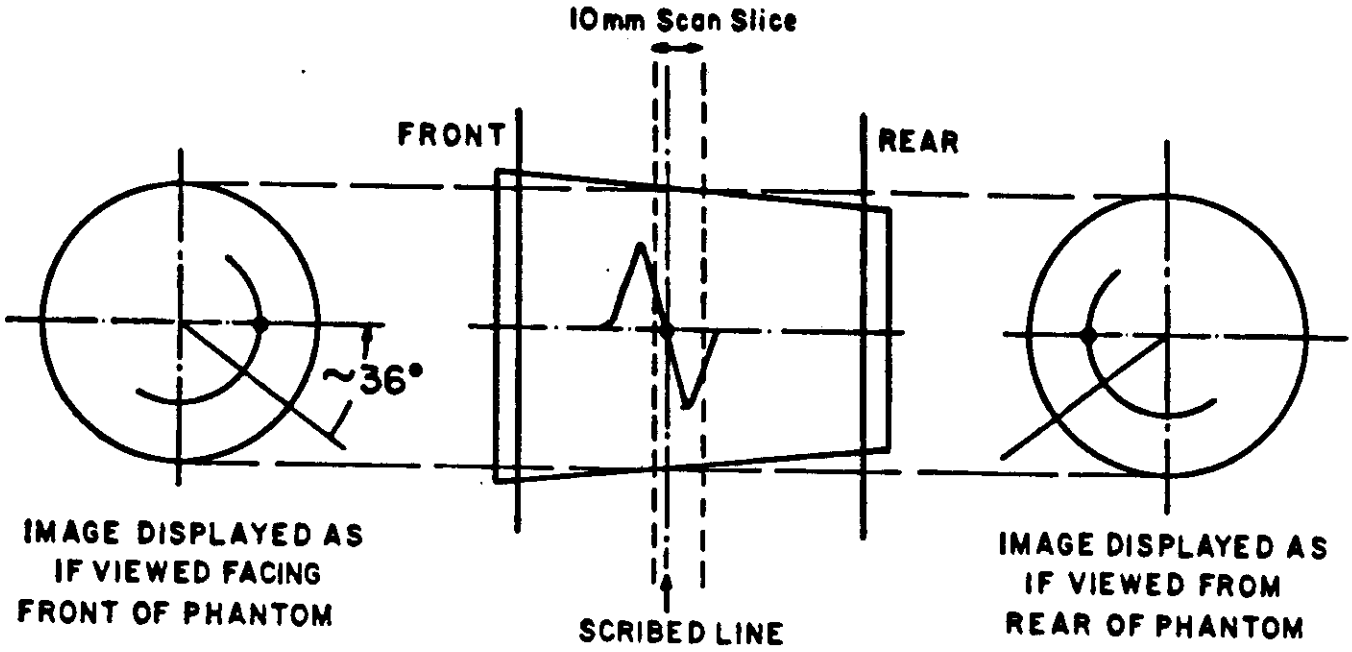
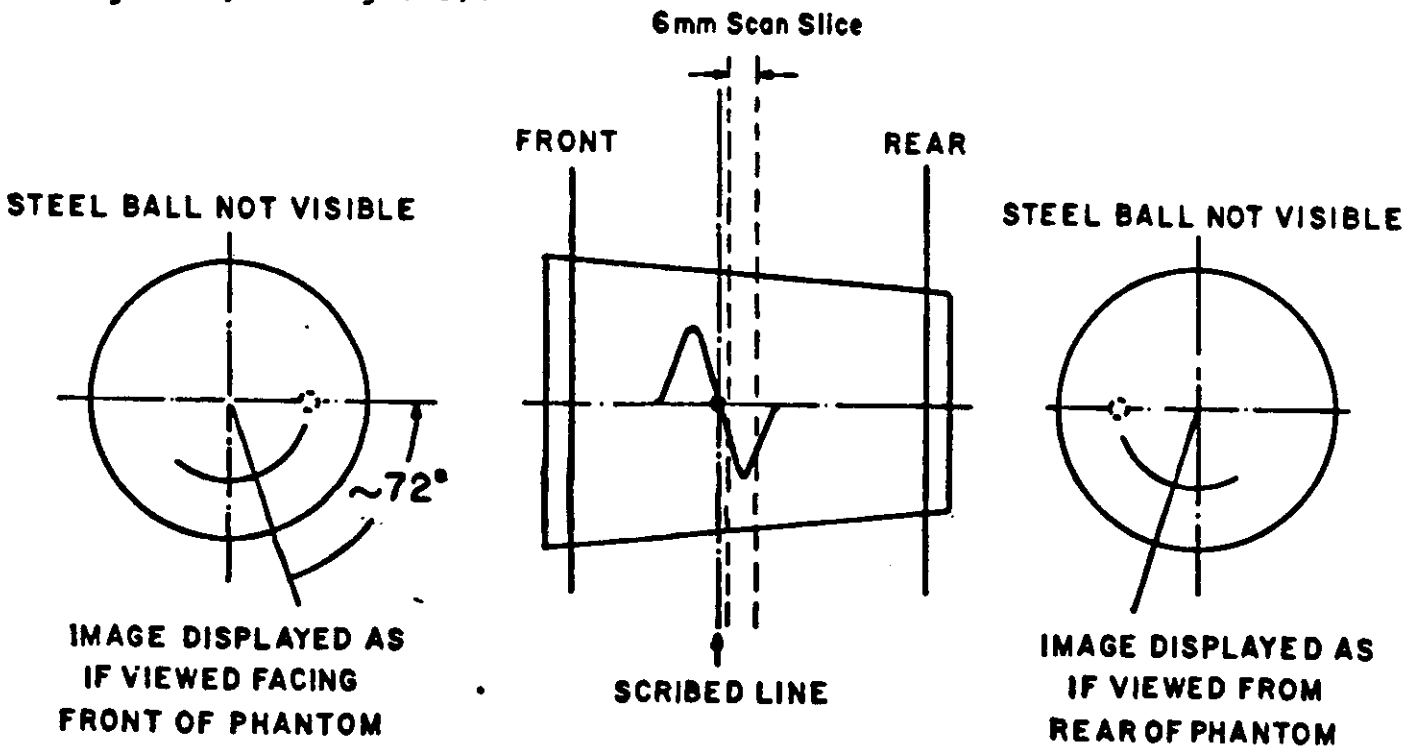


Fig. 10B. Improper position of the phantom for a 1 cm slice width. For an image displayed as if viewed facing the front of the phantom, the arc must be moved 90 degrees clockwise. Thus the phantom must be moved out of the gantry 5 mm (90 degrees ÷ 18 de grees).



10C. Improper position of the phantom for a 1 cm slice width. For an image displayed as if viewed facing the front of the phantom, the arc must be moved 36 degrees counter-clockwise. Thus the phantom must be moved into the gantry about 2 mm (36 degrees ÷ 18 degrees).



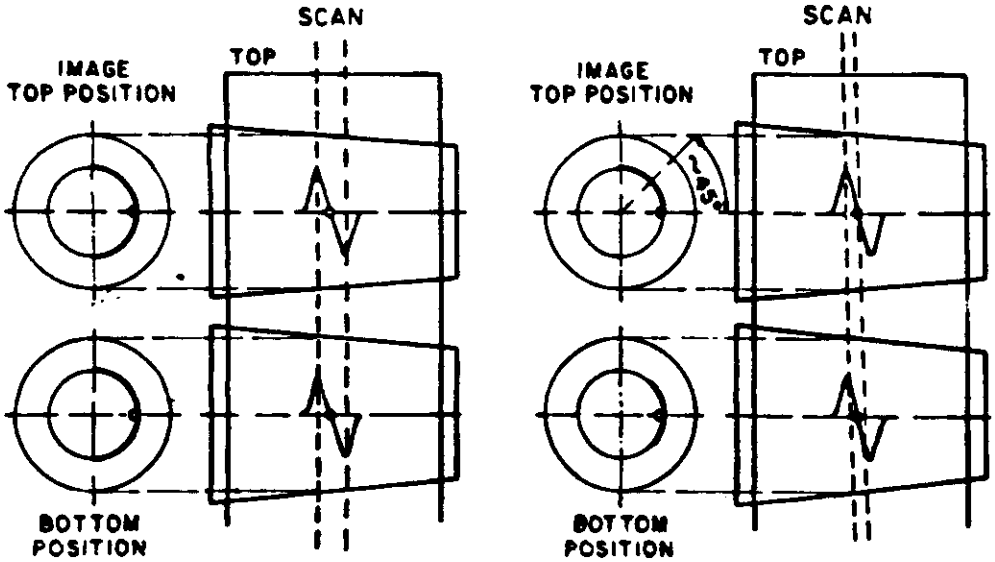
10D. Improper position of the phantom for a 6 mm slice width. For an image displayed as if viewed facing the front of the phantom, the arc must move 72 degrees counterclockwise. Thus the phantom must be moved about 4 mm into the gantry (72 degrees ÷ 18 degrees).

To avoid confusion "into the gantry" and "out of the gantry" must be properly defined. Imagine positioning yourself in the scanning room and looking at the phantom so that you see it in the same right-left orientation as the image presented on the viewing screen. If on the viewing screen the images of the steel balls are on the right side of the displayed arcs then you would be positioned facing the front of the phantom and could read the insert labels. If the images of the steel balls are on the left side of the displayed arcs, then you would be positioned facing the rear of the phantom and could not see the insert labels. Having positioned yourself in this way, "into the gantry" means "away from you"; "out of the gantry" means "towards you".

Fig. 10 shows four simulated images of a single helix together with schematic representations of the scan slice relative to the helix. Not that in Fig. 10D the image of the steel ball does not appear, but its position is known if the insert is properly oriented. In general, the steel ball image will not appear if the position error is more than half of the slice width.

Fig. 11 shows how the use of two HELIX Inserts in a vertical plane permits the vertical tilt of the phantom to be determined. A similar comparison using two helix inserts in the horizontal plane can be used to determine the horizontal tilt. NOTE: Vertical tilts can usually be avoided by properly leveling the top surface of the phantom using the supplied level.

Fig. 11. Displayed arc and steel ball images for two different slice widths and positions of the phantom.



11A. 1 cm slice width properly positioned in the top and bottom positions.

11B. 5 mm slice width properly positioned at the bottom, but the top must move 2.5 mm out of the gantry (45 degree ± 18 degrees). This indicates a tilt from the vertical.

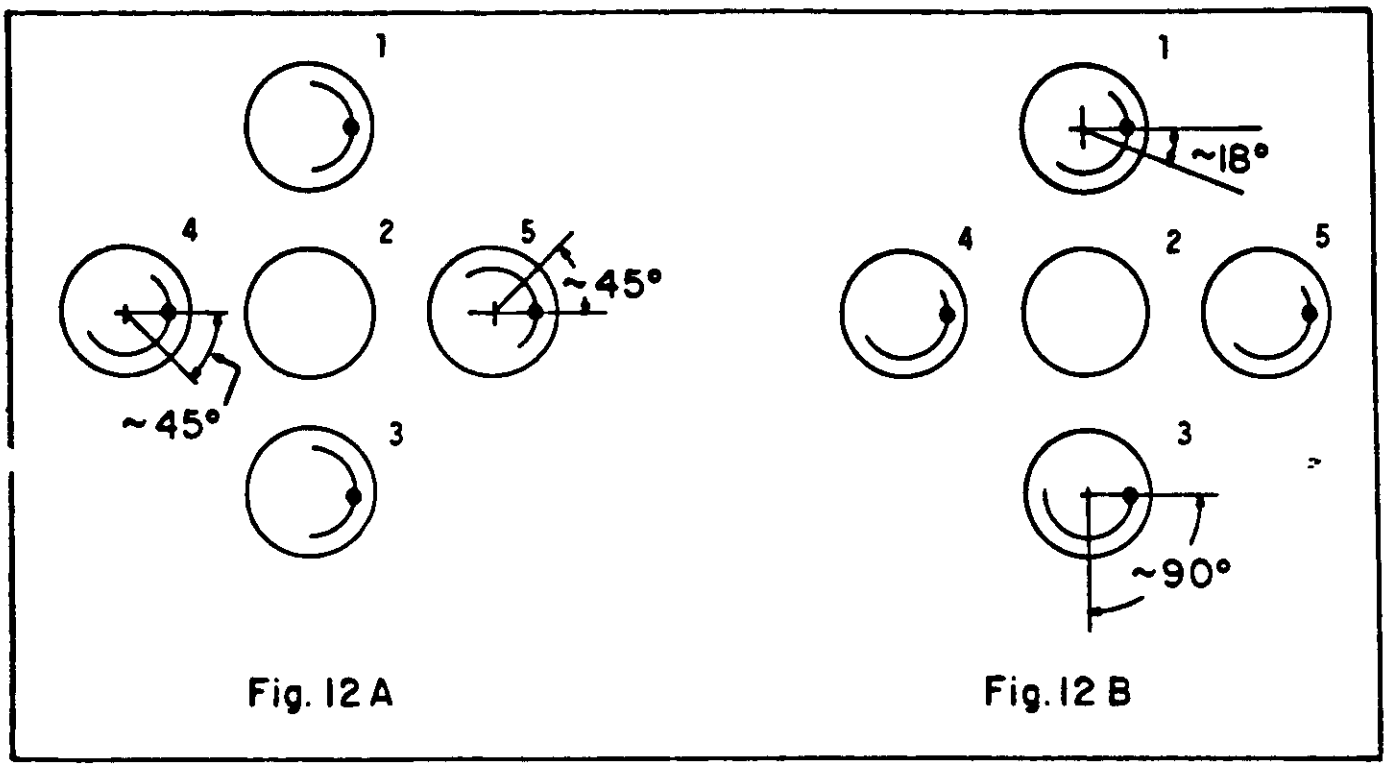


Fig. 12. Illustrations of angulation and correction.

By using the four HELIX Inserts the tilt of the phantom with respect to the scan plane can be accurately determined both horizontally and vertically.

Fig. 12A shows a phantom with horizontal tilt as indicated by the uncentered arcs displayed by the helicies in positions 4 and 5. The angular error for each of these two helicies is estimated at 45 degrees. Therefore, 45 degrees divided by 18 degrees indicates a 2.5 mm positioning error. To properly center the phantom, the left side must move 2.5 mm into the gantry while the right side moves 2.5 mm out of the gantry.

Fig. 12B shows a phantom image where none of the steel balls are in the scan plane. There is an angular error of 18 degrees at position 1; in position 3 the error is 90 degrees. The top of the phantom must be moved into the gantry 1 mm (18 degrees ÷ 18 degrees), while the bottom must come into the gantry 5 mm (90 degrees ÷ 18 degrees).

NOTE: The tests which are most sensitive to errors caused by phantom tilt are 6 - High Contrast Spatial Resolution; 7 - Edge Response - MTF: 11 - Impulse Response (Point Spread Function).

To perform Test 1, Localizing Light Accuracy, the scribed line on the top of the phantom should be visible and not obscured by tape, straps, or bolusing. If the CT scanner being tested employs a localizing light outside of the scan plane, then positioning accuracy is dependent on both the localizing light accuracy and accurate incrementation of the patient table. With such a scanner, secure the phantom to the patient table or head holder and place the four HELIX Inserts in the outermost phantom cavities, inserting them firmly with a slight twisting motion. Level the phantom's top surface, and move the patient table so that the localizing light falls over the scribed line on the phantom. Then move the phantom into the CT scanner gantry the prescribed distance for proper positioning. Scan the phantom and note the error in positioning indicated by the HELIX insert in the top position. This is the error in the patient positioning system. For scanners utilizing a localizing light directly in the scan plane the same procedure can be followed except, of course, the phantom is not moved into the gantry after the localizing light is positioned over the scribed line. A better method for this kind of scanner is to reposition the phantom after the initial scan, trying to correctly position it in the scan plane as previously described. When a scan of the HELIX Inserts indicate that the phantom has been correctly positioned, simply turn on the localizing light and measure the distance between the light's mark on the phantom and the scribed line. This will yield the localizer's error. The phantom is now properly positioned for any further tests. If these additional tests are to be performed with a bolus around the phantom, it is best to do Test 1 with the bolus on the phantom, carefully peeling back the bolus at the top of the phantom to see the scribed lines.

So far only positioning of the phantom with respect to the scan plane has been discussed, but for all tests it is also important that the phantom be properly centered within the reconstruction circle. This is easily checked in the scan image and the patient table can then be moved up or down to correct errors. If the outside limit of the reconstruction circle is not visible in a scan image it can usually be made to appear by adjusting the CT window level to a lower value (negative values). Lack of centering may result in a shading of the image of the phantom yielding a different apparent density for the top and bottom section of the image (Fig. 14). While mild shading will not affect Test 1, it may significantly alter the results of other tests.

2. IMAGE SLICE WIDTH

This test utilizes the HELIX Inserts. Their proper use has been described in the previous section on Test 1. To obtain the most accurate results, phantom tilt should be checked using the four HELIX Inserts and minimized. The slice width determined from the Helix Insert image is the fraction of a complete circle that is imaged times 20mm. Equivalently, 18 degrees of helix arc is

displayed for each mm of slice width. This arc can be visually estimated or measured with a protractor. A transparent plastic polar graph obtainable from book stores with engineering supplies makes a good protractor, or use the one supplied with the CT kit.

For the most accurate measurements of slice width the use of the Aluminum Ramp Insert is recommended, especially for small slice widths. Its use is described in Test 10, Slice Sensitivity Profile. It is especially simple to use with scanners which have cursors with the capability to measure distances on the image.

3. SPATIAL UNIFORMITY AND TEMPORAL STABILITY (CT NUMBER ACCURACY OF WATER)

The spatial uniformity of a CT image and the day to day consistancy of CT numbers and noise values obtained from a homogeneous phantom are simple, effective indicators of overall CT system performance. RMI's CT phantoms have the degree of homogeneity required for these tests.

These tests are performed after the phantom has been properly positioned and centered. The PLAIN Inserts (Part #460-010), which contain no test objects, are inserted into the phantom at all insert positions.

These tests should be performed at all the techniques (kVp, mA, times) and scan sizes employed clinically by the scanner. Good performance at one technique and scan size is no guarantee of good performance at any other technique and scan size.

Ideally when a uniform phantom of water or water equivalent material is scanned the resultant phantom image should everywhere demonstrate a CT number of zero. Of course, even in the best of actual images all the image pixels would not be zero because of image noise; still the average CT number over any significantly large region should be close to zero no matter where in the image that region is located.

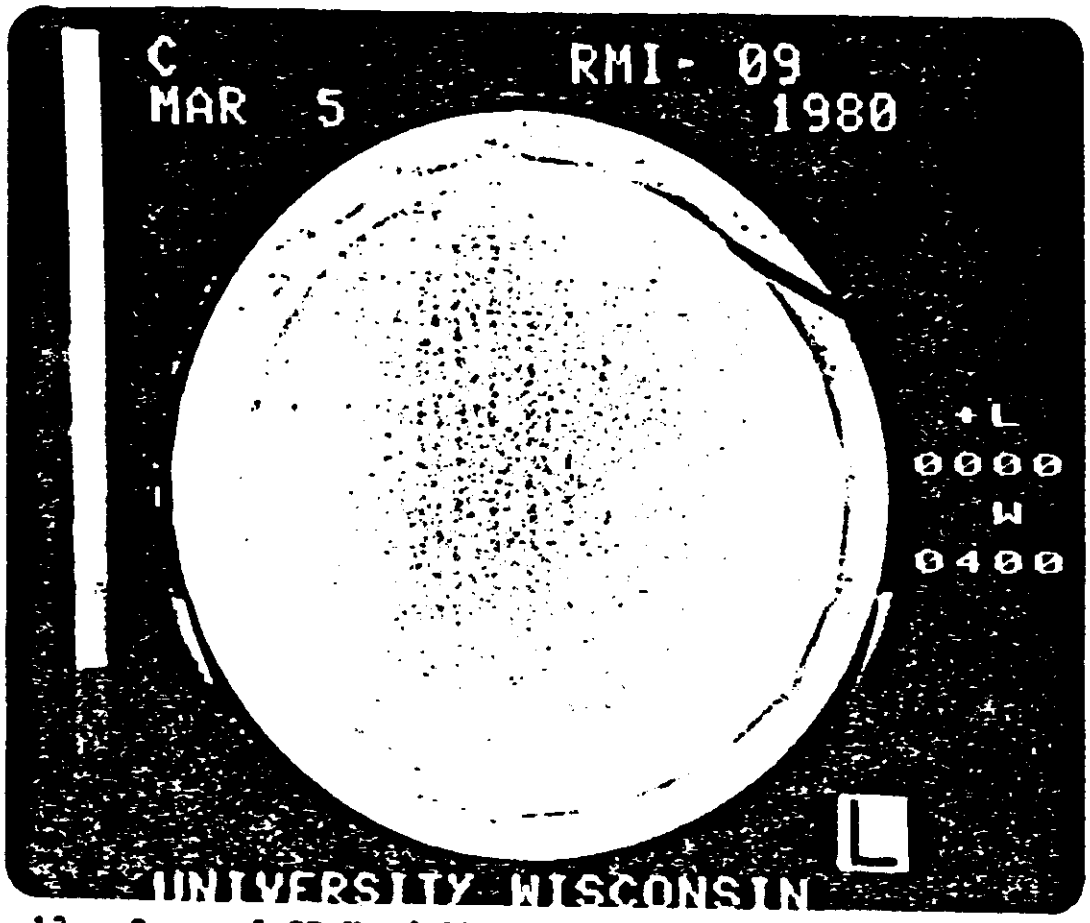


Fig. 13. Scan of CT Head Phantom containing five PLAIN Inserts.

If a CT image of a homogeneous phantom is analyzed by finding the average CT number in various regions distributed over the image, the degree to which these average CT numbers all agree is a measure of the spatial uniformity of the image; if this test is repeated daily in an identical manner, the degree to which the test results of CT numbers remain unchanged over time is a measure of temporal stability of the scanner.

These tests can be easily performed by using the RMI CT Phantom with PLAIN Inserts in all insert cavities. The phantom is positioned (as described in Test 1) and carefully centered in the middle of the scan field. A scan is performed and average CT numbers are determined at various positions in the image. One good method is to measure the average CT number at each insert position, adjusting the region of interest so that it is centered on the insert and covers a substantial fraction of its area. It is important not to make this measurement using the CT number from only one or just a few pixels - deviations of CT number will occur simply because of image noise. A sufficient number of pixels should be averaged so that the error in the average number due to noise is negligible. (Averaging over a minimum of 25 pixels is usually necessary.) It is good practice to average, not over the entire insert, but over an area somewhat within its boundary. If this is done one does not have to worry about CT

number variations that might occur at the insert-cavity interface.

Spatial uniformity can then be determined by observing the range spanned by the measured average CT numbers ($\overline{CT\#}$). Spatial uniformity may be expressed simply as the difference of the maximum and minimum $\overline{CT\#}$ measured; or it may be expressed as a percentage variation in linear attenuation coefficient (LAC) that the CT# difference represents relative to water's LAC:

$$SU = \frac{\overline{CT\#}(max) - \overline{CT\#}(min)}{\text{CT scale factor}} \times 100\%$$

Where "CT scale factor" is discussed on Page 4.

For example, if a head phantom is used to test a CT scanner with a CT scale factor = 1000, and the $\overline{CT\#}$'s measured are 16, 19, 22, 14, and 21, then

$$\overline{CT\#}(max) - \overline{CT\#}(min) = 22 - 14 = 8$$

$$SU = \frac{\overline{CT\#}(max) - \overline{CT\#}(min)}{\text{CT scale factor}} \times 100\%$$

$$= \frac{8}{1000} \times 100\% = 0.8\%$$

The temporal stability of the CT scanner can be determined by day to day comparisons of the $\overline{CT\#}$'s obtained by the spatial uniformity (S.U.) test. A suggested method is to compute the average value of the CT#'s obtained from the spatial uniformity test, call this CT#(ave).

$$CT\#(ave) = \frac{\sum_{i=1}^{N} CT\#(i)}{N}$$

Where N is the number of averaged areas (i.e. the number of inserts).

Then each day the value Δ CT# = CT#(ave) - CT#(actual) is computed where CT#(actual) is the true CT# of the phantom material. (Ideally CT#(actual) is zero, in practice it may be slightly different from zero due to minor manufacturing variations. With each RMI phantom is included a statement of the actual CT# of the water equivalent material.) Δ CT# may show small fluctuations around zero over a period of time. The temporal stability (T.S.) can also be expressed as a percent error relative to water by the following expression:

$$T.S. = \frac{\Delta \text{ CT\#}}{\text{CT Scale Factor}} \times 100\% = \frac{CT\#(ave) - CT\#(actual)}{\text{CT Scale Factor}} \times 100\%$$

The acceptable range for spatial uniformity and temporal stability for any particular CT scanner may be supplied by the manufacturer. In any case, both S.U. and T.S. should be less than 1.0%.

Fig. 13 shows a CT scan of the RMI Head Phantom containing all five PLAIN Inserts in a spatial uniformity/temporal stability/noise scan. The phantom is properly centered and bolused in this scan. It is important to note that some scanners will demonstrate significant artifacts if a phantom is imaged off center or if the phantom is smaller than the scan circle. Shading artifacts may be caused by improper centering, and fall-off of CT numbers toward the edge of the phantom may result from scanning a phantom smaller than the scan circle (Fig. 14). These artifacts may be removed by careful centering of the phantom and applying bolus around it to fill the scan circle. In any case, if the RMI phantoms are treated in the same way as patients, bolused or unbolused, they will demonstrate the type of artifacts that will be present in patient scans. In fact, the RMI phantom can be used to investigate the sensitivity of a scanner to positioning artifacts by intentionally scanning it in off-centered positions (Fig. 14).

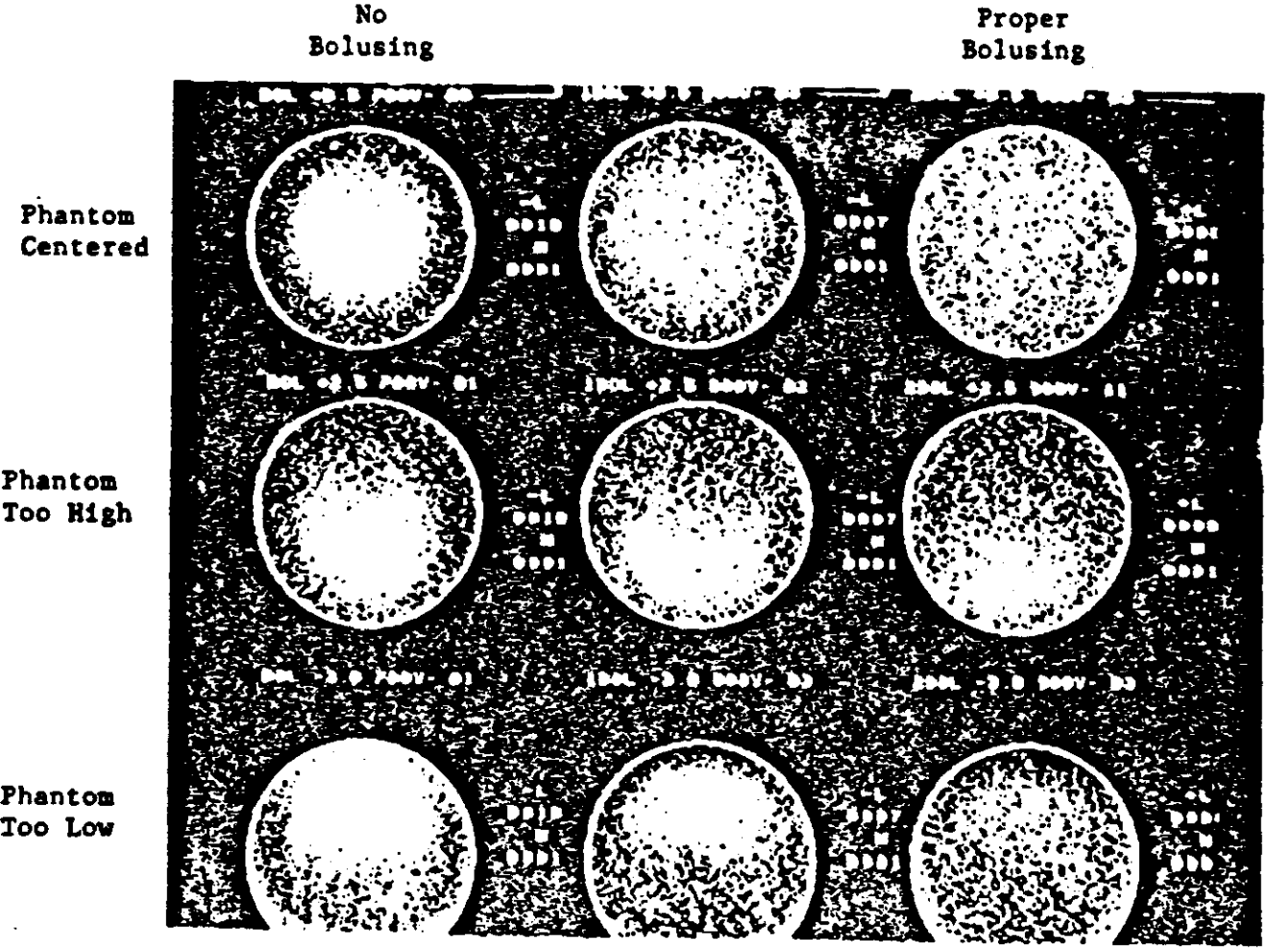


Fig. 14. Examples of image shading due to improper centering and improper bolusing.

4. NOISE

CT image noise is a result of the quantum photon nature of x-rays. It is the equivalent of quantum mottle in conventional radiographic images. As a result of this noise the CT image of a uniform phantom will demonstrate pixel CT number fluctuations even in a perfectly adjusted scanner. CT image noise is usually expressed as the standard deviation of the CT numbers of a large number of pixels taken from the image of a uniform phantom. The actual number of pixels used for the calculation is not important as long as it is sufficiently large (greater than about 25). Most scanners will allow you to select a region of interest and then compute for you the average CT number and standard deviation for all the pixels within that region of interest.

Mathematical Note:

The standard deviation (σ) is given by the following formula:

$$\sigma = \sqrt{\frac{\sum_{i=1}^{N} (CT\#(i) - \overline{CT\#})(2)}{N - 1}}$$

Where CT#(i) is the CT# of the i(th) pixel of the sample, $\overline{CT\#}$ is the average CT# of the sample, and N is the number of pixels in the sample. For a given region of interest about 63% of the pixels will have CT numbers between $\overline{CT\#} - \sigma$ and $\overline{CT\#} + \sigma$. About 95% will be between CT# - 2σ and CT# + 2σ.

The test for CT image noise can be done simultaneously with the spatial uniformity/temporal stability tests, using the same scan image: when values for $\overline{CT\#}$ are read out for the different regions of interest, also read out the values given for the standard deviation or noise.

The fluctuation in CT numbers due to noise can be expressed as fluctuation in apparent linear attenuation coefficient relative to the LAC of water, by using the following expression:

$$\text{Noise} = \frac{\sigma}{\text{CT Scale Factor}} \times 100\%$$

(see Page 7 for an explanation of "CT Scale Factor")

For example if σ = 5.0 and the CT Scale Factor is 1000 then,

$$\text{Noise} = \frac{5.0}{1000} \times 100\% = 0.5\%$$

Many scanners specify noise this way rather than in terms of standard deviation.

It is important to note that the CT image noise is strongly affected by the technique setting (kVp, mA, time), noise smoothing filters in the image reconstruction (which are sometimes at the discretion of the CT operator), patient or phantom size, and pixel size. For meaningful results and comparisons, a CT scanner should be checked in the clinically used modes and only identical modes compared over time.

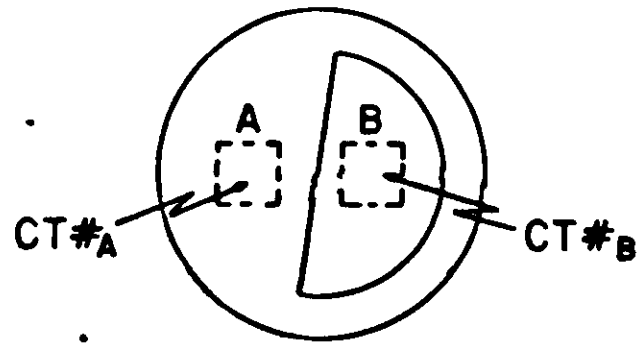
A measurement of image noise is one of the most important indicators of proper scanner function: day to day measurements of image noise are an essential part of CT quality assurance. Any significant increase in noise (at identical techniques) is a sure sign of image degradation and scanner malfunction. It may indicate, for example, a change in kVp or mA calibration, tube degradation before failure, or something as simple as a loose collimator cutting into the scan slice and reducing the amount of x-rays reaching the detectors.

5. CONTRAST SCALE (CT SCALE FACTOR)

As previously discussed on Page 4, the CT number of an image pixel is derived from the measured x-ray linear attenuation coefficient of the corresponding point in the subject being scanned. The relationship between linear attenuation coefficient (μ) and CT number is given by the following expression:

$$\text{CT\# of Substance } \underline{x} = \frac{\mu(x) - \mu(H2O)}{\mu(H2O)} \times \text{CT Scale Factor}$$

where, for most scanners, the nominal value of the "CT Scale Factor" is either 500 or 1000. The actual value of the CT scale factor can be determined by performing the following contrast scale test.



AREAS OF INTEREST TO MEASURE

Fig. 15. Illustration of the EDGE/CONTRAST SCALE Insert showing regions of interest.

The EDGE/CONTRAST SCALE Insert (illustrated in Fig. 15) is constructed of two materials A and B, that differ in linear attenuation coefficient (μ). After the insert has been scanned and the CT#s of the materials A and B determined, the actual "CT Scale Factor" can be calculated from the equation:

$$\text{CT Scale Factor} = \frac{|CT\#(B) - CT\#(A)|}{\Delta\mu/\mu(H2O)}$$

where $\Delta\mu$ is the difference in linear attenuation coefficient for the two materials: $\Delta\mu = \mu(A) - \mu(B)$; $\mu(H2O)$ is the linear attenuation coefficient for water, CT#(A) and CT#(B) are the measured CT numbers of materials A and B. For the EDGE/CONTRAST SCALE Insert 460-011, $\Delta\mu$ for the two materials is nominally 10% of $\mu(H2O)$, or $\Delta\mu/\mu(H2O) = .10$. The actual value of $\Delta\mu/\mu(H2O)$ for a given EDGE/CONTRAST SCALE Insert is indicated on the QC test sheet included with each phantom kit. EDGE/CONTRAST SCALE Insert is indicated on the QC test sheet included with each phantom kit.

The results of this test can be expressed in another way as the percent change in linear attenuation coefficient, relative to water corresponding to a change of one CT number. This measure, which will be called contrast scale (CS), is calculated from the equation:

$$CS = \frac{\Delta\mu/\mu(H2O)}{|CT\#(B) - CT\#(A)|} \times 100\%$$

As an example, suppose CT#(A) = 100 and CT#(B) = 6. Then with $\Delta\mu/\mu(H2O) = .100$ for the EDGE/CONTRAST SCALE Insert in use:

$$\text{CT Scale Factor} = \frac{94}{.100} = 940$$

$$CS = \frac{.100}{94} \times 100\% = 0.106\%/CT\#$$

To perform the contrast scale test, place the EDGE/CONTRAST SCALE Insert into the center cavity of the phantom. (There may be a slight position dependance of contrast scale with some scanners. This can be investigated by placing the test insert into other positions in the phahtom.) You may find it suitable to place the test insert in an off center position if a protocol is developed employing other test inserts during the same scan. If an edge response or MTF test is to be performed simultaneously with the contrast scale test, then the test insert should be oriented so that the interface makes a small angle (about 5 degrees to 10 degrees) to the vertical (Fig. 19). The label on top of the insert shows the orientation of the edge within the insert. After the

insert is in place and the phantom is properly positioned and centered (See Test 1), take a scan of the phantom. (Exact positioning in the scan plane and tilt elimination is not critical for this test. However, the phantom should be carefully centered in the scan field to prevent shading artifacts.)

The image is analyzed by finding the average CT number value for a region of interest in the center of each half of the test insert (Fig. 15) and then applying the calculation discussed above. It is important that the regions of interest be large enough to average image noise and give an accurate figure for the average CT#, yet one should be careful to avoid averaging in pixels close to the interfaces. An abnormal variation of the value obtained for the contrast scale (or CT scale factor) may indicate the need for scanner recalibration or, perhaps, signal a more serious scanner malfunction.

6. HIGH CONTRAST SPATIAL RESOLUTION

High contrast spatial resolution can be evaluated by using the HIGH CONTRAST RESOLUTION insert (Part #460-012B). This insert contains a pattern of seven rows of straight, parallel rods. A row contains five rods of equal diameters with a center to center spacing of twice the rod diameters. The six rows contain rods of 1.5, 1.25, 1.00, 0.8, 0.5, 0.4 diameter. Fig. 16 shows the insert configuration. The rods are air contrast holes surrounded by water equivalent plastic. The resulting contrast is 100%. The rods are 4 cm long and extend through the entire slice width, if the phantom is properly positioned in the CT scanner. An optional High Contrast Insert (460-012C) with a pattern of six rows of rods 2.0, 1.75, 1.5, 1.25, 1.0, and 0.75 mm in diameter is also available. A scan of this optional insert is shown in Fig. 17.

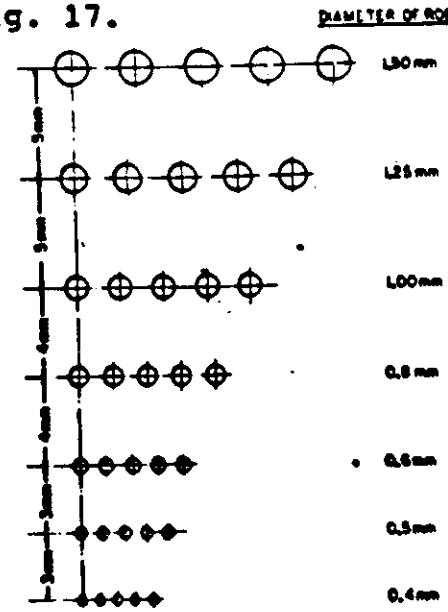


Fig. 16. Diagram of HIGH CONTRAST RESOLUTION INSERT (#460-012B) showing rod geometry.

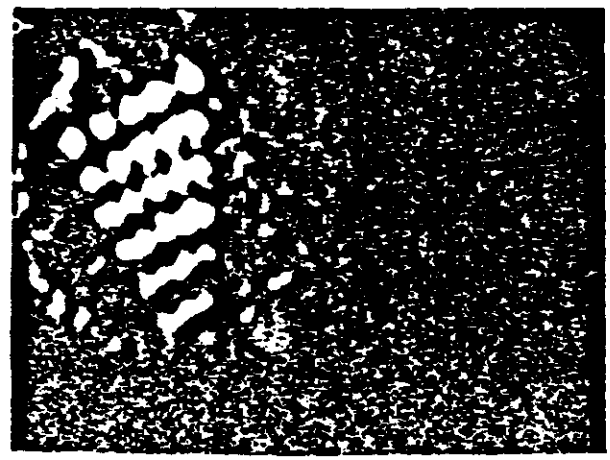


Fig. 17. Scan of HIGH CONTRAST RESOLUTION Insert - 100% contrast (#460-012C).

The resolution test may be done in any of the phantom's insert positions, and may be used to test both vertical and horizontal resolution by rotating the insert. The orientation of the rod pattern is indicated on the insert label. It is important that any phantom tilt with respect to the scan plane be eliminated before performing this test (see Test 1). The presence of phantom tilt may cause blurring of the test pattern due to partial volume effects and result in this test yielding a poorer resolution figure than deserved by the scanner.

The resolution determined with the rod pattern will depend somewhat on the viewing conditions and the resolution criteria used (Fig. 18). Therefore it is important that a fixed protocol be developed for evaluation of these images so that consistent comparisons can be made. A row of rods may be considered resolved if all five rods can be perceived with some discernible spacing or lowering of density between them. No two rods should merge into an apparent single entity, nor should the pattern of rods be grossly distorted. Aliasing, or "false resolution", may rarely occur where the rods appear to be separated, but in fact their positions are distorted: for example, there may only appear to be four rods with their imaged position in between their actual positions in the phantom. Rows exhibiting such "false resolution" should not be considered resolved.

Fig. 18. Scan of the same HIGH CONTRAST RESOLUTION Insert - 100% contrast (#460-012C) at two different, viewing conditions.

To give examples of some possible protocols for viewing conditions, suppose that in an image of the HIGH CONTRAST RESOLUTION Insert the background measures approximately CT# = 0 while the largest rods have a CT# of about 100.

Protocol 1: Set window width to 100 and window level to 50. Reduce window level if necessary to perceive resolution of smallest resolvable row.

Protocol 2: Set window width to smallest value ("measure", "0", or "black-white image"). Allow any window level. Image of row

is resolved if by varying the window level an image can be obtained of five black objects with white separation between them.

In any case it is probably best to make the resolution determination from the viewing screen rather than from film images to allow the viewer to alter the viewing parameters if necessary.

7. EDGE RESPONSE - MTF

High contrast spatial resolution (Test 6) is a good test for image sharpness that can be easily and quickly implemented. However, more complete information about image sharpness can be obtained by utilizing the EDGE/CONTRAST SCALE Insert to measure edge response and modulation transfer function (MTF). This is a test, usually performed by a physicist, which involves mathematical manipulation of the individual CT numbers in the vicinity of the image of an edge. It requires access to the individual CT numbers by means of a print out or C.R.T. dis-play. The procedure to determine the edge response, taking its spatial derivative to obtain the line spread function, and then doing a fourier transformation to arrive at the modulation transfer function is described in an article by P.F. Judy, (Ref. 19).

To perform this test with the RMI CT Phantom, first check that the phantom is properly positioned and centered (See Test 1). It is very important that the phantom not demonstrate any tilt with respect to the scan plane as this would blur the image of the edge due to partial volume effects. Place the EDGE/CONTRAST SCALE Insert into the center cavity (2) of the phantom. The insert should be oriented so that the interface makes a small angle (about 5 degrees to 10 degrees) to the vertical (Fig. 19). In this way the number of data points defining the edge response is increased, and test accuracy is improved. The label on top of the insert shows the orientation of the edge within the insert. After the insert is properly positioned, take a scan of the phantom. A "Low Noise" scan (high mA and/or long time) is preferred for the most accurate test results. Reducing the image noise by increasing the mA or scan time should not affect the image sharpness as measured by the edge response or MTF. However, image noise on some scanners can also be affected by a choice of image reconstruction algorithm image smoothing options. These adjustments will affect <u>both</u> image sharpness and image noise. Their effect on changing the results of this test should be recognized..

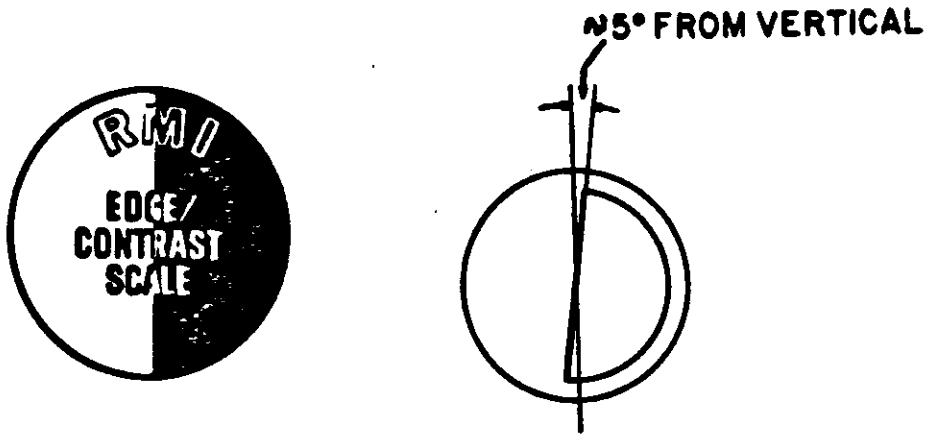


Fig. 19. Proper orientation of EDGE/CONTRAST SCALE Insert for Edge Response - MTF test.

This test can also be performed by placing the EDGE/CONTRAST SCALE Insert into a cavity other than the central one. Significantly different results may be obtained in these off-center positions. Tests may be done with the edge in both the vertical and horizontal orientations to observe image sharpness in the tangential and sagittal directions.

8. MECHANICAL ALIGNMENT

The ALIGNMENT Insert (Part #460-014) contains an aluminum pin with a diameter of 6 mm (1/4") (Fig. 20). This insert can be scanned in any of the positions within the phantom. Evaluate the test simply by examining the image of the pin to see if it contains artifacts. Scans of the insert are shown in Fig. 21. Many of the artifacts seen in this test are similar or identical to those caused by patient motion or internal organ motion. Correctable artifacts due to scanner misalignment may be falsely attributed to patient motion and ignored unless a mechanical alignment test is performed (using a <u>stationary</u> phantom rather than a <u>moving</u> patient).

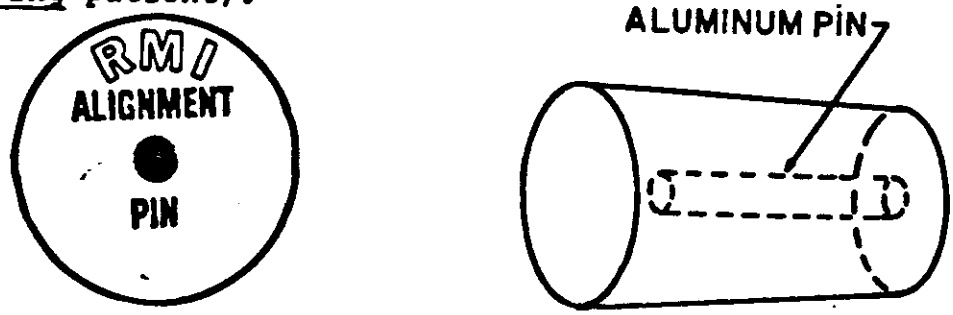


Fig. 20. Diagram of ALIGNMENT Insert (Part # 460-014) and label.

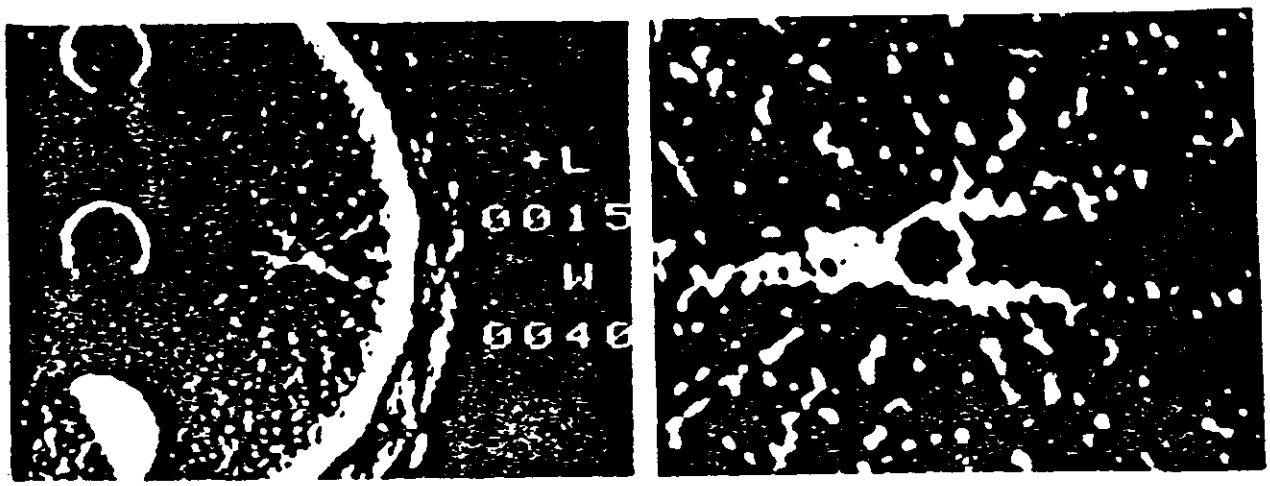


Fig. 21. Scans of ALIGNMENT Insert (#460-014).

An important component of a CT scanner is the device which monitors the varying position and angle of the x-ray source relative to the detectors. This device must be carefully adjusted to typically less than 1 mm tolerances otherwise image artifact will result (Ref. 35). Misalignment of the detectors relative to the x-ray source or to the center of rotation, and errors in scanner motion (perhaps due to faulty computer control) can all add to image artifacts. Different scanner geometries and motions may result in different forms of artifact using the ALIGNMENT Insert. With second generation (translate-rotate) CT scanners for example, two common artifacts are the "Tuning Fork" artifact and the spoke pattern artifact (Fig. 21). Your CT service person should be able to interpret the artifacts and make the proper adjustments to reduce or eliminate them.

9. CT NUMBER LINEARITY - STABILITY

This test involves scanning various materials with known linear attenuation coefficients (μ) and determining their CT numbers from the image. A plot is then made of CT number versus μ, which should be a straight line. This test is somewhat similar to the contrast scale test, but covers a larger range of CT numbers and involves four measurement points, instead of two.

The LINEAR Insert (Part #460-020) along with 1" (2.5 cm) diameter Teflon, polyethylene, and acrylic rods (Part #460-020C, B, and A respectively) are used in this test (Fig. 22). Optional rods of other materials are available from RMI. The LINEARITY Insert should be placed in the center position (2) of the phantom. Three scans are taken of the phantom one with each of the three plastic rods inserted into the LINEARITY Insert. Finally a scan is taken with a PLAIN Insert in the center position. The average CT number of a region of interest within each of the four samples is determined and then plotted on a graph versus the attenuation coefficients known for the materials (Fig. 23). (These attenuation coefficients will be a function of the scan kVp.) Values of the attenuation coefficients for various materials are given in References 1, 22, 23, 24, 26, 32, 46, and 47.

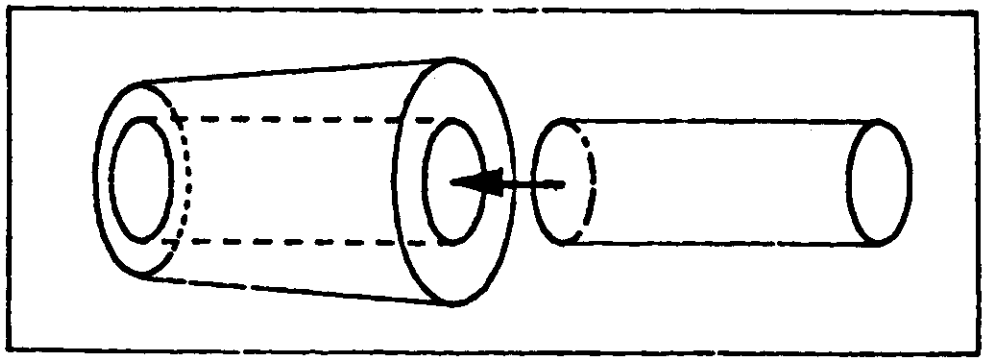


Fig. 22. Schematic diagram of LINEARITY Insert (Part #460-020) and plastic rod.

Optionally the CT numbers of the measured materials can be plotted vs. the attenuation coefficient of the material divided by the attenuation coefficient of water (Fig. 24). If plotted in this way, the slope of the curve is the "CT Scale Factor" discussed on Page 7 and in Test 5. The reciprocal of the slope times 100% is the contrast scale (CS) defined in Test 5. In any case, the resultant plot should ideally be a straight line with no change in slope over time. Depending on the beam hardening correction employed by the scanner under test, the measured CT# of Teflon may be slightly depressed and fall below the curve drawn from the three other points. The presence of beam hardening artifact in the Teflon can be verified by raising the window level so the Teflon can be visualized and observing a "cupping" artifact: The Teflon rod appears to have a lower CT# in its center than near its periphery.

Another way to use the LINEARITY Inserts is to simply monitor the temporal stability of the CT numbers, keeping a record of the measured CT numbers and noting any change in time.

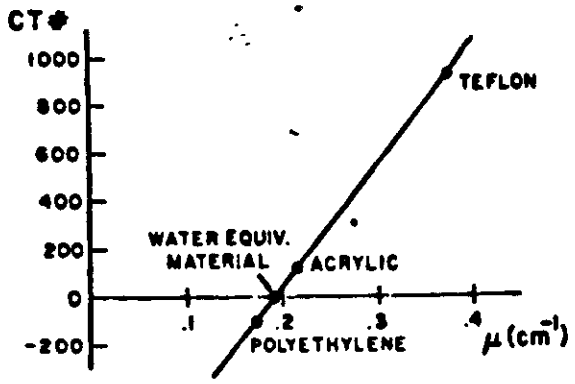


Fig. 23. Example of linearity plot, CRT# versus μ of various plastics.

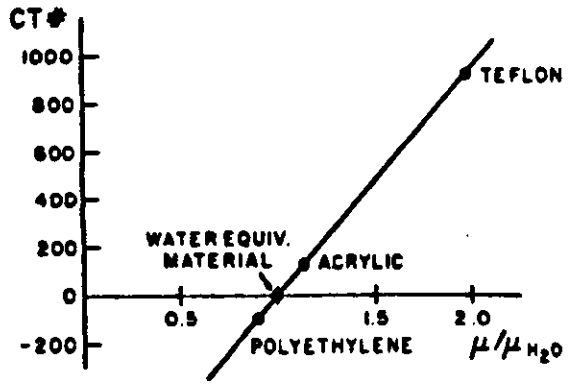


Fig. 24. Example of linearity plot, CT# versus μ / μ (H20) for various plastics.

10. SLICE SENSITIVITY PROFILE

For this test the ALUMINUM RAMP Insert (Part #460-017A) is employed. This insert contains a 0.5 mm thick strip of aluminum oriented at 26.6 degrees to the scan plane. (Fig. 25). The label (Fig. 26) shows the orientation of the ramp in the insert and Fig. 27 shows a scan image of the ALUMINUM RAMP Insert. The ramp has a small hole in it that is positioned in the plane of the circumscribed line on the phantom when the insert is firmly placed into one of the phantom's cavities.

Although the HELIX Insert allows a quick measurement of slice thickness, more accurate results, along with information on the slice sensitivity profile, can be obtained using the ALUMINUM RAMP Insert. It is especially useful for accurate measurements of the smaller slice thicknesses now in common use with modern CT scanners.

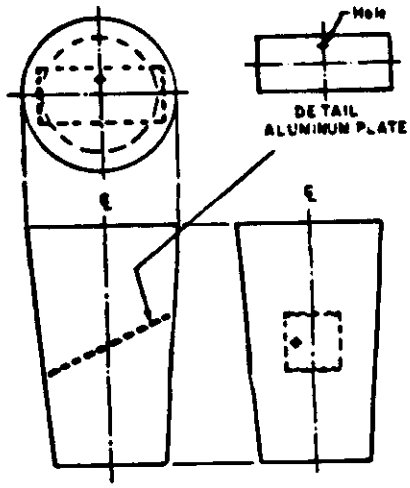


Fig. 25. Schematic of ALUMINUM RAMP Insert (Part #460-017A).

Fig. 26. Label which shows the orientation of the aluminum ramp in the insert.

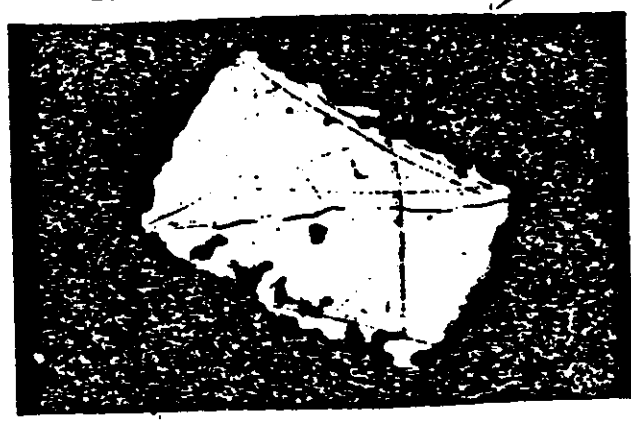


Fig. 27. Scan of ALUMINUM RAMP Insert.

Since the ramp is oriented at 45 degrees to the scan plane, the CT number profile along its length directly yields the slice sensitivity profile: one millimeter along the length of the Ramp image is equivalent to one millimeter depth through the slice thickness. To obtain the slice sensitivity profile or the slice width, first properly position the phantom, eliminating any tilt relative to the scan plane. If this test is performed with the phantom tilted, inaccurate results will be obtained. Next, firmly position the ALUMINUM RAMP Insert into one of the phantom cavities. For the most accurate measurements, it is best to

rotate the insert so that the length of the ramp does not lie along the horizontal or vertical directions. (IF the ramp did lie along these directions the test accuracy would be limited by the pixel size.) Take a scan of the phantom; best results are obtained with a high resolution, high mA, and/or long time scan. If a number printout of the ramp image is available the CT numbers can be used to plot the slice sensitivity profile (Fig. 28).

A simpler measurement is a direct determination of the slice width as the full width half maximum (FWHM) of the sensitivity profile (Fig. 28). This is easily done with any scanner that can measure distances between two points in the image, by using the following procedure:

1) Set the window width at "zero" or "measure" to give a black-white display with no gray shades.

2) By adjusting the window level, determine the average CT number [CT(max)] of the most dense part of the ramp (at the midline of the ramp image).

3) Determine the average CT number (CT(min)) of the background beyond the end of the ramp image.

4) Adjust the window level to the number midway between CT(max) and CT(min).

5) Measure the length of the ramp at that window level. The length obtained in this way is the FWHM of the sensitivity profile: the slice width.

If the scanner does not have the capacity to measure distances on the image, the FWHM can still be determined by following steps 1 through 5 and then producing a film of the phantom image, preferably a magnified view of the ALUMINUM RAMP Insert. Measure the imaged length of the ramp and divide it by the measured diameter of the insert image. Multiply this number by 45 mm (the actual diameter of an insert at the scan plane) to obtain the FWHM.

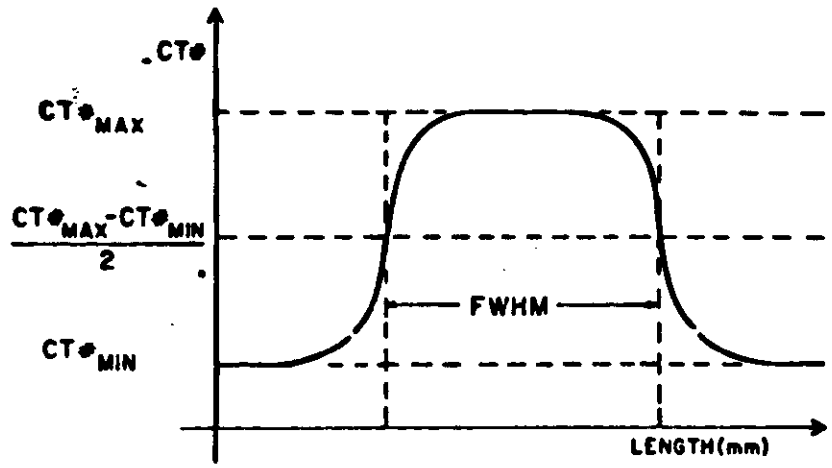


Fig. 28. Plot of Slice Sensitivity Profile.

11. IMPULSE RESPONSE (POINT SPREAD FUNCTION)

The optional IMPULSE Insert (Part #460-021) contains a 0.25 mm diameter steel wire surrounded by water equivalent plastic (Fig. 29). Since the wire is embedded in solid plastic, it cannot sag with time and its straightness is assured. This insert can be used to measure the impulse response (point spread function) and also the MTF of the scanner system. For MTF measurements, however, use of the EDGE/CONTRAST SCALE Insert (See Test 6) will generally give more accurate results less affected by scanner noise.

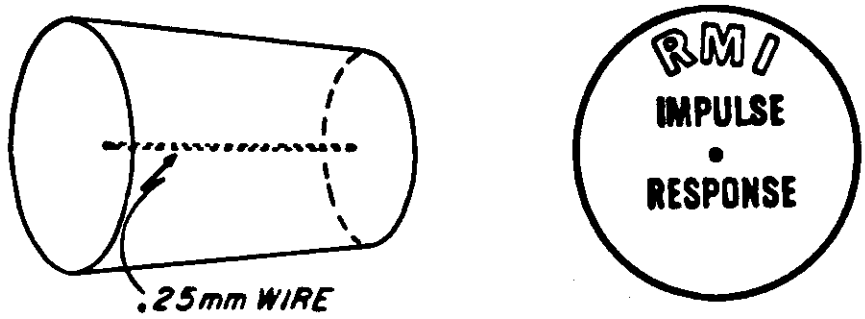


Fig. 29. Schematic of IMPULSE RESPONSE Insert and label.

It is important that the phantom be properly positioned so that any tilt with respect to the scan plane is eliminated before this test is performed. After the IMPULSE Insert is positioned into any of the phantom cavities and a scan performed (high mA and long time), the point spread function can be obtained from a printout of the CT numbers in the vicinity of the wire's image. A method of using the IMPULSE Insert to quantitatively investigate image sharpness without the need of a printout is given in the following protocol:

1) Set the window width to "zero" or "measure" to obtain a black-white display with no gray shades.

2) Increase the window level until the wire's image completely disappears, than slowly decrease the window level until the first pixel appears. Record the CT number at which it appeared.

3) Decrease the window level and record the CT numbers at which all subsequent pixels appear (Fig. 30), along with their appearance ranking: 1st, 2nd, 3rd, etc.

4) Plot the CT number of a pixel's appearance vs. (the square root of its appearance ranking) times (the pixel size) (Fig. 31).

The resultant graph will be closely related to the point spread function (it <u>is</u> essentially the point spread function if that function is spatially symmetric).

PIXEL RANKING	CT NO.
1	405
2	296
3	234
4	166
5	131
6	115
7	97
8	50
9	33
10	21
BAKGROUND	-6

Fig. 30. Sample record of CT# of pixel appearances.

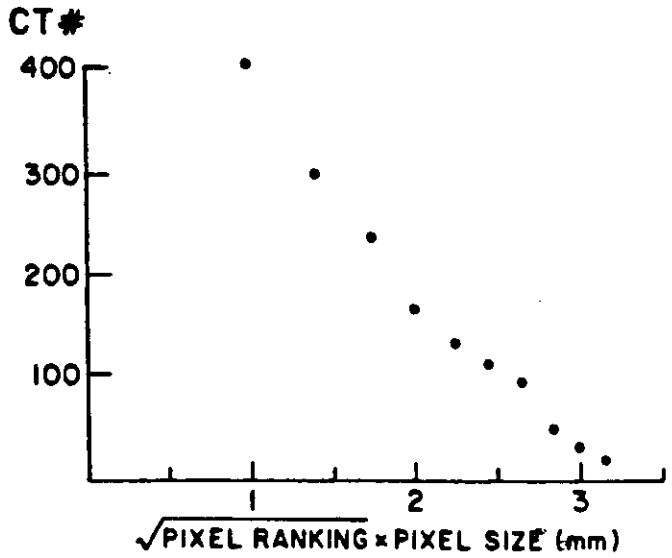


Fig. 31. Graph of pixel appearances.

12. LOW CONTRAST DETECTABILITY

One of the most significant properties of CT scanners is their ability to perceive the presence of low contrast lesions that are quite invisible by any other radiographic method. Therefore, monitoring low contrast detectability is an important component of CT quality assurance and even more essential to scanner intercomparisons and acceptance testing. Noise measurements usually correlate well with detectability measurements in monitoring a particular CT scanner and imaging mode over time. However, noise measurements by themselves may give very little information as to how two different scanners compare in their low contrast performance. The only valid way to make the comparison in this way is usually by direct measurement of low contrast detectability.

The LOW CONTRAST DETECTABILITY Insert (Part #460-013A) is made of two epoxy resins which provide a contrast of approximately 0.6%. An insert with 0.3% nominal contrast is also optionally available (Part #460-013B). The insert consists of a cylinder of resin A, containing holes of the following sizes: 2, 2.8, 4, 5.6, and 8mm (Fig. 32). The holes increase in size geometrically by multiples of 1.41 ($\sqrt{2}$). Resin B has been poured around the cylinder of Resin A and also fills the holes in Resin A.

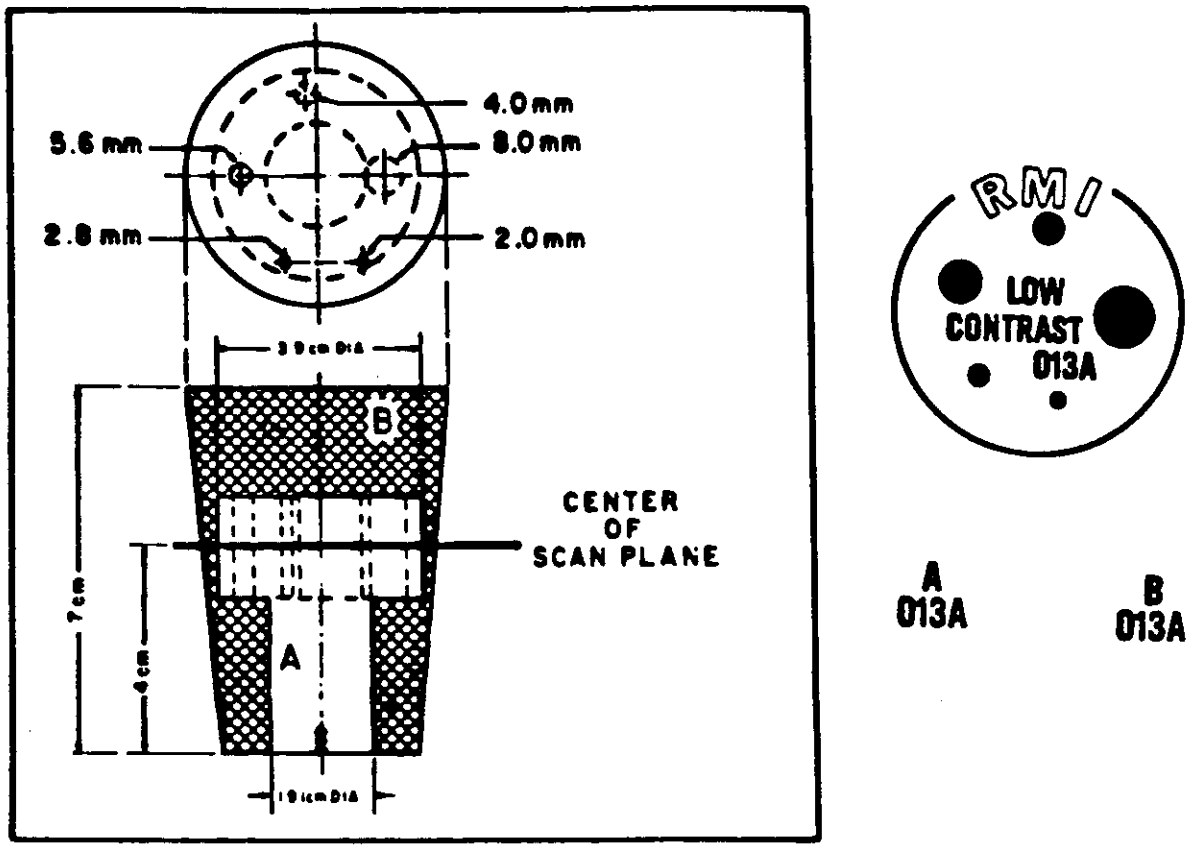


Fig. 32. Diagram of LOW CONTRAST DETECTABILITY Insert (Part #460-013) and label set.

Scanning the LOW CONTRAST DETECTABILITY Insert (LCDI) should begin only after the RMI phantom has been properly aligned in the scan plane, including tilt elimination (See Test 1). Because a low contrast detectability test is very subjective by its nature, the best results are obtained if the LCDI is scanned several times to test a particular scan technique or if several LCDI's are placed in the phantom and scanned simultaneously. In this way several images can be inspected to determine the detectability limit and test reliability is improved.

There is an important psycho-physical visual phenomenon concerning perception of low contrast detail which one should be aware of when performing this test: Perception of low contrast images is strongly affected by the viewing distance. Increasing the viewing distance (up to some limit) will generally allow the perception of lower contrast objects. This effect may be observed by looking at a CT image of the LCDI. If viewed up close at a normal viewing distance possibly none of the low contrast target objects may be visible. But as the viewing distance is increased some of these objects may then come into view.

Since the low contrast detectability test is intended as a test of the CT scanner and not of the human visual system, it is essential that the test observer be allowed full freedom of viewing distance to permit best visualization of the test image.

This may pose a problem if the test evaluation is attempted using the image of the CRT viewing screen. There may simply not be enough room to back up and increase your viewing distance for best visualization (up to 10 feet may be necessary). In this case it is best to use films for the test evaluations. Use a window width of from 1/20 to 1/10 of the CT scale factor (See Page 4) and a window level approximately equal to the imaged CT number of the low contrast detectability insert.

Included with the LOW CONTRAST DETECTABILITY Insert are two sample cylinders, one of resin A, and the other of resin B. These cylinders will fit into the LINEARITY Insert (Part #460-020). To verify the actual percent contrast presented by a particular LCDI for a particular kVp and scanner, scan the Resin A and B cylinders in the central phantom positions (using successive scans). The difference in average CT numbers for the two cylinders divided by the "CT Scale Factor" and multiplied by 100% will yield the contrast possessed by the LCDI. The composition of the resins used in this insert indicate that there should be very little if any, variation of contrast with kVp, though this energy independance can be verified by the above method. The actual contrast of an individual insert may be slightly different than the nominal 0.6% due to subtle manufacturing variations which cannot be ignored at these low contrast levels. However, the larger cylinders of resin A and B allow you to verify the actual contrast level for yourself.

13. BEAM HARDENING

Artifacts in CT images due to beam hardening effects have been often discussed in the CT literature. These artifacts are due to the inhomogeneity (polychromaticity) of the x-rays used in CT scanning. The CT reconstruction process assumes a unique value for the linear attenuation coefficient at every point in the object. Highly attenuating objects in the scan field shift the existing x-ray spectrum to a higher effective energy for x-rays which have paths through these objects. Since the linear attenuation coefficients of all the materials in the x-ray beam are energy dependent (and the form of the energy dependence itself varies with the effective atomic number of the material), inconsistencies in the image reconstruction result due to non-unique values of attenuation coefficient perceived by different x-ray paths through the same object. The result of these inconsistencies is image artifact.

The nature and magnitude of beam hardening artifacts can be investigated using the COMPACT BONE ROD (Part #460-020H). This insert is a 1" (2.5 cm) rod of simulated compact bone material which can be inserted into the LINEARITY Insert (See Fig. 22). After these inserts are placed into one of the phantom's cavities and scanned, the resultant image can be investigated for both internal and external artifacts. Internal artifacts (within the image of the bone insert) can be observed by raising the window level so that the insert will be displayed within the gray scale.

Commonly a cupping artifact will be observed due to beam hardening effects. The center of the bone image will demonstrate a lower CT number than its periphery (Fig. 33). External artifacts (outside of the bone image) may include raising or lowering of the imaged CT numbers in the vicinity of the bone image (Fig. 34). These artifacts may be caused by beam hardening or in part by such things as detector afterglow, x-ray scatter, or quirks in the image reconstruction algorithm. Misalignment streaks similar to those seen with the ALIGNMENT Insert (Test 8) may also be visible.

If two high density objects are placed into the scan field, beam hardening effects will commonly cause a low density streak to appear between the two objects (Fig. 35). This type of artifact is also of some clinical significance. For example, it is partially the cause of the intrapretrous lucency (Hounsfield artifact). It can be investigated using two LINEARITY Inserts and two bone inserts.

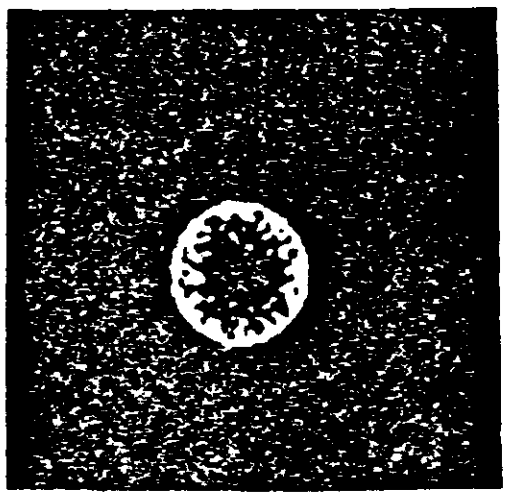


Fig. 33. Cupping artifacts in Bone Rod Image.

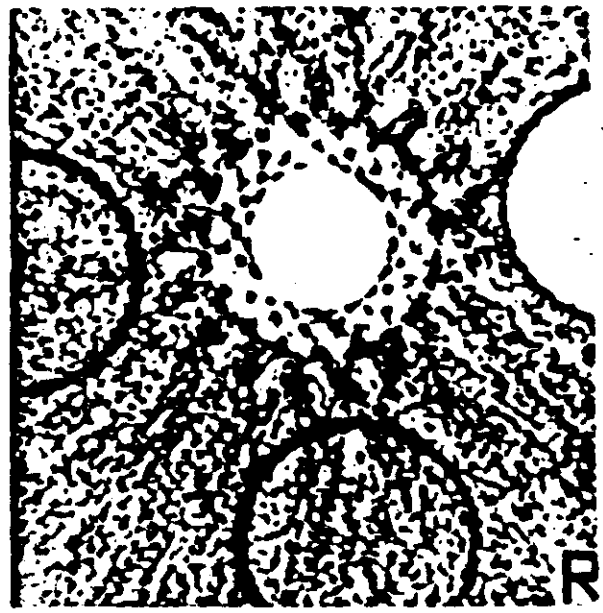


Fig. 34. Scan of Bone Rod at tissue window level.

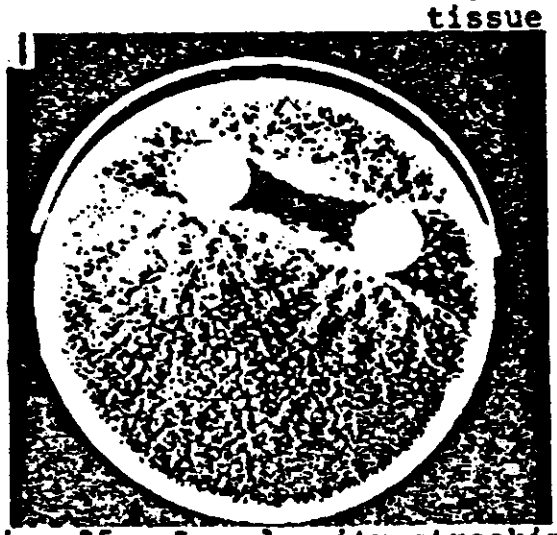


Fig. 35. Low density streaking artifacts between two Bone Rods.

14. DOSIMETRY

The DOSE Insert (Part #460-019) along with a low energy pocket dosimeter or ion chamber can be used in RMI's CT Phantoms to measure integral dose. Alternatively, the TLD HOLDER (Part #460-018) can be loaded with TLD chips and placed into the DOSE Insert to measure the dose profile.

The DOSE Insert is essentially a PLAIN Insert with a hole bored into it to accept the TLD HOLDER (Fig. 36). It will also accept pocket dosimeters or ion chambers of diameter 14.5 mm or less. RMI can supply DOSE Inserts with holes bored to the customer's specification to fit chambers of various diameters.

In preparation for dosimetry measurements the scribed line around the CT phantom should be placed in the middle of the scan slice. The phantom's positioning should be verified by a scan using the HELIX Inserts (As Explained in Test 1), and corrected if necessary.

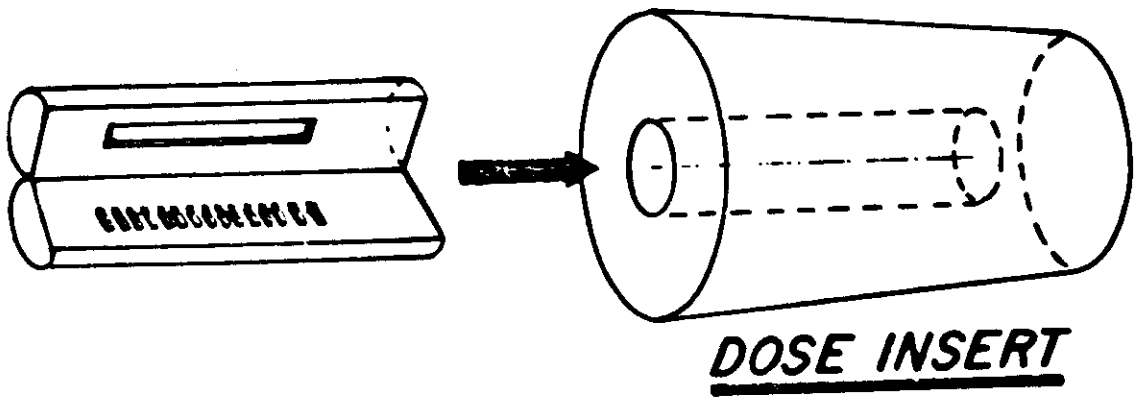


Fig. 36. DOSE INSERT with TLD HOLDER.

To measure the dose profile at any of the insert positions; load the TLD HOLDER with 15 TLD chips, insert the TLD HOLDER fully into the DOSE Insert and place the insert into the desired phantom cavity. The middle TLD will be automatically positioned in the plane of the phantom's exterior scribed line; if the phantom has been properly positioned, the middle TLD will be at the scan slice. Surface dose profile measurements can be performed by taping the TLD HOLDER to the surface of the phantom, making sure it is aligned parallel to the scanner axis (perpendicular to the scan plane).

The TLD HOLDER will hold 15 TLD chips to cover a 40 mm width. The geometry of the TLD chip positions is given in Fig. 37. The central chips have center to center separations of 2 mm; this increases to 4 mm for the outermost chips.

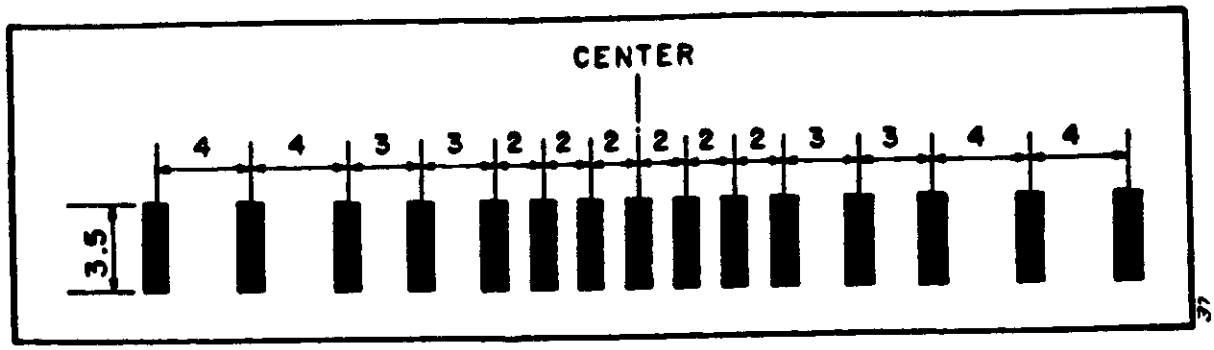


Fig. 37. Geometry of TLD chip positions. All indicated dimensions are in mm. TLD chip cavities are 1 mm wide.

To measure integral dose at positions within the phantom, insert a low energy pocket dosimeter or ion chamber into the DOSE Insert so that the center of its active length coincides with the middle of the scan slice (40 mm above the bottom of the insert) see Fig. 36. RMI can supply DOSE Inserts to fit any cylindrical chamber of diameter 25 mm or less or can supply individual PLAIN inserts that can be bored out by the customer. To order a custom DOSE Insert from RMI, specify the chamber diameter and the distance from the center of the active length to the physical end of the chamber.

The DOSE Insert plus dosimeter or ion chamber can be placed into any of the phantom cavities, or alternatively the dosimeter or ion chamber can be placed directly on the surface of the phantom to measure surface integral dose.

How long of an active length is necessary for the chamber to accurately measure integral dose: Surface dose is produced principally by the primary beam that has been collimated to the selected slice thickness. Thus for a 8 mm nominal slice thickness, the band of principal irradiation around the phantom's surface would ideally have a width of 8mm; in practice while some scanners approach this ideal closely, other scanners irradiate up to twice as wide an area: In this case up to 16 mm wide. For surface integral dose measurements a chamber with an active length of 3 to 4 times the nominal slice thickness should be sufficient to detect, essentially, the entire integral surface dose for a single slice exposure.

At points interior to the phantom scattered radiation contributes significantly to total dose so that the dose profile becomes wider than at the surface. In addition, a significant fraction of the total integral dose is contained in the long tails of the dose profile which are due to scattered radiation. If a chamber of active length 3 to 4 times the nominal slice thickness is used at an interior position it will not record the entire integral dose, but rather somewhat less than the total. In any case, the fraction of the total that it does detect will be essentially fixed for a particular insert position and scanner unit, and will not be very sensitive to slight variations in chamber positioning. For this reason this method is still useful as a quick quality assurance check to detect any changes in

dosimetry over time.

Calculating integral dose:

Of course ion chambers measure exposure in units of Roentgen (R) and to obtain absorbed dose for soft tissue (in units of rad) you should multiply the indicated exposure by 0.86. The units of integral dose are rad - cm. To obtain integral dose use the following formula:

$$I = 0.86 \times R \times L$$

I is the integral dose in rad - cm.

R is the exposure reading in Roentgen from the ion chamber-exposure meter or pocket dosimeter due to a single slice exposure

L is the active length of the chamber in centimeters.

From the integral dose value obtained by a single slice exposure you can calculate the average dose due to multiple slices by dividing the integral dose (I) by the table increment. Alternatively you can measure this average dose directly by multiple exposing of the chamber as you increment the table. The total table motion should exceed the active length of the chamber.

The Capintec PM-05 ion chamber performs quite well when used with the RMI Phantom for integral dose measurements. It has an active chamber length of 4.7 cm, and fits precisely into the regular DOSE Insert.

15. TABLE INCREMENTATION AND LOCALIZING LIGHT ACCURACY, DOSE PROFILE USING FILM

This test measures the dose profiles of the various nominal slice thicknesses and also determines the accuracy of the localizing light(s) and table incrementation mechanisms. All of this data is quickly and accurately documented on a few sheets of film.

In addition to the RMI CT Phantom, you will need the following materials to collect data for this test: A few sheets of Ready Pack Kodak X-omat V therapy verifications film or its equivalent (TL film will generally result in overexposure unless very low technique factors are available), a sheet of flexible lead-vinyl shielding about 15 cm square, a felt tip pen or marker, and a pin. 25 cm x 30 cm Ready-Pack film is quite suitable. It can be cut into two 25 cm x 12.5 cm strips and taped closed in the dark. Using the pin, each of the films should be punctured at it's two ends with some code to indicate left and right orientation. A third pin marking should indicate the center of the "HEAD" side of the film.

Before beginning this test the RMI Phantom should be properly centered in the scan plane. The phantom should be positioned so that the plane indicated by the localising light is about 1 cm from the edge of the phantom. A sheet of Ready-Pack film should be taped onto the top of the phantom according to the orientation marks on it. The position of the localizing light(s) should then be marked on the paper of the Ready Pack using the felt tip pen. If the localizing light is outside the scan plane, move the phantom into the scan plane using normal procedures. Perform a single scan using the smallest available slice thickness. Next, securely tape the flexible lead sheet over the Ready-Pack so that it covers only the right side. Make four more exposures, incrementing the table in the usual manner before each scan. Shift the flexible lead sheet so that it now covers three more exposures, incrementing the table by the same amount but in the opposite direction before each scan. Remove the Ready-Pack and make several punctures in it along the marks which indicate the position of the localizing light(s). This procedure can be repeated with another Ready-Pack film if you wish to test the accuracy of other incrementation distances or other methods of the table incrementation.

After the films have been developed, the following data can be obtained from each film: 1) The error in the localizing light(s) at the left, top, and right side of the phantom can be measured by comparing the positions of the puncture marks with the center of the first exposed strip on the film. 2) By measuring the distances between the exposed strips on the left and right sides of the film, the table incrementation accuracy and reproducibility in each direction can be determined. 3) By comparing the exposed strips on the left and right sides, you can check the positioning accuracy that is maintained by the table when reversing its direction of motion. A problem here will show itself as a non-coincidence or "jog" in the exposed strips on the

two sides.

To measure dose profiles place a Ready-Pack film on the properly centered phantom, as before, and make a single exposure for each different slice thickness. Increment the table between scans so that the profiles do not overlap. Analysis of the developed film can be done in two ways. A quick estimate to the irradiation slice width can be made by visual inspection of the exposed strips, although an accurate ruler can be used. More precision is possible using a 6x comparator (such as Edmund Scientific's No. 30,169.) To obtain an accurate estimate the film must not be overexposed: Optical densities from about 0.6 to 1.2 are the most suitable. As a general rule, you must be able to see through the exposed region fairly easily. (Overexposed films will usually give estimates for the radiation slice width which are too large; In this case, radiation intensities significantly less than one-half the peak value will still produce densities which are perceived as dark and will be included within the visually estimated thickness.)

A second, more accurate, method of dose profile analysis utilizes a microdensitometer for determining a density profile for each exposed strip. This density profile can then be changed into a relative dose profile using the characteristic density vs. exposure curve for TL film. Since this curve will vary with processing conditions, greatest accuracy is obtained by measuring it at the same time as the first dose profile test. Attach a Ready-Pack to the top of the RMI phantom. A flexible lead sheet must be placed directly under the film pack. Place a second lead sheet over the top and one side of the film. Using an mAs technique of about one-fifth of that used to expose the first dose profile test film, make six exposures without incrementing the table. After each exposure, move the top lead sheet slightly so that the exposed film will show six stepped densities produced by 1, 2, 3, 4, 5, and 6 times the base exposure. From this film you can obtain the characteristic curve. When using the microdensitometer method, the dose profile test film must be exposed to a much higher density than was required for the visual inspection method. Densities above 2.0 are suggested to keep most of the dose profile away from the "toe" of the characteristic curve.

The above methods give a radiation slice thickness or dose profile measured at the phantom's surface. The radiation slice thickness should be the same as the image slice width so that the full thickness that is being irradiated is also being imaged, without wasting radiation. To compare the <u>radiation</u> and <u>image</u> slice thicknesses. the radiation slice thickness must be <u>geometrically</u> referred to the isocenter of the scanner (or, equivalently, the center of the phantom since the phantom was centered in the scan field). To do this the radiation slice thickness (measured at the phantom's surface) must be multiplied by the following factor:

$$C = \frac{L}{L-r}$$

Where L is the distance between the iso-center and the x-ray focal spot and r is the radius of the phantom. This same factor must be used to correct the distance scale of the measured dose profile.

16. SPECIAL TESTS

A. keV Determination

By scanning materials of differing effective atomic number and determining their imaged CT numbers it is possible to measure the effective photon energy of the CT x-ray beam (Ref. 44). The liquids employed for this test can be scanned in the RMI PLASTIC TUBE (#460-023A). The PLASTIC TUBE has an angled filling neck which keeps bubbles outside of the scanned area when the tube is filled with liquid. The PLASTIC TUBE fits into the KEV LIQUID Insert (Part #460-023).

B. CT Number Measurements of Test Materials

The KEV LIQUID Insert with PLASTIC TUBE can, of course, be used to measure the CT number of any liquid. Solid test inserts can be cast or machined for fit of almost any dosimeter or solid material for CT scanning.

C. Image Sharpness

The WEDGE Insert (Part #460-022) contains a wedge of material with a nominal 10% contrast to its surroundings. The wedge tapers (See Fig. 38) to a very thin edge; the length of the wedge visible in the image is a function of the image quality.

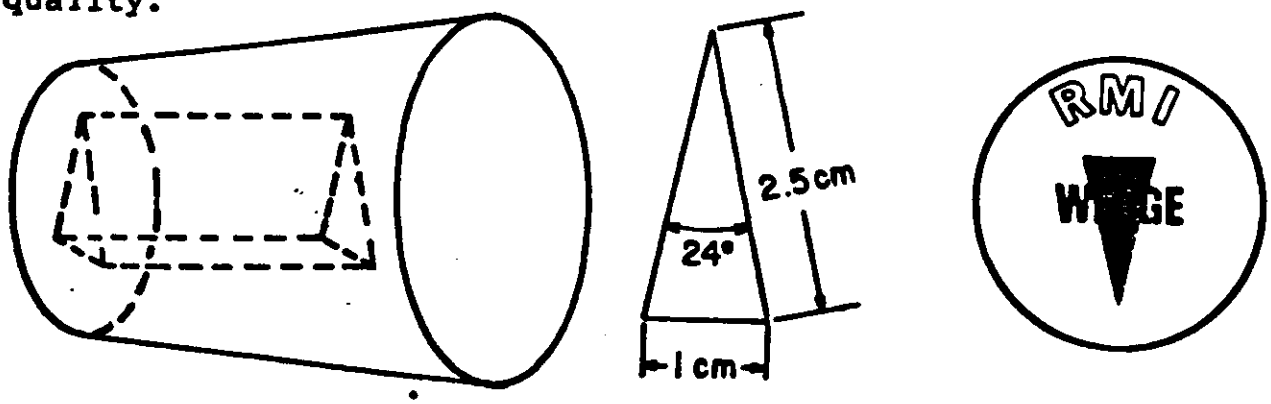


Fig. 38. Diagram of the WEDGE Insert, Part No. 460-022 and label.

References:

Text References and General References on Computed Tomography, Phantoms, and Quality Assurance.

1. Aronow, S. "Studies on Physical Phantoms in the EMI Scanner", Journal of Canadian Association of Radiologists, 27 (3): 149-156, Sept. 1976.

2. Bellon, E.M., F.D. Miraldi, and E.J. Weisen, "Performance Evaluation of Computed Tomography Scanners Using a Phantom Model", American Journal of Radiography, 132(3):345-352, March 1979.

3. Bischof, C.J. and J.C. Ehrhardt, "Modulation Transfer Function of the EMI CT Head Scanner", Medical Physics 4(2):163-167, Mar/Apr 1977.

4. Brooks, R.A. and G. DiChiro, "Theory of Image Reconstructionn in Computed Tomography", Radiology, 117(3):561-572, Dec. 1975.

5. Brooks R.A. and G. DiChiro, "Principles of Computer Assisted Tomographer (CAT) in Radiographic and Radioisotopic Imaging", Physicists in Medicine and Biology 21(5):689-732, Sept. 1976.

6. Chesler, D.A., S.J. Riederer, and N.J. Pelc, "Noise Due to Photon Counting Statistics in Computed X-Ray Tomography", Journal of Computer Assisted Tomography. 1(1): 64-74, Jan. 1977.

7. Chew, E., G.H. Weiss, R.A. Broolks, G. DiChiro, "Effect of CT Noise on Detectability of Test Objects", American Journal of Raidography, 131(4):681-685, Oct. 1978.

8. Christensen, E.E., T.S. Curry III, and J.E. Dowdey, An Introduction to the Physics of Diagnostic Radiology, Chapter 24: "Computed Tomography" P. 329-360, Lea & Febiger, 1978.

9. Cohen, G. and F.A. DiBianca, "The Use of Contrast-Detail-Dose Evaluation of Image Quality in a Computed Tomographic Scanner", Journal of Computer Assisted Tomography, 3(2):189-195, Apr. 1979.

10. Constantinou, C., F.H. Attix, B.R. Paliwal, "A "Solid Water" Phantom Material for Radiotherapy X-Ray and X-Ray Beam Calibration", Medical Physics, 9, (3). May/June 1982.

11. Gamberelli, J., G. Gverinel, L. Chevrot, and M. Mattei, Computerized Axial Tomography, Springer-Verlag, 1977.

12. General Electric, Introduction to Computed Tomography, General Electric Company, Medical Systems Division, Milwaukee, WI 53201.

13. General Electric, CT/T Technology Continuum Update: The Significance of Dose and Sensitivity Profiles, General Electric Publication 4998, 1979.
14. Gordon, R., G.T. Herman, and S.A. Johnson, "Image Reconstruction from Projections", Scientific American, 233(4):56-68, Oct. 1975.

15. Hanson, K.M. "Detectability in Computed Tomographic Images", Medical Physics, 6(5):441-451, Sept/Oct 1979.

16. Hospital Physicists Association Diagnostic Radiology Topic Group, "Measurement of the Performance Characteristics of Diagnostic X-Ray Systems Used in Medicine: Part III, The Physical Specification of Computed Tomography X-Ray Scanners: Measurement and Use or the Associated Performance Parameters", The Hospital Physicists(1) Association, 47 Belgrave Square, London (1981).

17. Johns, H.E., J. Battista, M.J. Bronskill, R. Brooks, A. Fenster, and M. Yaffe, "Physics of CT Scanners: Principles and Problems", International Journal of Radiation Oncology, Biology, and Physics, 3:45-51, 1977.

18. Jucius, R.A., and G.X. Kambic, "Radiation Dosimetry in Computed Tomography (CT)", SPIE Vol. 127, Optical Instrumentation in Medicine. VI:286-295, 1977.

19. Judy, P.F., "The Line Spread Function and Modulation Transfer Function of a Computed Tomographic Scanner", Medical Physics, 3(4):233-236, Jul/Aug 1976.

20. Judy, P., S. Balter, D. Bassano, E.C. McCullough, J. T. Payne, and L. Rothenberg. Phantoms for performance Evaluation and Quality Assurance of CT Scanners, AAPM Report #1, American Association of Physicists in Medicine, 1977.

21. Kak, A.C., "Computerized Tomography with X-Ray, Emission, and Ultrasound Sources", Proceedings of the IEEE, 67 (9):1245-1279, Sept. 1979.

22. McCullough, E.C., H.L. Baker, O.W. Houser, and D.F. Reese, "An Evaluation of the Quantitative and Radiation Features of a Scanning X-Ray Transverse Axial Tomograph: The EMI Scanner",Radiology, III(3):709-715, June 1974.

23. McCullough, E.C., "Photon Attenuation in Computed Tomography", Medical Physics, 2(6):307-320, Nov/Dec 1975.

24. McCullough, E.C., J.T. Payne, H.L. Baker, et al. "Performance Evaluation and Quality Assurance of Computed Tomography Scanners with Illustrations from the EMI, ACTA, and Delta Scanners", Radiology, 120(1):173-188, July 1976.

25. McCullough, E.C., and J.T. Payne, "X-Ray Transmission Computed Tomography", Medical Physics, 4(2):85-98, Mar/Apr 1977.

26. McCullough, E.C., "Factors Affecting the Use of Quantitative Information from a CT Scanner", Radiology 124910:99-107 July 1977
27. McCullough, E.C. and J.T. Payne, "Patient Dosage in Computed Tomography", Radiology, 129(2):457-464, Nov. 1978.

28. McCullough, E.C., "Specifying and Evaluating the Performance of Computed Tomography (CT) Scanners", Medical Physcis 1(4):291-296. Jul/Aug 1980.

29. Maravilla, K.R. and M.S. Pastel, "Technical Aspects of CT Scanning", Computerized Tomography, 2(3):127-144, 1978.

30. Millner, M.R., W.H. Payne, R.G. Waggener, W.D. McDavid, M.J. Dennis and V.J. Sank, "Determination of Effective Energies in CT Calibration", Medical Physics, 5(6):543-545, Nov/Dec 1978.

31.Nalcioglu, O., R.Y. Lou, and M.G. Yalcintas, "X-Ray Filtration, Noise, and Artifacts in Computerized Tomography", SPIE Vol. 152, Recent and Future Developments in Medical Imaging:26-32, 1978.

32. Phelps, M.E., M.H. Gado, and E.J. Hoffman, "Correlation of Effective Atomic Number and Electron Density with Attenuation Coefficients Measured with Polychromatic X-Rays", Radiology, 117 (3):585-588, Dec. 1975.

33. Riederer, S.J., N.J. Pelc, and D.A. Chesler, "The Noise Power Spectrum in Computed X-Ray Tomography", Physics in Medicine and Biology, 23 (3):446-454, May 1978.

34. Scudder, H.J. "Introduction to Computer Aided Tomography", Proceedings of the IEEE, 66 (6):628-637, July 1978.

35. Shepp, L.A., S.K. Hillal, and R.A. Schulz, "The Tuning Fork Artifact in Computed X-Ray Tomography", Physics in Medicine and Biology, 23 (3):446-454, May 1978.

36. Speller, R.D., D.R. White, R. Veerappan, A.J. Coleman, and P.E. Waller, "A Survey of 29 EMI CT Machines in Britains", British Journal of Radiology, 54, (639):232-240, Mar. 1981.

37. Speller, R.D., D.R. White, C.K. Showalter, et al., "An Evaluation of CT Systems from Ten Manufacturers', British Journal of Radiology, 64(648):1053-1061, Dec. 1981.

38. Swindell, W. and H.H. Barrett, "Computerized Tomography: Taking Sectional X-Rays", Physics Today, 30 (12)32-41, Dec. 1977.

39. Ter-Pogossian, M.M., et al. (ed.), Reconstruction Tomography in Diagnostic Radiology and Nuclear Medicine, University Park Press, 1977.

40. Ter Pogossian, M.M., "Basic Principles of Computed Axial Tomography", Seminars in Nuclear Medicine, 7 (2):109-127, Apr. 1977.

41. White, D.R., "The Formulation of Tissue Substitute Materials using Basic Interaction Data", Physics in Medicine and Biology, 22 (5):889-899, Sept. 1977.

42. White, D.R., R.J. Martin, and R. Darlison, "Epoxy Resin Based Tissue Substitutes", British Journal of Radiology, 40 (599):814-821, Nov. 1977.

43. White, D.R., "Tissue Substitutes in Experimental Radiation Physics", Medical Physics, 5 (6):467-479, Nov/Dec 1978.

44. White, D.R. and R.D. Speller, "The Measurement of Effective Photon Energy and "Linearity" in Computed Tomography", British Journal of Radiology, 53 (625):5-11, Jan. 1980.

45. White, D.R., R.D. Speller, and P.M. Taylor, "Evaluating Peformance Characteristics in Computerized Tomography", British Journal of Radiology, 54 (639):221-231, Mar. 1981.

46. Zatz, L.M., "The Effect of kVp Level on EMI Values", Radiology, 119 (3):683-688, June 1976.

47. Zatz, L.M. and R.E. Alvarez, "An Inaccuracy in Computed Tomography: The Energy Dependence of CT Values", Radiology, 124 (1):91-97, July 1977.

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