AAPM REPORT NO. 39

SPECIFICATION AND ACCEPTANCE TESTING OF COMPUTED TOMOGRAPHY SCANNERS

Published for the
American Association of Physicists in Medicine
by the American Institute of Physics

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Report of Task Group 2
Diagnostic X-Ray Imaging Committee

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May 1993

Published for the
American Association of Physicists in Medicine
by the American Institute of Physics

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Further copies of this report may be obtained from:
American Association of Physicists in Medicine
335 East 45th Street
New York, NY 10017

International Standard Book Number: 1-56396-230-6
International Standard Serial Number: 0271-7344

Published by the American Institute of Physics, Inc.
335 East 45th Street, New York, NY 10017

Printed in the United States of America

Preface

The Computed Tomography (CT) Scanner Task Group was formed to provide a reference document of suggested procedures for acceptance testing of CT systems for clinical medical physicists. The AAPM first addressed performance testing of CT systems with the publication of AAPM Report No. 1, “Phantoms for Performance Evaluation and Quality Assurance of CT Scanners”. Since publication of Report No. 1 in 1977, CT system designs have evolved and research has improved our understanding of principles and limitations of CT systems. These changes in technology and our understanding of them prompted the formation of this task group.

This report is intended to be a source of acceptance testing procedures, as well as an educational tool for those medical physicists whose primary responsibilities lie in areas other than diagnostic imaging. In addition to acceptance testing procedures, the related issues of specification writing and radiation shielding design are also discussed.

I. Introduction

The medical physicist can perform a critical role throughout the selection, purchase and acceptance of any complex expensive medical imaging system, such as an x-ray computed tomography (CT) scanner. Early involvement in the writing of the bid request can ensure that the appropriate technical data are supplied by the vendors in a format permitting a reasonable comparison. The expertise of the medical physicist can be used to analyze these specifications so that the nontechnical user can make an informed decision about which equipment best matches the clinical needs. Once the purchase decision is made, the physicist should be involved in site preparation and shielding design to ensure that radiation protection requirements are met and appropriate reports are submitted to regulatory authorities. Prior to acceptance of the system by the user, the physicist should make a thorough evaluation of the system performance and radiation characteristics to be certain that the system meets the specifications of the manufacturer and any additional requirements spelled out in the purchase document. Finally, the medical physicist can employ acceptance test data for x-ray machine inspection/registration reports required by some regulatory authorities.

The primary concerns of clinical medical physicists in diagnostic radiology are image quality, radiation dose, and radiation protection. This document, therefore, is concentrated on scanner characteristics which directly or indirectly influence one of these concerns. Emphasis is placed on performance characteristics that can be quantified by a medical physicist using widely available instruments and apparatus. Testing methods described in this document are suggestions and should not be interpreted as the only acceptable methods. Some procedures may require modification to address unique design features of a given scanner. The experienced physicist will find that different tests can be combined to make best use of limited scanner access time.

The acceptance test is a series of measurements performed by the physicist to verify that a CT system conforms to vendor technical specifications or to specifications mutually agreed upon by buyer and vendor. Often, a proviso for acceptance testing is written into the bid request which indicates who will do the testing, what tests will be performed, and what level of performance is acceptable to the buyer.

A. THE BID REQUEST

The bid request is a document submitted by the purchaser to potential CT system vendors which lists, in generic terms, system capabilities desired by the purchaser and any special business terms or conditions required. The physicist should ensure that sufficient technical detail is requested to allow a comparison of performance specifications. In this regard, a typical bid request should solicit the following:

- A full set of technical specifications detailing system performance in terms of spatial resolution, image noise and radiation dose for all standard scan settings.
- A set of typical multiformat images annotated with scan parameters used, and showing anatomy of clinical interest to the purchaser.
- A list of local purchasers to facilitate site visits.
- Typical architectural layouts indicating space, weight, electrical, and thermal requirements.
- Typical radiation exposure levels around the CT gantry for shielding design.
- Details of warranties, service contracts, and projected costs of contract and non-contract service over the first three years after warranty expiration.

Comparison of technical specifications can be difficult when supplied in dissimilar formats. In the interest of uniformity, a sample bid questionnaire is included as Appendix A.

B. EVALUATING NEEDS OF THE USER

Specific clinical needs to be determined include: patient load, type of patients examined (e.g., ambulatory, inpatient, adult, pediatric) and principal kinds of examinations anticipated. This provides insight into throughput, image archiving requirements and shielding design, and has an impact on system cost. More expensive systems may have more detectors, more pow-

erful computers, more flexible software, faster scan and reconstruction times, and better spatial resolution. Options such as separate diagnostic consoles, multiformat cameras, fast-cycle reconstructions, image archiving devices and special purpose software can add significantly to the total price. A careful technical evaluation can provide maximum value to the purchaser by matching the system configuration to actual needs.

C. EQUIPMENT SELECTION

After bids are received, technical differences between them can be simplified by tabulating technical specifications of each scanner for a "side-by-side" comparison. This should be accompanied by a non-technical commentary which relates relative importance of specifications to the user's specific needs. "Trade-offs" of performance levels should be pointed out, i.e., the dose and spatial resolution at optimal settings for low contrast discrimination, the dose and noise level at optimal spatial resolution settings, etc.

Prior to the final purchase decision, a site visit may be appropriate, particularly if new or unfamiliar systems arc considered. If a system employs significantly new technology, a few selected performance measurements by the physicist can prove useful.

Because CT systems are inherently complex and prone to failure, the availability and quality of service should weigh heavily in the purchase decision. Since service quality varies from one locale to the next, it is wise to confer with other local users prior to purchase. Warranties should be carefully reviewed and compared with regard to duration and components covered. Twelve-month warranties are typical for hardware and may or may not include x-ray tubes. Service contracts are commonly priced as a percentage of total system price (typically 8 to 12%).

D. SPECIFICATION WRITING

Once the purchase decision is made, the buyer may request that the physicist write performance specifications into the purchase agreement In some institutions, specifications are sent to vendors with the bid request. In any case, the following should be kept in mind:

> Specify only parameters with a measurably impact on some aspect of performance.

- Ensure that specifications are within the performance capability of the CT system in question.

- Be certain that a reliable, acceptable means of measurement is available for each specification.

- Either describe measurement methods or reference standard techniques to allow the manufacturer to know how a given parameter is to be tested.

Specifications may also address system reliability by requesting a guaranteed "up-time" of 95% for the warranty period. The proviso may also include penalties for excessive down-time, such as one month of warranty extension for each 1% of down-time beyond the specified 5% (see Appendix A).

II. Shielding Design

Like any radiation emitting system, consideration must be given to protection of personnel and others in the immediate vicinity of the CT system. Unfortunately, the characteristics of a CT system do not readily lend themselves to the standard shielding design methods of NCRP Report No. 49[i]. Two methods will be described: direct use of isoexposure contours, and an isotropic point source method adapted to that of NCRP Report No. 49.

Manufacturers can provide measured exposure levels in the gantry vicinity under average scan conditions, typically in the form of isoexposure contours, (see Figure 1). Isoexposure curves show measurements of exposure per scan in air while scanning a cylindrical phantom. Measured values typically represent scatter from the portion of the phantom within the scan field and do not suggest significant contributions from leakage or transmitted primary radiation. All barriers may be reasonably treated as secondary barriers.

Isoexposure contours may be used directly in shielding design, in which case two drawings are necessary, one in the horizontal plane (floor plan) and one in the vertical plane (elevation plan) scaled to the architectural plan. Exposure contour drawings can be superimposed on the architectural plan, and exposure levels either read directly or extrapolated (inverse square) to occupied areas, Exposure per scan values are then scaled to the total weekly workload of the scanner. A linear correction should be used, if slice width is different from that under which exposure contours were measured.

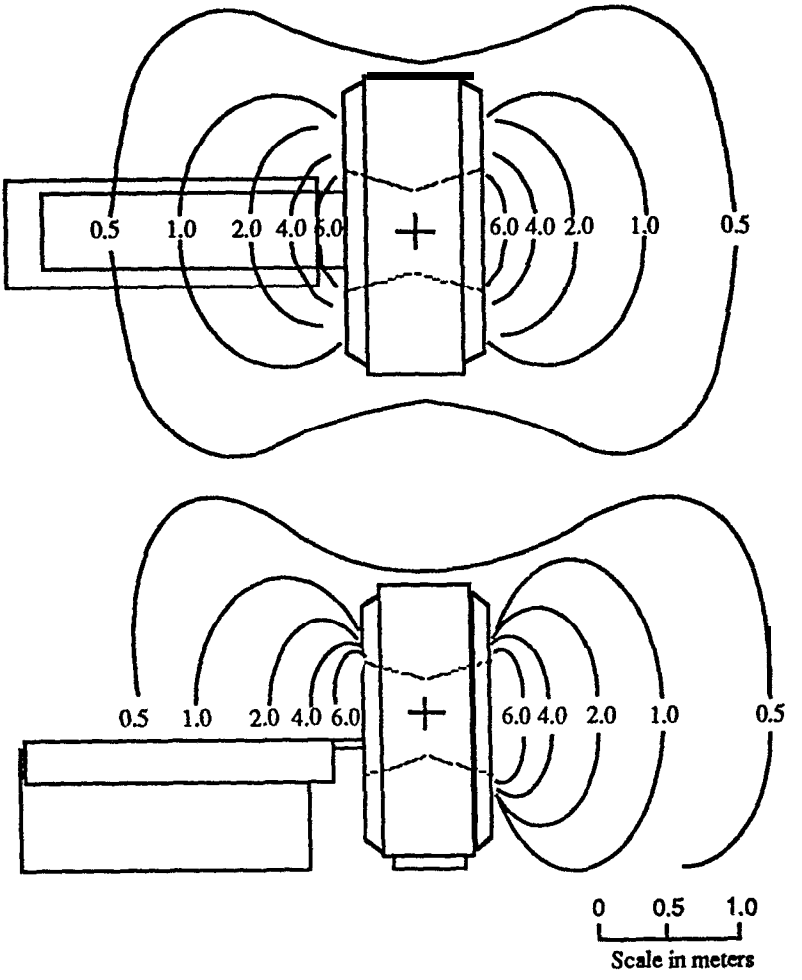

Figure 1: Typical isoexposure contours around CT gantry during scanning. Units are arbitrary, depending on scanner and scan conditions. (Generally available from scanner manufacturer).

An alternative shielding design method adapted from the standard methodology of NCRP No. 49 is described as follows:

For simplicity, secondary radiation is assumed to be isotropically emanating from the center of the gantry axis (the gantry isocenter). This ignores inherent shielding in the "shadow" of the gantry, hence is conservative in certain directions. The assumed exposure rate (X,) is the "worst case" measured value at one meter from the gantry isocenter in mGy/mA-min (air kerma), for the largest slice width on a body size phantom. As shown in Figure 1, the worst case exposure rate is at an angle of roughly 45° to the gantry axis. Some typical values of X_s normalized to one meter for both body and head scans are listed in Table 1. The required barrier transmission (B), at any location to be shielded, can then be obtained from the expression[1]:

$$B = \frac{P \cdot d^2}{X_s \cdot W \cdot T} \qquad [1]$$

where: P = weekly design exposure rate (mGy/week)
W = workload (mA-min/week)
T = occupancy factor
d = distance from isocenter to barrier location

Workload can be calculated in mA-minutes/week as:

$$W = \frac{R_S \cdot R_p \cdot Q}{12} \qquad [2]$$

where R_S = average number of scans per patient
R_p = average number of patients per 8 hour day
Q = average mAs per scan

Workloads for 120-130 kV machines may range from 5,000 to as high as 20,000 mA-min/week for heavily used systems, As in most shielding circumstances it is wise to be reasonably conservative, overestimating workloads to allow for changes in technique or patient load.

Given the barrier transmission (B) the required shielding thickness of a specific material can be computed from the formalism of Archer et a1.[2]:

$$t = \frac{1}{\alpha\gamma} \ln\left[\frac{B^{-\gamma} + \frac{\beta}{\alpha}}{1 + \frac{\beta}{\alpha}}\right] \qquad [3]$$

where t is the required barrier thickness (in mm for lead, cm for all other materials) of specific shielding material. B is the barrier transmission, while a, b, and g are fitting constants from Simpkin[3]. The coefficients appropriate for CT energies and beam qualities for a number of common structural and shielding materials are reproduced in Table 2. Use of these coefficients is illustrated in the following example:

Example:

A control area is 2.5 m from the gantry isocenter for a system where heavy use is anticipated at a tube potential of 140 kVp. Assuming a workload of 20,000 mA-min/wk, a (conservative) design air kerma of 0.1 mGy/wk, an occupancy factor of 1 and X_s (Table 1, GE 9800B, 42 cm phantom) of $3.8 \cdot 10^{-3}$ mGy/mA-min @ 1 m, the barrier transmission from Equation [1] is then:

$$B = \frac{(0.1)(2.5^2)}{(3.8 \cdot 10^{-3})(2.0 \cdot 10^{4})(1)}$$

$$B = 0.008$$

The coefficients for lead at 140 kVp from Table 2 are entered into Equation [3] as follows:

$$t = \frac{1}{(2.7667)(1.0281)} \ln\left[\frac{.008^{-1.0281} + \frac{10.855}{2.7667}}{1 + \frac{10.855}{2.7667}}\right]$$

$$t = 1.18 \text{ mm Pb}$$

Similarly, using the fitting coefficients from Table 2 for concrete, gypsum board, steel and plate glass results in required thicknesses of 10 cm, 27.0 cm, 1.01 cm and 12 cm, respectively. In practice, the method is considerably simplified by incorporating the mathematics of Equation [3] and the coefficients of Table 2 into a simple computer program or spreadsheet.

Table 1: Normalized scatter air kerma rates (X_s) in mGy/mA-min measured at 1 m for selected CT scanners, and conditions as indicated.

Scanner	Tube kVp	Phantom Diameter	Slice Width	X_s* mGy/mA-min
GE CT/T8800[a]	120	42 cm	10 mm	$1.9 \cdot 10^{-3}$
GE CT 9800B[a]	140	25 cm	10 mm	$2.9 \cdot 10^{-3}$
GE CT 9800B[a]	140	42 cm	10 mm	$3.8 \cdot 10^{-3}$
GE High Speed Advantage[a]	140	42 cm	10 mm	$2.6 \cdot 10^{-3}$
Elscint Elete 2400, Elect 2400[b]	NA	30 cm	10 mm	$1.7 \cdot 10^{-3}$
Picker IQ/PQ Series[c]	130	22 cm	10 mm	$2.6 \cdot 10^{-3}$
Siemens HiQ[d]	133	20 cm	10 mm	$1.7 \cdot 10^{-3}$
Siemens Somatom Plus[d]	137	20 cm	10 mm	$1.1 \cdot 10^{-3}$
Siemens Somatom AR.T[d]	NA	20 cm	10 mm	$2.9 \cdot 10^{-3}$
Toshiba TCT 300S[e]	120	NA	10 mm	$7.3 \cdot 10^{-4}$
Toshiba TCT 900S[e]	120	NA	10 mm	$1.5 \cdot 10^{-3}$

* Assumes 1 mGy = 0.1 R
a. GE Medical Systems, Milwaukee, Wisconsin
b. Elscint Inc., Hackensack, New Jersey
c. Picker International Inc., Cleveland, Ohio
d. Siemens Medical Engineering Group, Erlangen, Germany
e. Toshiba America Medical Systems, Tustin, California

Table 2: Shielding coefficients for computing thickness of shielding material required for Equation [3]. Coefficients yield thickness (t) in cm for all materials except lead, where t is in mm.

	α		β		γ	
kVp ⇒	120	140	120	140	120	140
Material						
Lead	2.7394	2.7667	13.229	10.8550	.9429	1.0281
Concrete	.3916	.3422	.6204	.4628	1.1792	1.1691
Gypsum	.1435	.1251	.4563	.2980	1.5952	1.4336
Steel	2.1157	2.0370	15.838	20.2740	.7270	1.2825
Plate Glass	.4050	.3627	.5567	.3807	1.5438	1.5712

III. Performance Evaluation

Performance of any scanner varies with operating conditions, and most systems provide more combinations than can be practically tested. Nevertheless, the acceptance test must acquire sufficient data to adequately characterize the system's performance. Before beginning, consider the following:

- Review clinical needs, scanner specifications and design principles to determine the most important performance variables to test, and include only those that could affect an important performance measurement.

- Devise a short list of 4-5 "standard" scanning conditions, either suggested by the manufacturer or by local users. Typically this would include: standard body, standard head, spine, pediatric body and pediatric head, as appropriate to the installation. All pertinent performance tests should be done under each standard condition to provide a performance baseline for comparison.

- Determine a few more scanning conditions that test performance limits, i.e., the fastest scan, the highest resolution scan, the lowest noise scan, etc., and relate results to baseline conditions.

Table 3: CT image quality and electromechanical acceptance tests.

Test	Priority	Comment
Localization Light Accuracy	Essential	Affects clinical utility. Can be combined w/ Table/Gantry Tilt test
Alignment of Table to Gantry	Optional	Useful if misalignment is suspected
Table Gantry Tilt	Optional	Implications in QCT
Slice Localization From Radiographic (Scout) Image	Altern. to Table/Gantry Tilt	Requires special test object
Table Incrementation	Essential	Implications in image quality and patient dose
Radiation Profile	Essential	Affects in patient dose
Sensitivity Profile	Essential	Measures accuracy of imaged slice width, implications in image quality
mAs Linearity	Optional	Some implications in noise scaling with dose
kVp Accuracy	Optional	Error can be problematic in QCT but is difficult to measure
Image Noise	Essential	Can substitute with Low Contrast Resolution test
Field Uniformity	Important	Implications in image quality and QCT
Scan Field Dependence (for QCT)	Optional	Implications for QCT with some scanner designs
High Contrast Res. - Resolution Pattern	Essential	Requires resolution test pattern from CT vendor or third party
High Contrast Resolution - MTF	Altern. to Resolution Pattern	More quantitative than resolution pattern test but requires MTF test tool and interpretation software
Low Contrast Resolution	Alternative to Image Noise	Requires special phantom, can replace noise measurements but requires observer performance test
Display and Hard Copy Image Quality	Essential	Requires SMPTE pattern as CT image data file or external video generator
Radiation Dose	Essential	Requires dosimetry phantoms and CT ion chamber or TLD system

- Determine both benefits and trade-offs of a given protocol, e.g., in the highest resolution scan determine the consequences in image noise, dose, presence of artifacts, etc.

Performance may also be influenced by the properties of the computer system, and by major electromechanical components, i.e., x-ray generator, x-ray tube, collimators, alignment lights, patient table, etc. Computer systems are evolving too rapidly for inclusion, but consideration of image quality is preceded by a discussion of direct and indirect performance measures of major electromechanical components. It is suggested that electromechanical tests precede those for image quality. Major discrepancies may influence image quality or dose and should be corrected prior to proceeding with other tests. An overview of the principal tests, their alternatives and priority are listed in Table 3, together with pertinent comments. For convenience, a listing of tools required for all tests is listed in Table 4.

Scanner performance and the interpretation of certain tests are influenced somewhat by scanner design. For the purposes of this document the following conventions are used (most but not all system configurations will fall into one of the following categories):

Second Generation Scanner: A mostly outmoded design with multiple detectors which operates in a translational motion during image acquisition, with rotation between projections. Generally employs a stationary anode x-ray tube and a constant potential x-ray generator.

Third Generation Scanner: The most common system design which uses a x-ray tube collimated to a fan beam directed at an arc-shaped array of detectors on the opposite side of the patient. Both x-ray tube and detector array rotate during image acquisition. Generally employs a rotating anode x-ray tube, and the generator may or may not be pulsed. Xenon gas detectors are only used with third generation designs. May employ slip rings to permit continuous rotation for spiral scanning. Some less expensive versions employ an asymmetric fan beam coupled to a detector array which samples half of the gantry aperture.

Fourth Generation Scanner: A scanner design which incorporates a ring of stationary detectors in which the x-ray tube rotates in a concentric path inside the detector ring. Generally employs a rotating anode x-ray tube and a constant potential generator. May employ slip rings to permit continuous rotation for spiral scanning. A variation on this design uses a nutating

motion of the detector ring so that the x-ray tube can be placed outside the detector ring. In this configuration, the detector ring tilts back away from the position of the x-ray tube as it rotates (so the beam doesn't intercept the outer ring). Opposite the position of the x-ray tube the detector ring tilts forward to intercept the beam transmitted through the patient.

Table 4: Equipment required for CT acceptance tests.

Tools Needed	Tests
Adhesive Tape	Most tests
Protractor accurate to 1°	Table/Gantry Tilt test, Slice Localization From Radiographic Image test
6-12" carpenter's spirit level with vertical and horizontal indicators	Table/Gantry Tilt test, others
3-4 sheets of 24 x 30 cm (10"x 12") prepackaged therapy localization film	Localization Light Accuracy, Table/Gantry Tilt and Radiation Profile tests
Film backing plate, i.e., 1 cm thick 25 x 31 cm acrylic plate	Localization Light Accuracy, Table/Gantry Tilt and Radiation Profile tests
Plumb bob	Alternative method for Alignment of Table to Gantry
Sharp pin or needle	Localization Light Accuracy test
Tape measure with mm ruling	For Table/Gantry Tilt, Alignment of Table to Gantry
Crossed-wire test object (Fig. B-4, Appendix B)	For optional Slice Localization from Radiographic Image test
10-30 cm ruler with mm ruling	For Table Incrementation test
Bent paper clip	For Table Incrementation test
Sensitivity profile test object (Figure B-1, Appendix B, CT or phantom mfr.)	For Sensitivity Profile test
Voltage divider, display unit, storage oscilloscope, hard copy device	For optional Invasive Generator tests
Noninvasive kV measuring device configured & calibrated for CT use	For optional Noninvasive Generator tests

Tools Needed	Tests
Test tube stand	To hold CT ion chamber for optional mA Linearity test
Head and body water or acrylic uniformity phantoms (Appendix B)	For Image Noise and Uniformity tests
Elliptical uniformity phantom with bone mass inserts (Figure B-2, Appendix B)	For optional part of Uniformity test and optional Scan Field Dependence (for QCT) test
Resolution pattern test object (CT or phantom manufacturer)	High Contrast Resolution - Resolution Pattern test
MTF test object (Figure B-4, Appendix B, CT or phantom mfr.)	For High Contrast Resolution - MTF test
Rose pattern phantom (Figure B-5, Appendix B or phantom mfr.)	Required for optional Low Contrast Resolution test
SMPTE pattern on CT image data file	Display and Hard Copy Image test
X-ray film densitometer	Display and Hard Copy Image test
Electrometer with 10 cm CT ion chamber	For standard ion chamber dosimetry method, and mA linearity test
32 cm body and 16 cm head dosimetry phantoms, optional 8 cm infant phantom (dosimeter and phantom mfrs.)	Required for dosimetry measurements
LiF TLD-100 ribbons, readout system and TLD holders (Figure B-6, Appendix B)	For optional TLD dosimetry method
Dosimetry alignment rod (Figure B-7, Appendix B, dosimeter and phantom mfrs.)	Required for dosimetry measurements

A. PERFORMANCE OF ELECTROMECHANICAL COMPONENTS

CT scanners incorporate specialized x-ray generators and a number of electromechanical components designed to localize and define the scan plane to the patient's anatomy. This category includes gantry and patient table motions, scan alignment lights and x-ray beam collimators. Since the x-ray generator and electromechanical components are subject to miscalibration, mechanical misalignment or other problems, careful testing is required.

1. Scan Localization Light Accuracy

Purpose: To test congruence of scan localization light and scan plane.

Patient anatomy to be scanned is often defined by scan alignment lights. Alignment lights may be located within the gantry at the slice plane, outside the gantry at a reference distance from the scan plane, or both. If both internal and external alignment lights are supplied, and are independently aligned, both should be tested.

Tools needed: Sheet of therapy localization film, film backing plate, adhesive tape and sharp pin or needle.

Method:

- Tape film to backing plate with film edges aligned parallel to plate edges. Tape plate in position oriented vertically along long axis of table and raise table to head scan position. If both internal and external alignment lights are provided, position plate so that both lights are visible on film surface (if possible).

- Turn on internal alignment light and mark light location on film by piercing film pack with pin at several points along middle of illuminated line. Repeat for external light, but put unique pattern of pin pricks near each illuminated line for later identification.

- Expose film at inner light location using narrowest slice setting at ~50-100 mAs, 120-140 kVp. For external light, increment table to light position under software control and repeat scan.

- For time savings, combine this test using same film for Gantry/table Tilt test (Section III.A.3); use same setup but fresh film for Radiation Profile (Section III.A.6.a).

Interpretation: Measure alignment error from holes in processed film to midpoint of radiation field. In absence of manufacturer's specifications, error should not exceed ±2.0 mm. Accuracy of external light systems depends on both table incrementation accuracy and light alignment, since they are designed to indicate a plane a specific distance from the scan plane.

2. Alignment of Table to Gantry

Purpose: To ensure that long axis of table is horizontally aligned with a vertical line through the scanner rotational axis.

Rationale: Misalignment can cause image artifacts with large patients if patient extends out of the sampled area.

Tools needed: Tape measure and adhesive tape. Optional - a plumb bob (string with a pointed weight on the end).

Method:

- Place piece of tape on middle of tabletop, locate table midline with ruler and mark on tape.
- With table and gantry untilted, extend tabletop into gantry to tape mark.
- Locate vertical midline of gantry opening with tape measure across largest horizontal diameter or alternatively hang plumb bob from top midpoint of gantry opening. Mark gantry midline on adhesive tape on tabletop.

Interpretation: Measure horizontal inaccuracy from marks on adhesive tape. In absence of manufacturer's specifications, gantry midline should be within ±5 mm of table midline.

3. Table/Gantry Tilt

Purpose: To determine accuracy of tilt indicators and to ensure that specified tilt can be accomplished under clinical conditions. (See alternative Slice Localization From Radiographic (Scout) Image test).

Rationale: Tilt indicators on gantry or table may not correspond to actual tilt or to that indicated on computer display. Some scanners may collide with the patient at extremes of tilt under some clinical conditions. Most scanners accomplish nonorthogonal scan planes by tilting the gantry, but some angle the table. Most table-tilting systems angle the table relative to the horizontal plane; however, pivoting of the table relative to the vertical plane is also seen.

Tools needed: Carpenter's spirit level, protractor, adhesive tape, sheet of therapy localization film, film backing plate, and tape measure with millimeter ruling.

Method:

- Tape film centered to side of acrylic plate, with film edge carefully aligned to edge of plate.
- Lower table to body scan position.
- For systems which tilt the gantry or table in vertical plane: Align plate with long axis of table, parallel to vertical plane (verify with spirit level). Fix in position with tape extending from table edge to table edge across top of plate.
- For tables tilting in horizontal plane: Place plate flat on tabletop with plate edge aligned to table long axis; tape into position.
- Move tabletop into gantry opening: center plate to alignment light.
- Expose the film at inner light location using narrowest slice setting at ~ 50-100 mAs, 120-140 kVp.
- Tilt gantry (table) to one extreme from operators console: record angle indicated on gantry (table) and operators console, repeat exposure.
- Measure gantry clearance from closest point on gantry to table midline.
- Tilt to opposite extreme, record indicated angles, repeat exposure and clearance measurement.

Interpretation: Measure tilt angles between density lines on processed film using protractor; express angles relative to density line from untilted position. In absence of manufacturer's specifications, tilt inaccuracies should be less than 3°; consoIe and gantry angle indicators, however, should agree exactly. Ideally, gantry clearance at the extremes of tilt should permit the scanning of at least a 30 cm patient.

4. Slice Localization From Radiographic (Scout) Image

Purpose: To determine correspondence of localization image parameters with actual slice position and angle (see alternative Table Gantry Tilt test).

Rationale: Errors can be problematic in quantitative CT work in the spine. Using an initial localization image of the patient, the measurement slice must be accurately positioned in the center of the vertebral body at an angle parallel to the end-plates. Errors in axial position or angle can cause inclusion of the dense end-plates in the slice intended to measure only trabecular bone in the center region.

Tools needed: Test object with two 45° wire cross-hairs (see Figure B-3, Appendix B) and protractor.

Method:

- Incline test object disk at an angle to be tested (verify with protractor) and position it on the scanner table such that the wire cross-hairs are visible in a lateral view. Take lateral localization radiograph. Under computer control define slice position and slice angle to pass through intersections of the two cross-hairs.

- Scan phantom with narrowest slice and highest resolution settings. The correctly positioned image should show two dense objects in a vertical orientation, corresponding to intersections of the two sets of wires.

- If one or both cross hairs show the wires vertically separated from each other, repeat scan at one mm spacing and same tilt angle on either side of the first slice. This will permit determination of the displacement and angle error.

- Repeat procedure for an arbitrary inclination angle in the opposite direction.

Interpretation: Because the two wires in each cross-hair are. 45° apart and one wire is roughly perpendicular to the scan plane, the distance between the two in the image gives the table position error. Combining the position errors of both cross-hairs provides the gantry angulation error. This test

should be repeated periodically since indicators for table axial position, height and tilt can change over time.

5. Table Incrementation

Purpose: To determine accuracy and reproducibility of longitudinal table motion.

Rationale: Under computer control from operators console the patient table must be able to accurately and reproducibly move the patient to any indicated position in the scan field. Accuracy is critical since it determines relative locations of image sections and influences the multi-scan dose.

Tools needed: 10-30 cm ruler with mm ruling, bent paper clip for a pointer and adhesive tape.

Method:

- Tape ruler along tabletop edge, near foot end. Tape pointer (end of paper clip) on table frame opposite midpoint of ruler with pointer directed at ruler midpoint. Zero table position.
- Load table with 70-80 kg (150-180 lb), c.g., have assistant lie on table.
- From control console, note indicated table position. Under computer control move tabletop 300-500 mm in one direction, and back to original position.
- Record relative distance from starting position (if any) indicated by pointer and ruler.
- Repeat measurement twice more with same direction of table motion, then three times in opposite direction.

Interpretation: In the absence of specifications, both the standard deviation and the average error should be less than 3 mm, Note whether error is systematic (in one direction) or random.

6. Collimation

A CT scan samples a thin slab of tissue; ideally only the imaged slab is irradiated, and its thickness equals the selected slice width. Inaccuracies arise from scanner &sign limitations and calibration of collimator settings. Most systems collimate the z axis dimension of the x-ray beam at both the source (patient collimation) and detectors (postpatient collimation). Width of the imaged slice (from sensitivity profile) is determined by both sets of collimators. Width of the tissue irradiated (from radiation profile) is a function only of the prepatient collimator. Measurements of radiation and sensitivity profiles are best interpreted together.

a. Radiation Profile Width

Purpose: To determine accuracy of prepatient collimator settings. (See Section III. C for dosimetry).

Rationale: The radiation profile describes the distribution of radiation energy within a continuous medium along a line parallel to the scanner rotational axis. Like any x-ray field, radiation energy peaks in the center of the field and falls off diffusely at the edges. Edge fall-off is due to scatter within the field and collimator penumbra. The magnitude of penumbra depends on the z dimension of the focal spot (parallel to scan rotation axis) and focus-collimator-patient geometry. The z axis dimensions for rotating anode x-ray tube foci are typically less than 2 mm, but may be much larger in a stationary anode tube (rarely used in modem systems). Penumbra is typically much worse in the latter.

Tools needed: Sheet of therapy localization film, film backing plate and adhesive tape.

Method:

- Tape film pack to backing plate (Can use same film as Localizer Light test).
- Place plate on table with neither table nor gantry tilted. Raise table so that plate is at gantry center of rotation.
- Starting at about 2 cm from end of plate and about 2 cm apart, take one slice at each slice width, using 50-100 mAs @ 120-140 kVp

(density on processed film should be between 0.8 and 1.5). In systems with dual foci, use the smaller focus.

Interpretation: Measure width of density profile on processed film with a caliper. Alternatively, to improve measurement accuracy, project film image onto a rigid surface with overhead transparency projector and a transparent ruler for scaling. Measure each density profile at full width half maximum (FWHM), i.e., between penumbral midpoints on each field edge. Average at least 3 measurements on each profile. Density profiles may also be measured with a scanning microdensitometer.

b. Sensitivity Profile Width

Purpose: To determine actual width of the imaged slice.

Rationale: The sensitivity profile measures system response to an attenuating impulse as a function of z axis position, (through the slice plane). The sensitivity profile is a function of pre- and postpatient collimation, and appears as a blurred square wave which at FWHM should correspond to the nominal slice width. Because effective beam width varies across the scan plane, the sensitivity profile will vary somewhat depending on source-collimator geometry, and scan angle The sensitivity profile should be measured at a specific radius, with respect to the center of rotation, on the midpoint of the scan arc for <360° scanners, and ideally within a few cm of the radial location of radiation profile measurement.

Tools needed: Inclined metal ramp test object (Figure B-l, Appendix B) or similar device supplied by CT manufacturer.

Measurement: The test object (Figure B-l Appendix B) uses a pair of 0.1 mm copper foils at an inclination ratio of 5:1 (θ = 11.31°) to the scan plane, embedded in acrylic. The width of the sensitivity profile is:

$$W_s = t \cdot \tan(\theta) \qquad [4]$$

where t is the measured image width (at FWHM) of the inclined ramp in the slice plane. Test objects like that in Figure B-l have been shown to be accurate for slice widths from 1 to 10 mm[4].

- Center test object to gantry axis and align to scan alignment light. Be certain that neither table nor gantry are tilted (misalignment causes error in 0).

- Without moving tabletop or gantry, scan test object at each slice setting using a low noise technique and a standard head convolution kernel.

Measure sensitivity profile from scan images with one of the three following methods:

- In each image, determine average pixel values of metal ramp and acrylic background.

- get window width at <10 CT. units or minimum width setting; set window level halfway between measured average value for ramp and acrylic background.

If scanner has distance utility program with precision to 0.1 mm:

- **Measure** width of ramp images using the distance utility function, average, then multiply by tan θ.

If distance utility is unavailable or insufficiently accurate:

- Record each image on film using specified window settings, and large image magnification.

- Determine geometric correction factor as ratio of test object diameter to image diameter.

- Measure width of ramp on each film image, correct geometrically, then multiply result by tan θ.

If pixel values can be printed out or ported to a microcomputer:

- Obtain pixel values within a region of interest (ROI) over the ramp image. Record pixel values across image of each ramp and plot as a function of pixel spacing (correct for angulation of matrix with

ramp image if necessary). Measure sensitivity profiles at FWHM of each plot.

Interpretation of Radiation and Sensitivity Profiles: Test results should be interpreted with care since incongruence of radiation and sensitivity profiles may indicate collimator misalignment but may also be due to design constraints.

In some CT designs, slice width is varied only with prepatient collimation; postpatient collimation is fixed at the widest slice setting. In these cases, the sensitivity profile should be congruent with the radiation profile only at the widest slice, but will be wider than radiation profiles at narrower settings. In other designs the prepatient collimator width for the narrowest slice is the same as in the next larger slice. This is done because prepatient collimation can't be narrower than the focal spot z dimension without sacrificing output. In one case, the same prepatient collimation is used for both 1 and 2 mm slice widths; only postpatient collimation is changed for the 1 mm. The sensitivity profile at the 1 mm setting would be -1 mm wide but the radiation profile would be -2 mm wide.

For scans in which both pre- and postpatient collimators are varied, radiation profiles should be slightly larger than the sensitivity profile, mainly because of field edge penumbra in the former. In absence of manufacturer's specifications, sensitivity profiles with rotating anode x-ray tubes should be within ±0.5 mm of the slice setting, and the radiation profile should be less than 1 mm in excess of the slice width. Stationary anode systems may require a tolerance of 2-3 mm or more in the radiation profile, due to greater penumbra

Within these limitations, results should be interpreted as follows: Excess in the radiation profile indicates prepatient collimation set too wide, with significant implications in patient dose. An excessively narrow dose profile indicates a misaligned or excessively narrow prepatient collimator setting. This condition is usually accompanied by a narrowed sensitivity profile. An excessively narrow sensitivity profile by itself may indicate a misaligned postpatient collimator. Narrowed sensitivity profiles cause increased quantum noise due to diminished photon collection. An excessively wide sensitivity profile has an opposite effect on noise, and will cause some loss of z-dimension resolution with effects in partial volume averaging and in multiplanar reconstructions.

7. The X-ray Generator

The x-ray generator is integral to the CT system, and in most newer systems was specifically designed for CT. Typical CT systems operate at relatively few tube potential settings since each requires different calibration data for image reconstruction. Tube currents range from 10-600 mA, with mAs values typically between 100-1000 mAs. Non-pulsed x-ray generators (in second, fourth and some third generation designs) allow tube current and exposure (scan) time to be varied independently, Pulsed generators (some third generation systems) may vary mAs by changing pulse width, tube current, or number of pulses per scan. Because non-pulsed scanners may incorporate a beam shutter to avoid spectral problems during rise and fall of the high voltage waveform, the high voltage duration may be longer than the exposure time. Generator measurements should be considered optional, to be done if miscalibration is suspected. Kilovoltage measurements are particularly difficult to perform in the clinical setting and are subject to significant sources of measurement error. Invasive measurements with a voltage divider should be done only with cooperation of a qualified service engineer. For completeness, both invasive and noninvasive generator tests are described. but indirect noninvasive measurements are preferred in most cases.

Purpose: To determine accuracy of tube potential and mAs settings.

Rationale: Miscalibration of tube potential can cause error in CT numbers, particularly problematic in quantitative CT work. Miscalibration of tube current often results in tube potential error as well. Second and fourth generation systems are generally intolerant of within-scan voltage fluctuations. For these scanner types, high frequency cyclical fluctuations (within projection acquisition) can cause Moire patterns or other severe artifacts. Low frequency variations mainly increase image noise. Third generation systems (pulsed or non-pulsed) are relatively tolerant of tube potential fluctuation, though uncorrected random inter-pulse variation will increase image noise.

Invasive Measurements: These measurements should only be done after appropriate training or with the assistance of a qualified service engineer. It is assumed that the physicist is familiar with safety precautions and proper use of voltage divider and oscilloscope (see Rossi et al.[5]). Conventional voltage dividers cannot be used with grounded anode generators, generators with non standard high voltage cables, or systems where the high voltage transformer rotates with the x-ray tube.

Tools needed: Noninvasive tube potential measuring device calibrated for CT use, or voltage divider with kV and mA display and 2-3 m high voltage extension cables (e.g., Dynalyzer -Varian Power Grid and X-Ray Tube Products, Salt Lake City UT). Note: long cables can perturb generator calibration. Include also analog storage or digital oscilloscope (for use with voltage divider), and Polaroid oscilloscope camera or screen printer.

Method:

- With scanner system turned off, and no power supplied to generator, insert voltage divider into secondary side of high voltage transformer, according to manufacturer's recommendations (discharge static charge in cables to ground before further handling). If generator is pulsed, measure on secondary side of pulsing unit.

- Measure both tube potential and current at each generator power level for each calibrated kV setting (different scan times at the same power level need not be tested). Note focus selected with dual focus x-ray tubes and record mA and kV waveforms with oscilloscope.

Interpretation: In absence of manufacturer's specifications, for both pulsed and non-pulsed generators, tube potential should be within ±2 kV of indicated values for all power levels, and tube current should be accurate to within +5% of indicated levels.

NoninvasiveMeasurements:

Measuring Tube Potential: Noninvasive devices may not work well in CT if detector sensitive region is too large for narrow CT beams. Be certain that the device is calibrated for the kV range and beam quality to be tested (consult device manufacturer) and is set for DC or three-phase wave forms. Hardening corrections may be necessary with some instruments to compensate for heavily filtered CT beams.

Tools needed: Noninvasive tube potential measuring device calibrated for CT use.

Method:

- From operators console, rotate tube to overhead (AP) position. Put table at lowest scan position, move tabletop into gantry opening and place kV sensor on table. Align detector(s) to scan alignment light. If instrument detector is large, place it at bottom of

gantry opening where the field size is greatest, with tabletop out of field

- Operate CT system in service mode with tube and tabletop stationary (consult service engineer).

- Set widest collimator setting and expose detector. Proceed only if instrument obtains a reading without error indication.

- Measure tube potential at each generator power level for each potential setting. Record which focus is selected with dual focus x-ray tubes.

mA Linearity: Without specialized instruments (i.e., Hall Effect probes), tube current must be inferred indirectly using a mAs linearity measurement For a constant tube potential and slice width, the integral exposure should be a linear function of mAs.

Tools needed: CT ion chamber, electrometer with integral exposure and time duration capability and test tube stand or equivalent to hold ion chamber.

Method:

- Put tabletop just outside of scan field: place test stand on tabletop.

- Put ion chamber (without any attenuator) on stand and center parallel to rotational gantry axis. (Centering is critical if scan angle is variable, or if system has asymmetric fan beam). Ensure that chamber sensitive volume is within slice dimensions (not critical with CT chamber) and no other attenuator (including tabletop) is in scan field

- Set widest available slice width.

- If the generator is not pulsed and electrometer can also measure exposure duration: measure integral exposure and scan duration for all charge settings at each tube potential. Ensure that other pertinent factors (e.g., beam filter) remain constant and note which focus is selected with dual focus tubes.

- If generator is pulsed or if electrometer does not measure exposure duration: measure integral exposure for all available mAs settings, at each tube potential, with other factors constant.

Interpretation: For each tube potential, calculate mGy/mAs (or mR/mAs) then determine coefficients of linearity relative to mean of all values. Significant deviation from linearity may indicate miscalibration of potential, current, or exposure time. If the coefficient of linearity of mGy/mAs between the mean of all values and any single value (absolute difference divided by sum) is greater than 0.05 then significant miscalibration may be present. If time values are obtained, determine if measured times correspond to scan time settings. Errors with constant potential generators can indicate shutter problems.

B. IMAGE QUALITY

A reconstructed CT image is essentially a map of energy weighted x-ray attenuation values in the scanned tissue slice. Its accuracy in a practical scanner is constrained by intrinsic physical limitations of the system design. Finite sampling of the image space imposes limits on object spatial frequencies reproduced in the image. Dose constraints, and limits on x-ray tube output and detector efficiency, cause statistical uncertainties in attenuation measurements. Superimposed on these arc random and systematic errors from a variety of sources. One may classify these discrepancies into the following categories.

- ◊ Random Uncertainties in CT Number (Image Noise)
- ◊ Systematic Errors in CT Number (Artifacts)
- ◊ Spatial Frequency Limits (Spatial Resolution)

Each category is discussed in the following sections, together with suggested testing methods.

1. Random Uncertainties in Pixel Value

A CT image of a uniform object reveals both systematic and random variations in pixel numbers about some mean value. The random component of this variation is pixel noise. Its effect on the image is to place a lower limit on the level of subject contrast which can be distinguished by the ob-

server. Pixel noise is a critical limiting factor in CT since much soft tissue detail is low contrast in nature. Assuming that digitization error is insignificant in modem scanners[6], total random pixel noise (Np) is given by:

$$N_p \approx \sqrt{N_e^2 + N_q^2} \qquad [5]$$

Electronic noise (N_e) arises as random variation in detector signal prior to digitization; quantum noise (N_q) is due to random variation in numbers of detected x-ray quanta. Electronic noise is thermal in origin and if the scanning system is operating properly, can be considered to be approximately constant in magnitude. Because N_e is practically independent of operating factors, its principal effect is to constrain the level of improvement available by reducing N_q. Electronic noise increases in relative importance when large numbers of photons are detected, e.g., thin patients, high dose techniques (because quantum noise is small and electronic noise remains roughly constant). In most clinical circumstances image noise is dominated by quantum noise: however, an observed increase in image noise can be due to detector problems causing a larger electronic noise component.

Quantum noise (Nq) arises from statistical uncertainty in the finite number of transmitted x-ray photons (n) collected in forming the image[7], i.e.:

$$N_q \propto n^{-1/2} \qquad [6]$$

One or more rays in each projection traverse each pixel locus in the reconstruction matrix. The number of x-ray quanta contributing to each pixel depends on: quanta collected per ray, ray spacing and total number of projections. For objects of fixed size and attenuation, N_q is approximately[8]:

$$N_q \propto \frac{1}{\sqrt{[w^3 \cdot h \cdot Q]}} \qquad [7]$$

where w is the spatial resolution element, h is the imaged slice width, and Q is the x-ray tube mAs. Altering slice width changes the effective detector cross-section and therefore the numbers of photons collected. A change in mAs results in a linear change in numbers of transmitted quanta. With small pixels and optimal sampling and reconstruction, "w" is approximated by the effective ray width A_{eff} (see Section III.C.3.a). Generally, any scan pa-

rameter which results in improved spatial resolution causes an increase in noise. Conversely, factors which degrade resolution improve noise.

The number of quanta collected also depends on patient transmission, a function of body part thickness and density, and beam quality. In a scan of a cylindrical uniform phantom, noise in the center of the image would be greater than at the periphery, due to the greater transmission of the latter. This effect is less pronounced in scanners using “shaped” or ‘bow-tie” x-ray beam filters, which reduce the intensity of off-axis rays, flattening the noise distribution in a uniform phantom[7].

A final important consideration is total detector efficiency, i.e., the fraction of x-ray energy incident on the detector element resulting in detector signal. Total detector efficiency is the product of absorption and geometric efficiencies of the detector element. Scintillation detectors can have absorption efficiencies of 95-99% over the usual range of CT kVp's, whereas xenon detector absorption efficiency is typically on the order of 60%[9]. Geometric efficiency is the fraction of the detector cross-section occupied by the detector, excluding "dead space" due to interdetector septa. Scanners with variable detector width (to improve resolution) generally sacrifice geometric efficiency. Total detection efficiencies of commercial CT systems vary between 30 and 85%. For images of comparable resolution and dose, N_q is inversely related to total detector efficiency.

image noise also has spatial frequency content, i.e., noise amplitude varies with spatial frequency. As noted by Reiderer et al.[10], CT image noise spectra differ from white noise spectra in that very low frequency noise is virtually absent. Of more practical clinical significance is the behavior of higher frequency noise, particularly within two octaves (a factor of four in frequency space) of the MTF limits, since these frequencies tend to be more visually objectionable[11]. Edge enhancement kernels and high resolution modes tend to enhance higher frequency noise as well as high frequency information, hence are generally unsuitable for clinical situations requiring detection of low contrast lesions.

Noise is typically measured as the standard deviation of pixel values within an ROI on a scan of a water phantom. The statistical sample included by the ROI may differ from the number of pixels if pixel size differs greatly from resolution limits. For very small FOV’s, reduced pixel spacing is achieved by interpolation, not by finer sampling; therefore, more pixels than samples are obtained. Alternatively, if pixel size is larger than about twice

the limiting resolution, more samples than pixels are obtained. The actual number of samples within the ROI can be approximated by:

$$\#S \approx \frac{Area_{ROI}}{A_{eff}^2} \qquad [8]$$

where $Area_{ROI}$ is the area of the ROI in mm^2 and A_{eff} is the width of the effective sampling aperture in mm (see Section III.C.3.a). A sample size of 25 is adequate in most cases and can be obtained with an ROI of ~1 cm^2 (assuming A_{eff} is <2 mm).

Noise Measurement

Purpose: To assess the level of noise under simulated clinical conditions and its variation with different scanning parameters.

Rationale: Noise limits the perceptibility of low contrast detail.

Tools needed: Head and body water phantoms (those supplied with scanner will usually suffice).

Method:

- Perform calibration scan according to manufacturer's recommendations.
- Center head phantom to the gantry opening on table or use phantom mounting bracket if available.
- Scan phantom with standard head scan parameters; save raw data
- Examine image with a narrow window for obvious artifacts (streaks or rings). Repeat air scan if necessary.
- Record standard deviation of pixel values from ROI centered on phantom image.
- Repeat with body water phantom using body scan parameters.

To fully characterize image noise with a given scanner, determine the effect of each of the following parameters from the same size ROI in the center of the phantom:

- mAs - Using the head phantom and all other conditions normalized for head scans, scan at lowest and highest mAs setting, record standard deviation from ROI. Ensure that resolution factors are constant.

- Tube Potential - If multiple kVp's are available and calibrated, perform head scan using manufacturer's head scan parameters at each setting. Ensure that resolution factors arc constant.

- Resolution enhancement scanning modes: With head phantom, perform a series of scans after varying factors which alter resolution, e.g., ray sampling, A_{eff}, but not convolution kernel or pixel size. If possible, vary only one factor at a time. Ensure that resolution is not pixel size limited in lowest resolution mode, e.g., pixel size £ $A_{\text{eff}}/2$)

- Pixel size - With saved raw data set from head scan, using largest reconstruction matrix and standard convolution kernel, reconstruct image with largest and smallest FOV. Repeat with other available matrix sizes.

- Convolution kernel - Using the same raw data set, reconstruct image with each available kernel, using standard head scan FOV and matrix size; repeat with saved body scan data set.

Interpretation: Noise measurements (N) should be expressed as a percent of the effective linear attenuation coefficient (μ_w) of water, and corrected for the scanner contrast scale[31]:

$$N = \frac{\sigma \cdot CS \cdot 100}{\mu_w} \qquad [9]$$

where σ is the measured standard deviation of pixel values in the ROI and CS is the contrast scale factor. CS is defined as the change in effective linear attenuation coefficient for a given change in CT number, for two known materials. Normally one material is water, and the other is acrylic (poly-

methyl methacrylate), the common construction material for CT phantoms. Other (low Z) reference materials may be used. The contrast scale is then[31]:

$$CS = \frac{\mu_m(E) - \mu_w(E)}{CT_m - CT_w} \qquad [10]$$

where $\mu_m(E)$ and $\mu_m(E)$ are the linear attenuation coefficients of the reference material and water, CT_m, and CT_w, are measured CT numbers of reference material and water. For convenience, Table 5 lists densities and mass attenuation coefficients for a selection of materials from 40 keV to 100 keV. The effective photon energy of a 120-125 kV CT x-ray beam is typically on the order of 70 keV. The effective energy can be obtained from a CT number linearity measurement (see references 27 and 38).

Normalized noise measurements should be compared to manufacturer's specifications, and should vary as predicted by Equation [7]. Note that some scanners alter A_{eff} by masking the detector aperture in high resolution modes, which sacrifices geometric efficiency, therefore noise will increase more than predicted by Equation [7]. Significant deviations from predicted values can indicate detector problems, source fluctuation, collimator misalignment, generator miscalibration, etc.

2. Systematic Uncertainties in Pixel Value

Any systematic alteration of pixel numbers from expected value is essentially artifactual in nature. Systematic variation may be due to system malfunction, or physical or design limits. Artifacts due to systematic errors vary in appearance depending on their source but due to the nature of rotational geometry and the back projection process, are exhibited mostly as rings or streaks. Artifacts may also appear as subtle spatial variations in pixel value, particularly troublesome in quantitative CT. Other than spatial uniformity measurements, (see below) no specific tests for artifacts are recommended. An overview of common CT artifacts is as follows:

Table 5: Densities (g/cm^3) and mass attenuation coefficients (g/cm^2) for polystyrene[12] ($(C_8H_8O_2)_n$). acrylic[12] ($(C_5H_8O_2)_n$). soft tissue[13], water[13] and bone[13]. Values were interpolated to 1 keV intervals using a log interpolation (ix., linear interpolation of $\log(\mu/p)$).

Photon Energy	Polystyrene	Acrylic	Soft Tissue*	Water	Bone**
Density →	1.06	1.19	1.03	1.00	1.92
40	0.244	0.249	0.268	0.262	0.666
41	0.240	0.245	0.264	0.258	0.637
42	0.237	0.242	0.259	0.254	0.608
43	0.233	0.238	0.255	0.250	0.582
44	0.230	0.235	0.251	0.246	0.556
45	0.227	0.231	0.247	0.242	0.531
46	0.223	0.228	0.243	0.238	0.508
47	0.220	0.224	0.239	0.234	0.486
48	0.217	0.221	0.235	0.230	0.464
49	0.214	0.218	0.231	0.227	0.444
50	0.210	0.215	0.227	0.223	0.424
51	0.209	0.213	0.225	0.221	0.412
52	0.207	0.211	0.223	0.219	0.400
53	0.205	0.209	0.220	0.217	0.388
54	0.203	0.207	0.218	0.215	0.376
55	0.201	0.205	0.216	0.213	0.365
56	0.200	0.204	0.214	0.211	0.355
57	0.198	0.202	0.212	0.209	0.344
58	0.196	0.200	0.210	0.207	0.334
59	0.195	0.198	0.208	0.205	0.325
60	0.193	0.197	0.206	0.203	0.315
61	0.192	0.196	0.205	0.202	0.310
62	0.191	0.195	0.204	0.201	0.304
63	0.190	0.194	0.203	0.200	0.299
64	0.189	0.193	0.201	0.199	0.294
65	0.188	0.192	0.200	0.198	0.289
66	0.187	0.191	0.199	0.196	0.284
67	0.186	0.190	0.198	0.195	0.279

Photon Energy	Polystyrene	Acrylic	Soft Tissue*	Water	Bone**
67	0.186	0.190	0.198	0.195	0.279
68	0.185	0.189	0.197	0.194	0.274
69	0.184	0.188	0.196	0.193	0.270
70	0.183	0.187	0.195	0.192	0.265
71	0.182	0.186	0.194	0.191	0.260
72	0.181	0.185	0.193	0.190	0.256
73	0.180	0.184	0.191	0.189	0.252
74	0.179	0.183	0.190	0.188	0.247
75	0.178	0.182	0.189	0.187	0.243
76	0.177	0.181	0.188	0.186	0.239
77	0.177	0.180	0.187	0.185	0.235
78	0.176	0.179	0.186	0.184	0.231
79	0.175	0.178	0.185	0.183	0.227
80	0.174	0.177	0.184	0.182	0.223
81	0.173	0.177	0.183	0.181	0.221
82	0.173	0.176	0.183	0.181	0.219
83	0.172	0.175	0.182	0.180	0.217
84	0.171	0.175	0.181	0.179	0.215
85	0.171	0.174	0.181	0.179	0.213
86	0.170	0.174	0.180	0.178	0.211
87	0.170	0.173	0.179	0.177	0.209
88	0.169	0.172	0.179	0.177	0.207
89	0.168	0.172	0.178	0.176	0.206
90	0.168	0.171	0.177	0.175	0.204
91	0.167	0.171	0.177	0.175	0.202
92	0.167	0.170	0.176	0.174	0.200
93	0.166	0.169	0.175	0.173	0.198
94	0.165	0.169	0.175	0.173	0.196
95	0.165	0.168	0.174	0.172	0.195
96	0.164	0.168	0.174	0.172	0.193
97	0.164	0.167	0.173	0.171	0.191
98	0.163	0.166	0.172	0.170	0.189
99	0.163	0.166	0.172	0.170	0.188
100	0.162	0.165	0.171	0.169	0.186

*Average adult male soft tissue ICRU 44 formulation[13]

**Skeleton- cortical bone adult[13]

a. Image Artifacts

According to Joseph[14], artifacts are best classified by the nature of the error ma& in the scanning process and may be categorized as geometric errors, algorithm effects, or attenuation measurement errors. The following discussion is restricted to artifacts that can occur in clinical situations.

1) Geometric Artifacts

This broad category includes artifacts due to errors from inadequate ray sampling, and inconsistencies in the spatial position of rays or projections.

a) Aliasing

Aliasing usually appears as faint streaks radiating from bone edges or other high frequency objects, and may reinforce from several sources to produce a Moire pattern. When ray spacing is less than $l/2\ A_{eff}$ (the Nyquist criterion), the highest frequencies in the scanned object are under-sampled, so that they reappear (alias) as lower frequencies. Some aliasing is often seen in third generation minimum rotation (<360°) scans. It can be greatly diminished by use of scan modes which reduce ray spacing within projections. In second or fourth generation designs, this may be done by slower scanning speeds or higher frequency detector sampling. In third generation designs, the Nyquist criterion is met with the 1/4 ray offset of the center of rotation, so that opposing projections (in 360° scans) are offset by 1/2 ray width. Aliasing can also be reduced in third generation designs by combining two adjacent projections into a single finer sampled projection (as in some high resolution modes of certain third generation systems). Aliasing is only secondarily affected by use of a smoothed convolution kernel.

b) Edge Gradient Streaks

This artifact is a consequence of sharp discontinuities in attenuation within the finite width of the sampling ray, and appears as a lucent streak extending for a short distance from edges of high-density objects. It is much more prominent if the edge is long, thus is more common in scanning of non-physiological objects (some phantoms, mechanical prostheses, etc.). It is not possible to eliminate, although algorithms that suppress it are possible. It can also be reduced by the use of a harder x-ray beam, which effectively reduces the magnitude of the boundary contrast. This artifact is not thought to be significant in most clinical conditions but can be problematic in quantitative CT[15].

c) Geometric Misalignment

Mechanical misalignment in the detectors or x-ray source cause positional uncertainty (misregistration) of ray data in reconstruction. The effect is most apparent with centrally located high frequency objects, causing radiating streaks. The effect is actually an artifact only in parallel ray scanners. In rotate only systems, the effect is a degradation in spatial resolution. Provided that actual positions of all rays are accurately known, the effect may be eliminated by software correction, or in translate-rotate systems by mechanical alignment.

d) Motion Artifacts

The effects of patient motion are indistinguishable from mechanical misalignment since both result in ray positional uncertainty. With motion, the patient rather than the focal spot or detector is out of position. Motion artifacts commonly appear as long streaks from air, bones, or other high-contrast objects. Motion within a single projection causes a more severe artifact than motion varying gradually from one projection to the next. Because fourth generation and asymmetric-fan third generation scanners recombine partial projections separated in time, they are more susceptible to motion induced error within a single projection. In fourth generation designs, over scanning is done (at least in body scans) to reduce the contribution of the combined projections to the reconstruction. This solution works well but results in a higher dose.

2) Reconstruction Algorithm Effects

a) Point Spread Effect

In all practical scanners the margins of imaged objects are blurred due to the system resolution limits. This spreading of margins can increase the apparent size of the imaged object, particularly for small high contrast objects. Such blurring is an inevitable consequence of system resolution limits. Point spread effect is somewhat dependent on window settings and can influence dimensional measurements if not properly considered[16].

b) Edge Enhancement Artifacts

In some imaging circumstances it is advantageous to employ “edge enhancement” convolution kernels in image reconstruction. These kernels make edges more prominent by enhancement of spatial frequencies near the system resolution limits, One result is that the CT numbers across a high contrast margin may exhibit some oscillation and overshoot. A bone-soft tissue interface may exhibit elevation of the bone CT numbers and depression of soft tissue numbers at the margin. The magnitude of this effect de-

pends on system resolution and kernel frequency characteristics. For this reason, edge enhancement kernels are generally not appropriate for quantitative CT.

3) Attenuation Measurement Errors

This artifact is due to errors in the value of the measured quantity, rather than in the position of the measured data in the image matrix. Such errors may be due to detector measurement error, x-ray source problems or deviations from the assumed linearity of attenuation measurement.

a) Detector or Source Variations

Since detectors define ray position in third generation systems, inevitable temporal drift in detector response results in ring artifacts, which can be avoided by periodic calibration "air scans". Detector drift in second and fourth generation machines can be corrected "on the fly" and generally causes no obvious artifacts, but reduced detector output or failed detectors significantly increase image noise.

Near the end of its useful life, an x-ray tube can exhibit anode wobble causing cyclical variations in output. This produces a severe Moire pattern artifact in fourth and second generation systems but relatively little effect on third generation systems.

b) Nonlinear Attenuation Errors

The physical foundation of CT is based on attenuation measurements; attenuation is a (logarithmically) linear process provided that the detector response is linear, the beam is monochromatic and scatter is excluded. Deviation from this ideal results in the production of artifacts.

(1) Detector Nonlinearity

CT scanner detectors can exhibit saturation effects at very high detected x-ray fluence and high dark current at very low fluence. These effects, as well as hysteresis, can cause some deviation from linearity, particularly at detector dynamic-range limits. Schemes to subtract dark current, mathematical linearization methods and dynamic range reduction, greatly reduce detector non linearity in modem CT systems. Shaped or "bow tie" filters are used in some scanner designs to reduce beam dynamic range.

(2) Spectral Nonlinearity - Beam Hardening

Although CT x-ray spectra arc typically heavily filtered, some unavoidable beam hardening occurs in passing through the patient, resulting in some re-

duction in the measured attenuation. The magnitude of the effect depends on the thickness, density and atomic number of tissues in the ray path. In a uniform phantom, beam hardening produces the familiar "cupping" artifact where CT numbers in the center are lower than at the periphery. This artifact is commonly corrected by a "water" correction, usually with more success in phantoms than in patients. A "bone hardening" correction is also commonly employed in head scans, which additionally corrects the hardening effect within the skull. This correction is usually less successful where the bone is very thick, as in the base of the skull. Beam hardening artifacts in patients often appear as lucent streaks between dense bones or other high atomic number objects.

Shaped x-ray filters may produce subtle beam hardening if the filter shape is not matched to the hardening produced in the patient. These effects may be problematic in quantitative CT.

(3) Scatter Effects

Physicists are familiar with the necessity of "narrow beam" geometry for attenuation measurements to avoid detection of scattered x-rays. The effect in CT is classical, producing an apparent reduction in attenuation as in beam hardening. The resulting artifact is similar, i.e., lucent streaks between dense objects, but is independent of their atomic number. Unlike beam hardening, scatter artifacts worsen with increased slice thickness and larger patients. Scatter cannot be totally excluded even with well collimated detectors[17], although the magnitude of the artifact is generally less than that due to beam hardening. Since the design of fourth generation type scanners precludes source convergent interdetector collimation, they are more susceptible to scatter artifacts. Scatter artifacts may be partly corrected with knowledge of the scatter fraction[17]. One fourth generation design samples the scatter distribution just outside the scan plane.

b. Measurement of Systematic Pixel Error

1) Field Uniformity

Purpose: To determine the spatial uniformity of CT numbers in a uniform

Rationale: The uniformity test is a simple and direct approach to determining accuracy of the reconstruction process. To evaluate scan plane uniformity, a test object with appropriate dimensions and uniform attenuation is scanned under simulated clinical conditions.

Tools needed: Head and body uniformity phantoms of 15-21 cm and 30-32 cm diameters (see Appendix B). Optional: An elliptical body uniformity phantom to better approximate actual patient cross sections[18] (Figure B-2, Appendix B).

Method:

- Perform calibration scan according to manufacturer's recommendations.
- Place head phantom in the gantry centered to the rotational axis, using phantom mounting bracket if available.
- With nothing else in scan field, scan with standard head parameters and widest slice width.
- Examine image with a narrow window to rule out subtle artifacts (rings, streaks, etc.); repeat scan if necessary.
- Repeat test with elliptical and circular body phantoms using body scan settings.
- Repeat body phantom test with narrowest slice width.
- Repeat test at all calibrated kVp stations and with any operator selectable beam hardening or compensating (bow tie) filters. For compensating filters, test only with appropriate phantom (head phantom with head filter, etc.).
- Determine average CT number in an ROI of ~1.0 cm^2 in the center of the phantom and at four locations at the phantom periphery (make sure that ROI contains only water).

Interpretation: Add 1000 to each measured value and express nonuniformity at each peripheral ROI location as percent difference of value measured in the center. There is no universally accepted value for field uniformity, however, modem scanners can provide field uniformity of ±5 CT numbers or better under most scan conditions. For a visual display of nonuniformity, use the pixel profile utility to draw orthogonal profiles across the phantom then record hard copy image of phantom with superimposed profiles. Nonuniformity which worsens with slice width suggests a scatter effect[19], while

nonuniformity with change of phantom size or shape suggests inadequacy of the software uniformity correction.

2) Considerations in Quantitative CT

In quantitative CT (QCT), CT numbers are used directly for tissue characterization. In some QCT techniques such as xenon cerebral-blood-flow measurements and vascular-flow studies with iodinated contrast media, measurements are relative between successive scans in a series. Other QCT techniques such as bone mineral analysis and assessment of calcium in lung nodules are absolute measurements, where CT numbers are used directly for quantitation. Because bone mineral analysis (BMA) is a relatively stringent measurement and is the most commonly done QCT technique, the considerations for acceptance testing of a scanner for BMA will be discussed.

BMA is most commonly measured as trabecular bone mineral from average CT numbers in an ROI placed within thoracic and lumbar vertebral bodies. The measured CT number is calibrated by scanning a standard test object, containing materials with known mineral content In one method, the standard is placed in the scan circle with the patient and appears in the same image. In the other, a phantom with standards is scanned after removing the patient from the gantry. The fit method is susceptible to spatial variation in CT numbers across the scan field; the second is dependent on short term precision and does not exactly simulate the patient size or position (see Figure B-2, Appendix B). Both techniques require precise slice positioning from a radiographic (scout) image.

A widely used concurrently scanned calibration object[20] contains solutions of 0, 50, 100, and 200 mg/ml K_2HPO_4 in water for the calibration of bone. and ethanol for fat. This object replaces the table cushion, so that the patient is scanned lying upon it. For each scan slice, a calibration curve of CT numbers versus K_2HPO_4 concentration is fitted, and the bone mineral values obtained in the vertebral body ROI are computed in mg/ml K_2HPO_4 equivalent.

The reader may notice that a measurement of CT linearity is not included in this document, since deviations from linearity are rarely seen. Since the linearity measurement does yield the effective energy of the x-ray beam, it can be useful in some quantitative work (see references 27 and 38).

Many tests described in other sections have a direct bearing on scanner performance for QCT. The most important of these are: Table/Gantry Tilt or Slice Localization From Radiograph (Scout), Table lncrementation, Sensitivity Profile, X-ray Generator Calibration, and Field Uniformity tests. The following additional test directly addresses problems encountered in QCT of the spine.

Scan Field Position Dependence

Purpose: To determine the dependence of measured bone density on scan field location, and to derive correction if necessary.

Rationale: Some scanners, particularly those with shaped (bow tie) beam filters may **show** a marked influence of scan field position on measured bone density. Due to radial scan symmetry, scan field dependence can be tested in most scanners by varying table height. It may then be possible to construct a correction which varies as a function of table height.

Tools needed: Elliptical body phantom with 100 mg/ml K_2HPO_4 or other trabecular bone equivalent spinal insert (a circular phantom is not recommended as some measured effects may stem from phantom shape - see Figure B-2, Appendix B).

Measurement:

- Place phantom in gantry and measure distance from vertebral insert position to gantry rotational axis. Set body scanning conditions for use with BMA and reconstruct with a convolution kernel recommended by the manufacturer for QCT.

- Perform a series of CT slices with the 100 mg/ml K_2HPO_4 insert in the vertebra location, changing table height by 10 mm between scans. Cover range of table heights typically used for bone scans.

- Repeat for other kVp settings and other filter combinations appropriate for body scans (if available).

Interpretation: For each slice, determine mean CT number in the 100 mg/ml insert For each kVp and filter combination, plot this value as a function of radial distance from scan rotational axis. If variations appear to be random

(some random variation is expected), then this may indicate a short term reproducibility problem. Repeat scans at a higher mAs to see if improvement is seen. If systematic variation with table height is observed, develop a systematic correction from measured data as a function of radial position.

3. Spatial Frequency Limits

In any practical CT system, image noise and blurring place upper limits on the spatial frequencies of the patient reproduced in the image. The relative importance of noise and blurring is a function of image contrast. For very low contrasts, object visibility is primarily constrained by image noise, and is independent of blurring effects. At very high contrasts, noise effects are negligible, and object visibility is constrained only by blurring sources. Between these extremes, as contrast increases, the effect of noise on object visibility becomes less important while the influence of blurring sources grows. In a real patient, visualized tissues span a wide range of contrasts, therefore the evaluation of spatial frequency limits of a scanner should span a similar range. Two tests are proposed here, a high contrast resolution test using either the MTF or a resolution pattern for evaluation of blurring factors, and a low contrast resolution test to evaluate the effect of noise on perceptibility limits. It is suggested that high contrast resolution be evaluated by either the resolution pattern or the MTF; the low contrast resolution test may substitute for noise measurement.

a. High Contrast Spatial Frequency Limits

The amount of blurring in a CT image may be influenced by: system geometric resolution limits, ray sampling, pixel size, and properties of the convolution kernel. Geometric limits are described by Yester and Barnes[21] as:

$$A_{eff} = \frac{1}{M} \cdot \sqrt{a^2 + (M - 1)^2 \cdot s^2} \qquad [11]$$

where M is the geometric magnification at the center of rotation, a and s are the widths in the scan plane of the detector and focal spot, respectively. The effective aperture, A_{eff}, is effectively the sampling ray width in the scan plane, and is the major limitation on spatial resolution in a given system. It is apparent from Equation [11] that resolution can be improved by reducing focal spot size, or detector width, and if the focal spot is small enough, by increasing magnification. Since only the scan plane dimension of the focal spot influences resolution, x-ray tubes are often designed with rectan-

gular foci. with a larger z dimension to optimize loading. Some fourth generation scanners have dual foci x-ray tubes and the capability of reducing the detector width by masking, and thus can more easily vary spatial frequency limits. In most scanners, geometric magnification is constant, but one third generation design varies geometric magnification by shifting the center of rotation with respect to focal spot and detectors.

Spatial resolution is also influenced by ray spacing within a projection data set. As noted previously, ray spacing must allow overlap by $A_{eff}/2$ to prevent abasing; further reduction in sample spacing will produce some resolution improvement, within limits constrained by geometry. Improved ray spacing is technically simple with second and fourth generation designs, but in third generation designs is limited by detector spacing. This limitation is overcome in some implementations by combining adjacent projection sets in scans where projection spacing is very fine.

Pixel size may affect displayed image resolution if pixel dimensions are large with respect to A_{eff}. Although this is somewhat dependent on ray sampling and selected convolution kernel, pixels wider than $\sim A_{eff}/2$. tend to limit image resolution (pixel width = FOV/matrix diameter). Most modem scanners permit image reconstruction on a wide range of FOV's, and some provide different reconstruction matrices. Varying FOV and matrix size can therefore produce both pixel-size and non pixel-size limited images. (Note also that pixel size variation influences image noise only if resolution is affected).

The kernel convolved with projection data prior to back projection is designed to influence spatial frequency content of the data set and therefore affects spatial resolution (and noise). Most scanners provide a selection of kernels optimized for different imaging conditions. Kernels may be matched to spatial frequency limits governed by sample spacing and geometric factors, or may be designed to enhance or suppress certain frequencies. A "matched" (e.g., Shepp-Logan) kernel can be expected to deliver the most faithful image reconstruction without altering intrinsic noise and resolution content of the data set. Such filters are most appropriate for quantitative CT. Kernels which suppress high spatial frequencies (smoothing kernels) reduce image noise but cause some loss in spatial resolution. Alternatively those that enhance high spatial frequencies (edge enhancement kernels) produce some resolution improvement but also enhance high frequency noise.

1) Measuring Resolution With A Resolution Pattern

Purpose: To determine the high contrast spatial frequency limits of the CT scanner under various conditions.

Rationale: It is generally more practical to use a resolution pattern for evaluation of spatial resolution, then to perform measurement of MTF, if desired A number of patterns are available from different phantom manufacturers and can be used here. It is suggested that a pattern be selected with a resolution that exceeds the specified resolved frequency limits for the scanner in question.

Tools needed: Appropriate resolution test object from CT manufacturer or third party phantom manufacturer.

Method:

- Place test object on tabletop and raise table to align object to center of rotation and to scan alignment light. Make sure neither table nor gantry are tilted. If properly aligned on first scan, do not move object through remainder of test.

- Scan phantom in all standard head and body modes with 8-10 mm slice width. Save one set of raw data for both standard head and standard body modes.

- To evaluate resolution enhancement modes: With all other factors set for normal head scan, vary scan factors (one at a time) which alter resolution, i.e., ray sampling, focal spot size, detector aperture, etc., but not convolution kernel or pixel size.

- To evaluate convolution kernel effects: Reconstruct the saved head scan data set with each available convolution kernel.

- To evaluate pixel size effects - Using saved raw data set for standard head scan conditions, reconstruct with the largest image matrix and the "standard" convolution kernel. Repeat several times varying FOV from the smallest to the largest size available. Repeat if other matrix sizes are available.

Interpretation: Set a narrow window width, with window level halfway between the pixel values of the two materials in the resolution pattern. A wider window may be necessary for imaging modes where the MTF is distinctly different from a simple exponentia1[22]. For each image, determine the observed limiting resolution. If it is necessary to magnify the image to visualize the test object, use a non-reconstruction magnification, which does not alter actual (data) pixel size. Save images from acceptance test for comparison in future quality assurance measurements.

2) Modulation Transfer Function Measurement

Purpose: To derive the modulation transfer function as a measure of the limiting spatial resolution characteristics of a CT system.

Rationale: The modulation transfer function (MTF) mathematically describes the capability of the system to reproduce the range of spatial frequencies in the imaged object. The MTF can be measured from CT data in several ways, all of which require a separate computer for data analysis. An infinitely small, high contrast point (impulse function) theoretically contains all possible spatial frequencies and the mathematical description of its image, the point spread function (PSF), can be used to derive the MTF. Similarly, the profile across the image of an infinitely thin line (line spread function, LSF) or an extremely sharp edge (edge response function, ERF) can also be employed[23,24]. Assuming that resolution in the image plane is the same in all directions, one dimension of the PSF is equivalent to the LSF. The LSF may be obtained directly or as the first derivative of the ERF. The MTF is then the Fourier transform of the LSF.

The physicist should be aware that some manufacturers provide test objects and software for the determination of the MTF, which will suffice in most cases. The following discussion provides an alternative method, useful if not provided by the manufacturer.

Because pixel data are discrete, a fitting function is necessary to obtain a continuous LSF. Unless programs to derive the MTF are available on the CT system computer, or data can be ported to an off line computer, the process involves manual entry of many pixel values for each MTF. In most CT systems numerous factors influence its spatial resolution characteristics, but it is seldom practical to derive the 20 or 30 MTF's necessary for a complete description. Simple methods which allow the MTF to be derived from a small amount of data are therefore preferable. One such method is the deriva-

tion from the square wave response function of Droege and Morin[25], and another is the simplified method of Nickoloff and Riley[26]. Although the former method has considerable value, the test object is more difficult to fabricate. The method of Nickoloff and Riley will be described for a LSF taken from an image of a copper foil.

The following assumes that the LSF is reasonably approximated by a gaussian function, a satisfactory approximation for most clinical convolution kernels without a high degree of edge enhancement[26]. Pixel values are obtained from an ROI including the blurred foil image and entered into a function of the form:

$$y(x) = \sqrt{\ln\left(\frac{1}{p_x}\right)} \qquad [12]$$

Where p_x are normalized pixel values. A linear regression of y(x) as a function of distance from the peak of the foil image (in mm) is then obtained; a good fit, i.e., r>.99, indicates a gaussian LSF. If α is the slope of the regression line, the MTF at spatial frequency υ in cm^{-1} is then:

$$MTF(\upsilon) = e^{-\left(\frac{\pi \cdot \upsilon}{10 \cdot \alpha}\right)^2} \qquad [13]$$

The resulting MTF can then be expressed by plotting the MTF (υ). or as the computed modulation at a specific cutoff frequency (0.1 or 0.05 are suggested).

Tools needed: LSF or PSF phantom from CT manufacturer (or Appendix B, Figure B-4), and MTF computation program from CT manufacturer or method to extract pixel values from CT ROI, i.e., line printer, data link, etc.

Method:

- If PSF phantom and MTF computation program are available from the CT manufacturer, follow the provided procedure, otherwise:

- Place phantom in scanner gantry, with foil vertically oriented across center of rotation and aligned perpendicular to the scan plane. Make sure that neither gantry nor tabletop are tilted.

- Scan phantom with standard head scan settings, but at highest mAs and narrowest slice width (ensure that increase in mAs does not alter resolution).

- Examine image of plate with non-reconstruction magnification (i.e., without changing FOV) and narrow window for subtle streak artifacts, extending orthogonal to the foil image. If seen, realign phantom and repeat scan. (Streaks from ends of foil are often present and should not influence results). Do not attempt MTF if significant artifacts or obvious negative lobes adjacent to plate are present.

- Repeat under highest resolution head scan and standard body scan conditions.

- Place ROI across plate image near center of image and extract pixel values, making sure that ROI includes both "tails" of the blur distribution.

- Determine angle of plate image to that of pixel rows, find pixel spacing (FOV/matrix diameter), then correct for angle to determine X displacement along pixel row from peak of plate image. Normalize pixel values to range determined by acrylic background and peak value of plate. Fit values by least squares as linear function of Equation [12]. averaging over several parallel rows (3-5, depending on noise level).

- Determine MTF using slope of regression line *(a)* in Equation [13].

Interpretation: If measured MTF is significantly different from manufacturer's specifications, recheck procedure for errors and repeat if necessary. The foil must be perpendicular to the scan axis; a small error (<5°) can be significant. Before reporting discrepancies, be certain that measurement conditions do not differ from those used by the manufacturer.

b. Low Contrast Spatial Frequency Limits

Purpose: To determine the capability of the scanner to discriminate low contrast objects.

Rationale: Since much relevant soft tissue detail is low contrast in **nature,** this is perhaps the most clinically important performance test. The visibility of low contrast objects is constrained mainly by amplitude and frequency characteristics of the image noise[11]. For the human observer, perception of barely discernible objects in a noisy background is influenced by psychophysical considerations which vary between observers and are difficult to quantify. Subject contrast in CT is simply the difference in average CT numbers between two adjacent regions of the image. Since the CT number is related to the attenuation coefficients of water and the material in the voxel, subject contrast can be expressed as:

$$SC = \frac{k}{\mu_w(E)} [\mu_1(E) - \mu_2(E)] \qquad [15]$$

where $\mu_w(E)$, $\mu_1(E)$, $\mu_2(E)$ are the energy dependent linear attenuation coefficients of voxels containing water, material 1 and material 2, respectively, and k is the CT number scaling constant (1000). Producing an energy independent response would require that materials 1 and 2 differ by an energy independent parameter, i.e., physical density, and that the attenuation coefficients of the two materials differ from that of water by an energy independent constant[27]. Assuming materials 1 and 2 are identical in composition and differ only in density (ρ):

$$SC = \frac{k}{\mu_w(E)} \cdot \frac{\mu_1}{\rho_1}(E) \cdot [\rho_1 - \rho_2] \qquad [16]$$

A phantom module employing this principle is produced by The Phantom Laboratory (Phantom Laboratory Inc., Cambridge, NY). Another test object which can be fabricated is described in Appendix B (Figure B-5). This object produces variable density by the use of partial volume effect. The phantom is constructed from a stack of thicker cellulose acetate sheets, which encloses a thinner stack of live thin (0.1 mm) layers of acetate. Each sector contains a single row of holes radially increasing in diameter, from 1 mm to 32 mm in doubling steps. For a 10 mm slice width, (nominal) contrast between sector and holes increases circumferentially, from approximately 1 to

5 CT units by varying the number of pierced sheets in each sector. Opposing sectors are paired so that there are two holes of each size and contrast.

Tools needed: Low contrast detectability test object (Appendix B, Figure B-5) or other commercially available test object.

Method:

- Perform calibration scan according to manufacturer's recommendations.
- Place test object on tabletop and align to scan alignment light and center of rotation. Make sure object is parallel to scan plane and neither table nor gantry are tilted. When properly aligned, tape in position for remainder of test.
- Scan under standard head scan conditions; Save raw data set.
- Examine image with a narrow window for presence of obvious artifacts (streaks or rings). Repeat air scan if necessary.

To fully characterize contrast discrimination performance, assess effects of each of the. following parameters:

- mAs - Using all conditions normalized for head scans, scan at lowest and highest mAs setting (ensure that resolution factors do not vary with mAs).
- Tube potential - Using standard head scan parameters at each kVp, scan phantom but ensure that resolution factors remain constant.
- Resolution enhancement scanning modes - Perform a series of scans after varying factors which alter resolution, i.e., ray sampling, $A_{eff.}$ etc., but not convolution kernel or pixel size; vary only one factor at a time with all other factors for standard head scan.
- Convolution kernel - Using the same raw data set, reconstruct the image with each available kernel, using standard head scan FOV

and matrix size. Repeat for body scan and pediatric scan modes, with appropriate kernels and raw data sets.

Interpretation: Results depend on actual contrast levels and are critically dependent on display window settings. Actual contrast depends on sensitivity profile width, therefore it must be measured at each slice width used. Begin by examining all images to select ones with highest and lowest noise level. With lowest noise image and a ROI of 100 or 200 pixels, determine contrast as difference in mean CT# between largest hole in each sector and surrounding plastic. Repeat for each slice width. Set standard window settings using noisiest image; measure standard deviation from ROI in highest contrast hole. Set window width equal to contrast between highest contrast hole and surrounding material +5 standard deviations of ROI. Set window level midway between CT numbers of plastic and highest contrast hole. Keep same window settings for all images with same slice width. Record images on multiformat camera, in standard format used for clinical images. For each image, plot smallest perceptible hole in each sector as a function of contrast, on a semilog scale; it is suggested that a hole should be considered as perceived only if both holes of that contrast and level are seen. Since this is an observer performance test, it is preferable to average results over at least two observers, with viewing conditions carefully standardized. Because the positions of all holes are known *a priori,* readers have a tendency to overestimate contrast limits. Save all images for future reference in quality assurance testing.

4. Video Display and Multiformat Camera Image Quality

In most clinical circumstances, the physician's interpretation is accomplished from a transparency image recorded with a multiformat camera. Ideally, the transparency image reproduces the quality of the original image displayed on the system monitor, and the display monitor reproduces the available image quality. The following procedure employs the use of the Society of Motion Picture and Television Engineers (SMPTE) digital test pattern[28]. This pattern is available from most CT manufacturers as a stored image data file for setup and assessment of displayed and recorded images. Alternatively a separate pattern generator (available from accessory vendors) can be used. The following procedure describes the technique for optimization of video display screens and multiformat camera images.

a. Visual Display Screen Setup and Quality Control

Purpose: To ensure that all of the information in the video signal is displayed on the video display.

Tools needed: Society of Motion Picture and Television Engineers (SMPTE) digital test pattern[28] and x-ray film densitometer.

Method:

- Clean front surface of the cathode-ray tube (CRT), including the front and back surfaces of any anti-reflective screens present, with appropriate cleaner and a soft cloth.
- Reduce room illumination to the normal viewing level. (Note: Typical room illuminance levels should be on the order of 5 to 10 lux (cd/m^2) or lower)
- Display the SMPTE test pattern on the CRT.
- Adjust the window width to just encompass the range of numbers comprising the SMPTE test pattern.
- Adjust the window level to either the lower or middle value of the window (depending on the particular software) so that the entire test pattern is visible.
- Turn both brightness and contrast controls completely counter-clockwise.
- Turn the brightness control clockwise until the video raster pattern is just visible on the CRT.
- Turn the contrast control clockwise until the image is bright and clear, and both 95% and 100% patches are clearly separated. Do not increase contrast beyond point where alphanumerics become blurred or streaked on the display.
- Examine the displayed image for the following:

  a) The 5% patch should just be visible inside of the 0% patch;

b) The area of the 0% patch should be almost black with raster lines just barely visible;

c) The 95% patch should be visible inside the 100% patch;

d) The alphanumerics should be sharp and clear.

- Repeat procedure for each display, e.g., operator's console, physician's console, so that each display appears similar.

Note: Some video monitors do not have an adequate "black clamp". Consequently, the black areas of the image will increase in brightness as the contrast is adjusted. In this case, the brightness control will have to be turned counter-clockwise to maintain the brightness of the black areas as the contrast is increased.

b. Setup of Multiformat Camera Image

Purpose: To ensure that optimal image quality is reproduced on the hard-copy image (e.g., video hard-copy camera, laser camera or printer, video printer). Secondarily, to ensure long term reproducibility of hard copy image quality.

Method

- Using standard sensitometric methods, assure that film processor is processing at an optimum level, using the same type of film exposed in the hard-copy camera[29].

- For video hard-copy cameras, clean all optical surfaces with appropriate cleaner and lens cleaning tissue including front of the CRT, the folding mirror (lake extreme care if mirror is front-surface@, and other optical surfaces.

- Display the SMPTE test pattern at the window and level as described in the visual display set-up procedure.

- Adjust the hard-copy camera controls so that film densities correspond to those in Table 6, for the appropriate patches of the gray scale.

- Examine the hard-copy image and compare it to the visual display.

Interpretation: If the OD (optical density) of the 0% patch is increased beyond 2.45 ± 0.10, the visibility of the higher densities will be compromised. If more contrast is desired in the clinical images, window and level should be adjusted. With the densities produced on the film as described here. one can assure that all of the information in the video signal or digital data is being displayed so that it can be perceived by the observer.

Table 6: Optimal SMPTE test pattern optical densities for multiformat camera films.

SMPTE Patch	Video Camera	Laser Camera
0%	—	2.45±0.10
10%	1.80±10.10	2.00±0.10
40%	1.15±0.08	1.15±0.08
70%	0.50±0.05	0.65±0.05
90%	0.28±0.03	0.35±0.03

C. Radiation Dose

Purpose: To determine the radiation dose to tissues under different CT scan conditions.

Rationale: In a CT scan, rotation of the x-ray source results in a narrow band of radiation extending partially or completely around the patient. Within the slice volume, the dose distribution exhibits symmetry reminiscent of arc teletherapy. The shape of the distribution is a function of scan arc, which varies from 180° to >400° in clinical systems. At or near the skin surface, the single scan dose (D_s) varies with tube potential, beam filtration, mAs, source skin distance (SSD), and patient attenuation factors:

$$D_s \propto kVp^c \cdot mAs \cdot \frac{B}{SSD^2} \qquad [17]$$

where the exponent c (≈3) varies with type and shape of filtration, and B is the patient transmission. Dose in the center of the patient, however, is mainly determined by transmission. In practice, measurements with CT dosimetry phantoms show that patient transmission and SSD have opposing effects on surface dose, i.e., larger patients have smaller SSD's but reduced transmission, while smaller patients have greater transmission but larger SSD's. Systems using shaped or bow-tie beam filters produce lower surface doses for the same operating factors due to reduced beam intensity toward the fan edges. SSD varies widely from one scanner design to the next, and is variable in third generation scanners with variable magnification. Commonly SSD is smaller in fourth generation than in fixed geometry third generation scanners (not true for "nutating" ring fourth generation designs). Since clinical studies seldom consist of a single scan, the procedure dose is summed over all contributing scans since tissues irradiated in one slice receive contributions from other slices due to scatter and field overlap. For an infinite series of contiguous slices, Shope et al.[30] showed that the average dose parallel to the scan axis over an interval I in length is independent of position. The multiple scan average dose (MSAD) is defined as:

$$MSAD = \frac{1}{I} \int_{-I/2}^{+I/2} D(z)dz \qquad [18]$$

where D(z) is the dose at position "z" parallel to the z (rotational) axis in the infinite series. The Computed Tomography Dose Index (CTDI) is defined by the Center for Devices and Radiological Health (CDRH) as[31]:

$$CTDI = \frac{1}{nT} \int_{-7T}^{+7T} D(z)dz \qquad [19]$$

where n is number of slices per scan (1 in most newer systems) and D(z) is the dose at point "z" on any line parallel to the z (rotational) axis for a single scan. Quantitatively, the CTDI is the average dose over an interval of width "T" equal to the selected slice width, at a point (x,y) in the plane of the middle slice of a series of fourteen scans. It has been shown[30] that the limiting conditions do not require an "infinite" series, but only one large

enough so that slices at either end of the series arc sufficiently far to contribute negligibly to the middle slice. For conditions where the number of slices is not "infinite", the multiple scan average dose (MSAD) is defined a s[32]:

$$MSAD = \frac{1}{I}\int_{-mI/2}^{+mI/2} D(z)dz \qquad [20]$$

where "m" is the total number of scans in the clinical series. Note that this is a modification of the limits of integration of the original formula described by Shope et al.[30], as suggested by Spokas[33]. Equations [19] and [20] are identical if "m" is equal to 14, and the interval "I" between slices equals the slice width. Shope et al.[30] have also demonstrated that where the scan increment differs from the slice width, the MSAD can be obtained from the CTDI as:

$$MSAD = \frac{T}{I} \cdot CTDI \qquad [21]$$

Equation [21] does not hold if I = 0 or if I>>T. If I = 0 as in dynamic flow studies, the MSAD is a multiple of the single slice peak dose. In the opposite extreme where slices are widely separated, the MSAD is meaningless because of the essentially unirradiated regions between scans, hence the single slice peak dose should be used.

The CTDI or MSAD is usually measured with a special CT ionization chamber, although TLDs may also be used. In either case, measurements are made within standard CT dosimetry phantoms[31].

1. Ion Chamber Dosimetry

The use of the standard CT ionization chamber is considerably easier than TLD dosimetry, but requires some care. The standard "pencil" ionization chamber, first described by Jucius and Kambic[34], and Suzuki and Suzuki[35], employs the principle of volume averaging. These chambers are designed to extend through the slice plane and integrate exposure over the chamber's active length (10 cm in the standard chamber). The exposure measurement thus obtained is equivalent to the exposure at the midpoint of a series of

contiguous slices equivalent to the length of the chamber (i.e., ten 10 mm, twelve 8 mm slices, etc.).

Tools needed: Head and body CT dosimetry phantoms (see Appendix B), CT air ionization detector, and acrylic sleeves to lit within CT phantoms, adhesive tape and phantom alignment rod (Appendix B, Figure B-7).

Method:

- Place CT head dosimetry phantom on the patient table and align phantom axis parallel to scan rotational axis with slice plane through mid length of phantom.

- Put alignment rod in central phantom hole. Take standard head scan with a slice width of 5-6 mm, look at image to see that all three holes in alignment rod are clearly seen, and realign phantom if necessary. Tape phantom in position when aligned.

- Put ion chamber in hole near surface (1 cm depth hole) corresponding to maximum dose point. If rotation angle of scan mode to be tested is <360°, then rotate phantom so that one dosimeter position (at 1 cm depth) is aligned with midpoint of scan arc. If scan angle is >360° (overscan) then position dosimeter location at middle of overscan region. If this position is unknown, take several measurements with chamber at different positions. Note that with scanners with continuous rotation capability, the maximum dose position varies from one scan to the next; consult manufacturer before proceeding[36].

- Measure exposure integrals for all standard head techniques, at all calibrated kVp settings, and with any operator selectable beam filter combinations (appropriate for head scanning). Also measure for all available slice widths.

- Repeat with body and infant phantoms as appropriate, using above alignment procedure.

Interpretation: MSAD values should be determined for all standard scan conditions, and for any variable which might be expected to influence dose. In interpreting data determine whether parameter changes produce the expected result on dose, e.g., a linear change with mAs or slice width. Ensure

that all conditions are comparable when comparing with manufacturer's specifications.

2. TLD Dosimetry

The following assumes that the physicist is familiar with handling, annealing, calibrating and reading of thermoluminescent dosimeters. Collimation accuracy should previously have been checked (see Section III.A.6.). One should be aware that the thickness of the standard LiF TLD ribbon (-0.9 mm) can introduce averaging error in very thin sections and should not be used for slices less than 3 mm. The resolution of TLDs will not give better than 20% estimation of radiation profile width with small slices.

Tools needed: Standard head and body CT dosimetry phantoms, TLD holder(s) loaded with LiF TLDs (3 x 3 x 0.9 mm), phantom alignment rod (see Figure B-7) and suitably calibrated TLD reader with all necessary accessories. See Figure B-6 for a TLD holder with fixed spacing or use a holder with a hollow through which TLDs may be packed separated by acrylic spacers of precise dimensions that may be color coded for expediency[32]. Note for the holder shown in Figure B-6, TLDs can be loaded one at a time by sliding insert along slot in rod to individually expose slots for loading. Tape or screw insert into position when loaded. Reverse process to unload.

Method:

- Place CT head dosimetry phantom on the patient table. Align phantom with its axis parallel to scan rotational axis and with slice plane through mid length of phantom.

- Put alignment *rod* in central phantom hole (do not put TLDs in phantom until alignment and all other conditions are assured!).

- If rotation angle of scan mode to be tested is < 360°, then rotate phantom so that one dosimeter position (at 1 cm depth) is aligned with midpoint of scan arc. If scan angle is > 360° (overscan), then position dosimeter location at middle of overscan region. If these positions are unknown, determine them from peak exposure value measured with CT ion chamber in dosimetry phantom. (See above caveat for scanners with continuous rotation capability.)

- Take standard head scan with a slice width of 5-6 mm, look at image to see that all three holes in alignment rod are clearly seen and

realign phantom if necessary. Tape phantom in position when aligned.

- Making sure that all conditions arc set for standard head scan, put TLD holder in position at 1 cm depth and scan.

- Repeat alignment procedure and measurement for body and infant phantoms as appropriate.

Interpretation: For determination of MSAD from TLDs it is necessary to fit measured dose values to an appropriate function for integration. For scanners without excessive field edge penumbra (most systems with rotating anode x-ray tubes), a function consisting of a square wave superimposed on a single exponential function will suffice. The width of the square wave should be equal to the measured radiation profile width (see collimation tests). Data in the field center are fitted after subtracting scatter tails fitted to single exponential with symmetry about the center of the x-ray field. This is demonstrated in Figure 2 with measured TLD data. In this case, data fit a function of the form:

$$\frac{-FWHM}{2} \le z \le \frac{FWHM}{2}, \quad D_z = D_m + a \cdot 10^{-bz}$$

$$\frac{-FWHM}{2} > z > \frac{FWHM}{2}, \quad D_z = a \cdot 10^{-bz}$$

where: FWHM = Radiation profile width at half maximum
D_z = Fitted dose value at position z
D_m = Mean measured dose for z within ±FWHM/2
a, b = Coefficients of fitted exponential function

IV. Summary

The acceptance testing of a CT system is intended to ensure that the user receives a system with known capabilities and limitations, and which is functioning optimally at its design specifications. The methods that are described herein are designed to test all relevant aspects of system performance under conditions which best simulate anticipated clinical conditions. Tests are also designed to determine the limits of scanner performance, e.g., its highest spatial resolution, and highest contrast resolution conditions, as well as determine the compromises in radiation dose or other factors neces-

sary to achieve that performance level. Special attention is given to optimization of the multiformat camera image since the diagnosis is commonly done from these images. Once a system is deemed acceptable, the acceptance test data forms a benchmark for future quality assurance testing, so that any degradation in performance can be seen.

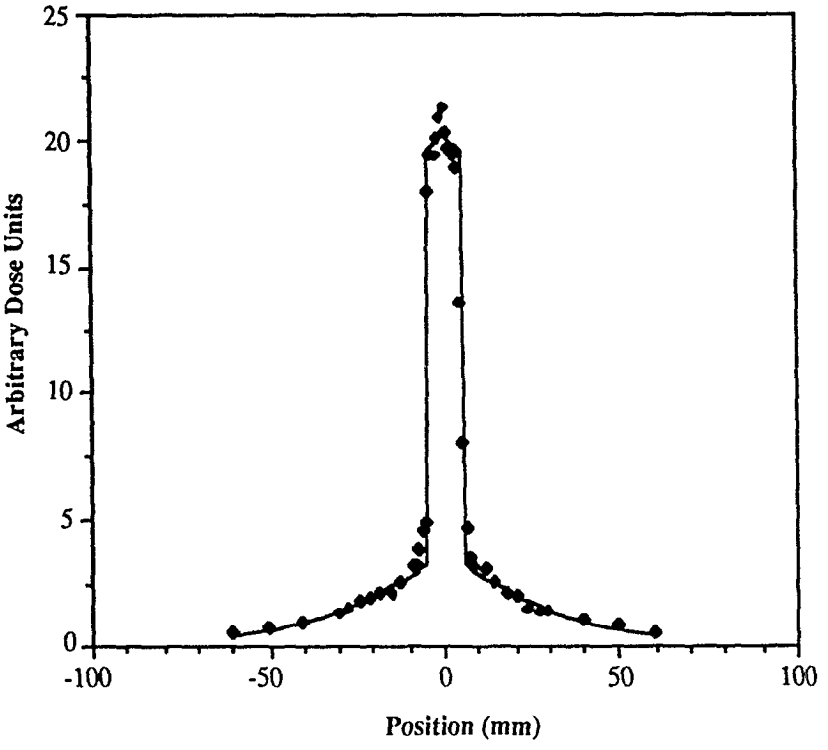

Figure 2: Measured TLD dose profile data at 1 cm depth in a head dosimetry phantom, measured with a third generation CT system, with a 10 mm slice width. The solid line represents the fitted curve. (Data courtesy of Center for Devices and Radiological Health, US FDA).

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Appendix A

SPECIFIC TECHNICAL AND PERFORMANCE INFORMATION FOR CT SCANNER BID SUBMISSION*

Manufacturer: _____

Model: _____

Address: _____

_____

_____

_____

Phone: (___)_____

Response prepared by:

Name: _____

Title: _____

Authorized Signature: _____

Date: _____

* Use one set of forms for each model bid.

A. SYSTEM ENVIRONMENTAL REQUIREMENTS

1. Electrical Power Sources: List voltage, power, and phasing for each; indicate locations on architectural drawings.

_____

_____

2. Power Conditioning: Give manufacturer and model numbers of power conditioning system provided:

_____

3. Air Conditioning Requirements:

Control Area: _____ BTU/hr

Gantry Area: _____ BTU/hr

Computer Room: _____ BTU/hr

Other _____: _____ BTU/hr

4. Mechanical Requirements:

a. Areas where raised "computer floor" is required:

_____

_____

b. Under-floor cable runways required: (Specify depth, width and locations on architectural drawings)

c. Total weight of equipment: _____ lb. (kg)

Gantry: _____ lb. (kg)

Control Console: _____ lb. (kg)

HV Generator & Controller: _____ lb. (kg)

Computer System : _____ lb. (kg)

Other _____ lb. (kg)

d. Minimum floor space required (entire system): _____ sq.ft.(m^2)

5. Plumbing Requirements:

a Number of drains required*: _____

b. Number of water inlets required*: _____

*Specify location, flow rate, temperature range, etc., on architectural drawings.

6. Physical modifications: Specify the extent to which facility modifications will be performed by the vendor, with respect to installation of electrical troughs, plumbing, clinical power, air conditioning, etc.

_____

_____

_____

7. Radiation Protection: Specify measured maximum exposure rate 1 meter in any direction from scan isocenter, for widest slice width and highest kVp using a cylindrical tissue equivalent phantom at least 20 cm in diameter.

Kilovoltage: _____ kVP

Slice width: _____ mm

Phantom diameter: _____ cm

Phantom material: _____

Air kerma: _____ mGy/mAs (mR/mAs)

B. SYSTEM CHARACTERISTICS

1. X-ray Generator:

a. Voltage waveform: Continuous: _____

pulsed: _____

b. kVp settings available (List):

_____

c. mA (mAs) stations available (list for each kVp):

settings at _____ kVp _____

settings at _____ kVp _____

settings at _____ kVp _____

settings at _____ kVp _____

d. Available Scan Times: Time Scan Angle

_____ s _____ °

_____ s _____ °

_____ s _____ °

_____ s _____ °

_____ s _____ °

_____ s _____ °

_____ s _____ °

_____ s _____

2. X-ray Tube:

a. Type: Rotating anode: _____

Stationary anode: _____

b. Focal spot sizes (Nominal):	*Scan Plane Dimension*	*Axial Dimension*
Focus #1	_____ mm	_____ mm
Focus #2	_____ mm	_____ mm

c. X-ray beam filtrations (operator variable - include both hardening filters and beam flattening or bow tie filters).

Material	***Thickness****	***Intended Use***

* Specify for hardening filters only.

d. Thermal Characteristics:

Housing cooling rate: _____ J/min

Anode cooling rate: _____ J/min

Anode heat storage capacity (cold): _____ J

Housing heat storage capacity: _____ J

Type of thermal overload protection system provided:

_____

e. Does x-ray tube employ a mechanical shutter? _____

3. Beam Collimation System:

a. List all available (nominal) slice thicknesses in mm:

_____

b. Slice width settings where prepatient collimator is adjustable in axial dimension _____.

c. Slice width settings where prepatient collimator is fixed in axial dimension _____.

d. Slice width settings where postpatient collimator is adjustable in axial dimension _____.

e. Slice width settings where postpatient collimator is fixed in axial dimension _____

4. Gantry:

a. Type of Scan Motion:

Rotate/translate: _____

Symmetric fan beam, rotating detectors:

Asymmetric fan beam, rotating detectors:

Fan beam, stationary detector ring:

Fan beam, nutating detector ring: _____

Other: _____

b. Variable geometric magnification available? _____

c. Continuous rotation available? _____

d. Gantry Aperture:

Maximum gantry aperture diameter: _____ cm

Maximum scan (sampled) diameter. _____ cm

e. Gantry Tilt (maximum):

Gantry top toward table: _____ °

Gantry top away from table: _____ °

Angulation accuracy: ± _____ °

f. Light-field Localizer:

Type: Laser: _____

Focused Light Beam: _____

Configuration: Transaxial : _____

Sagittal: _____

Coronal: _____

Position of transaxial localizer:

At scan plane: _____

External to scan aperture: _____

Accuracy of transaxial localizer* ± _____ mm

* Coincidence of light and x-ray field centers.

5. Patient Scanning Table

a. Maximum motions:

Longitudinal (full out to full in): _____ cm

Accuracy of table incrementation* ± _____ mm

Reproducibility*

*(Table loaded with 180 lb (80 kg) ± _____ mm

Minimum table height: _____ cm

Maximum table height: _____ cm

b. Location(s) of table position indicators:

Gantry: _____

Table: _____

Control console: _____

Scan image: _____

c. Table detachable from gantry? _____

Specify cost if optional: $_____

Cost of extra beds: $_____ ea

d. Table tilt (maximum): Head end up: _____ °

Head end down: _____ °

Angulation accuracy: - -

6. Detectors

a. Type:

Scintillator/photodiode: _____

Scintillator/PM tube: _____

Type of scintillator: _____

Pressurized xenon: _____

Other: _____

b. Number (exclude reference detectors): _____

c. Efficiency:

Scan Mode	*kVp*	*Geometric (%)*	*Total (%)*

d. Data sampling:

Scan Time	***# Projections***	***#Ray Samples****
_____ S		
_____ S		
_____ S		
_____ S		
_____ S		

* Give all values if independently variable.

e. Recommended calibration frequency:

"Air calibration" scans: _____

"Water calibration" scans: _____

7. Computer System:

a Image reconstruction time: (measure from scan start to completion of display, i.e., include scan time)*.

Scan Mode	*Reconstruction Matrix**	*Scan Time*	*FOV*	*Reconstruction Time*
Standard Head	_____	_____ s	_____ cm	_____ s
Standard Adult Body	_____	_____ s	_____ cm	_____ s
Highest Resolution	_____	_____ s	_____ cm	_____ s
FastestScan	_____	_____ s	_____ cm	_____ s

*Indicate when display matrix differs from reconstruction matrix.

b. Faster reconstruction options (Specify)

Option:_____ $_____

Performance (Optional conditions): *Scan Mode*	*Reconstruction Time*
Standard Head	_____ s
Standard Adult Body	_____ s
Highest Resolution	_____ s
Fastest Scan	_____ s

c. Simultaneous reconstruction and scanning? _____

d. Data storage and image archiving:

Device	MBytes	Storage Capacity* 512² Images	256² Images	Raw Data Files
Magnetic tape				
Magnetic tape				
Fixed disc drive				
Fixed disc drive				
optical disc				

* Uncompressed data files

List optional storage devices and additional cost:

_____

_____ $_____

Nondestructive data file compression available?

Compression ratio(s): _____

e. Convolution kernels (reconstruction filter functions):

Name	*Design Purpose*

Convolution kernels (continued):

Name *Design Purpose*

g. Image display system:

Pixels displayed (entire screen): Horizontal

Vertical

Image screen size (diagonal):

Operator's console: _____ in(cm)

Physician's console: _____ in(cm)

Gray scale bar displayed?

Alphanumeric information displayed
(Check where appropriate):

	On Image	***On Separate Data Screen***
Patient's name:	_____	_____
ID number:	_____	_____
Age:	_____	_____
Sex:	_____	_____
Date of exam:	_____	_____
Tie of exam:	_____	_____

Alphanumeric information displayed (cont.):	on Image	***On Separate Data Screen***
Slice #	_____	_____
kVp:	_____	_____
mA(s):	_____	_____
Scan time:	_____	_____
Slice width:	_____	_____
Bed position :	_____	_____
Bed increment:	_____	_____
Convolution kernel:	_____	_____
Gantry tilt angle:	_____	_____
Body side (R/L):	_____	_____

h. Diagnostic software features (check if standard, give cost if optional).

Feature	***Standard***	***Cost****
Square region-of-interest (ROI):	_____	$_____
Rectangular ROI:	_____	$_____
Circular ROI:	_____	$_____
Arbitrarily shaped ROI :	_____	$_____
Average CT number within ROI :	_____	$_____
Std. deviation of CT number:	_____	$_____
Histogram of CT numbers within ROI :	_____	$_____

Diagnostic Software Features (continued).

Feature	*Standard*	*Cost**
Distance measuring utility:	_____	$_____
Accuracy:	± ____ mm	
Grid overlay:	_____	$_____
Profile utility (CT number plot between image points):	_____	$_____
Highlighting of pixels within specific CT number range:	_____	$_____
Multiple image display (e.g., 2x2, 3x3):	_____	$_____
Gray scale inversion :	_____	$_____
Image reversal (left to right):	_____	$_____
Image inversion (top to bottom):	_____	$_____
Subtraction of two images:	_____	$_____
Reconstruction magnification (arbitrary FOV within limits):	_____	$_____
Non-reconstruction magnification:	_____	$_____
High density artifact removal:	_____	$_____
Programmable window settings:	_____	$_____
Multi-planar reconstruction:	_____	$_____
Arbitrary angle reconstructions:	_____	$_____
Dual windowing (simultaneous display of two CT number ranges):	_____	$_____

Diagnostic Software Features (continued).

Feature	*Standard Cost**
Three dimensional image display:	$
Surface rendering.	$
Transparency rendering:	$
Bone mineral density measurement?	$
Dual energy material decomposition:	$
Xenon (cerebral blood flow) imaging:*	$
cardiac gating:*	$
Radii therapy treatment planning:*	$
Compiler (Fortran, C, etc.) for research programming:	$
ACR/NEMA image transfer interface:*	$
Gamma correction to match CRT phosphor to sensitivity curve of film?	$
SMPTE pattern for QA:	$
Other features* (list):	
_____	$
_____	$
_____	$

*Include cost of additional hardware required.

8. Hardware Accessories: (check if standard, give cost if optional).

Feature	*Standard Cost**
Head holder:	$
Infant holder	$
Flat (radiation therapy simulation) table insert:	$
Other: (specify) _____	$
	$

9. Radiographic Scan Made:

a. Projections available: AP:

Lateral:

Arbitrary angle:

b. Maximum scan dimensions (at gantry axis):

length: mm

width : _____ mm

c. Software for scan localization from radiograph: _____

Localization of slice positions:

Accuracy: _____ mm

Localization of gantry (table) tilt:

Accuracy: ± °

10. Hard Copy Images:

a Standard multiformat camera provided: (Manufacture, model)

_____

b. Film sizes and display formats:

Film size(s)	***Display Format***
8” x 10” (20 cm x 25 cm)	1 on 1 _____
	4 on 1 _____
	9 on 1 _____
	other: _____
10” x 12” (25 cm x 30 cm)	1 on 1 _____
	4 on 1 _____
	9 on 1 _____
	other: _____
14” x 17” (35 cm x 43 cm)	4on1 _____
	9 on 1 _____
	16 on 1 _____
	other: _____
Other film size: _____	_____
	_____
	_____

c. Optional hard copy imaging devices available:

Device	*Cost*
_____	$_____
_____	$_____
_____	$_____

11. System Performance:

a Specification of Performance Data:

Spatial Resolution: Measured in cycles/cm at an MTF of 10%.

Image Noise: Measure within an ROI of ≈1 cm^2 centered within a 15-21 cm diameter cylindrical water phantom for head and pediatric scans, and a 30-32 cm phantom for adult body scans. Express as a percent of the effective linear attenuation coefficient of water, corrected for the scanner contrast scale[31].

Radiation Dose: Specify all dose data in cGy (rads) as either multiple scan average dose (MSAD) or computed tomography dose index (CTDI), check as appropriate:

CTDI _____

MSAD _____

Doses must be measured at a radial depth of 1 cm in acrylic phantoms meeting specifications of the U.S. CDRH (FDA)[31]. For all 360° scans measure at the 12 o'clock position in the phantom. Measure at mid scan arc for scans <360°, and at midpoint of overlap region for scans >360°.

Performance Conditions:

Scan Mode	*Reconstr. Matrix*	*FOV (cm)*	*Convol. Kernel*	*kVp*	*Scan Time*	*mAs*	*Slice Width*
Std.Head	_____	_____	_____	_____	_____	_____	_____
Std. Adult Body	_____	_____	_____	_____	_____	_____	_____
Best Resolution	_____	_____	_____	_____	_____	_____	_____

Performance Conditions (cont):

scan Mode	*Reconstr. Matrix*	*FOV (cm)*	*Convol. Kernel*	*kVp*	*scan Time*	*mAs*	*Slice Width*
Fastest scan							
Lowest Noise Body							
Lowest Noise Head							
Pediatric Head							
Pediatric Body							

Performance Specifications:

ScanMode	*Resolution cycles/cm*	*Noise % SD*	*Dose (cGy)*
Std. Head			
Std. Adult Body			
Best Resolution			
Fastest Scan			
Lowest Noise Body			
Lowest Noise Head			
Pediatric Head			
Pediatric Body			

c. Collimation performance: Measure sensitivity and radiation profiles in mm at full width half maximum (FWHM) within a radius of 5-15 cm of gantry axis. Tolerances should reflect manufacturer's range of acceptable error.

Nominal Slice Setting	*Sensitivity Profile Width Tolerance*	*Radiation Profile Width Tolerance*
(min) _____	_____ ± _____	_____ ± _____
_____	_____ ± _____	_____ ± _____
_____	_____ ± _____	_____ ± _____
_____	_____ ± _____	_____ ± _____
_____	_____ ± _____	_____ ± _____
_____	_____ ± _____	_____ ± _____
(max) _____	_____ ± _____	_____ ± _____

C. ADMINISTRATIVE DETAILS

1. Warranties:

a. Warranty Period (months beyond formal acceptance): _____

Exclusions:

x-ray tubes* _____

Other exclusions (specify):

_____

_____

*If excluded, give additional cost of x-ray tube warranty during base warranty period: $ _____

b. Normal service hours: _____ AM to PM,

_____ (day) through _____ (day).

2. Down Time:

a. Definition: Down time is defined as time when the scanner is unavailable for patient use due to failure of critical hardware or software component(s). Down time is defined over the base time

period from _____ AM to _____ PM,

from _____ (day) through _____ (day).

Excludes time for required preventive maintenance, component failure directly resulting from inadequate (owner- supplied) preventive maintenance or operation beyond performance specifications.

b. Guarantee: Down time shall not exceed _____ % of the base time period over any calendar month of the warranty period.

c. Penalty: The warranty period will be extended by _____ days for every 1% of down time beyond the guaranteed minimum.

3. Required Preventive Maintenance:

_____ hrs per week

_____ hrs every two weeks

_____ hrs per month

4. Service Contracts: (Use plans B and C as necessary for optional Contracts)

Plan A (check all that apply)

All parts excluding x-ray tubes: _____

x-ray tubes: _____

All labor from 8:00 AM to 5:00 PM Monday through Friday: _____

Night labor: between PM and AM, Monday through Friday: _____

Weekend and holiday labor _____

Cost: Year 1 after warranty: $_____

Maximum annual increase in years 2-5 after acceptance: _____%

Plan B (Check all that apply):

All parts excluding x-ray tubes: _____

x-ray tubes: _____

All labor from 8:00 AM to 5:00 PM Monday through Friday: _____

Night labor: between PM and AM, Monday through Friday: _____

Weekend and holiday labor: _____

Cost: Year 1 after warranty: $_____

Maximum annual increase in years 2-5 after acceptance: _____%

Plan C (Check all that apply):

All parts excluding x-ray tubes: _____

x-ray tubes: _____

All labor from 8:00 AM to 5:00 PM Monday through Friday: _____

Night labor: between _____ PM and ____ AM,
Monday through Friday: _____

weekend and holiday labor _____

Cost Year 1 after warranty: $_____

Maximum annual increase in years 2-5 after acceptance: _____%

5. Maximum Service Response Time (normal business hours):

_____ hrs.

6. Other Users:

If possible, provide list of names, addresses, telephone numbers and a contact person for 3 purchasers of the CT scanner model bid in this document.

Name : _____

Address: _____

_____

_____

Telephone No.: _____

Contact Person: _____

Name : _____

Address: _____

_____

_____

Telephone No.: _____

Contact Person: _____

Name : _____

Address: _____

_____

_____

Telephone No.: _____

Contact Person: _____

Appendix B

Phantoms for Acceptance Testing of X-Ray Transmission CT Scanning Systems

It was not the intent of the Task Group on Acceptance Testing of CT Scanners to devise a "new" AAPM phantom. However, for certain tests, commercial phantoms are not always adequate or are not available. The phantoms and test objects described in this appendix include generic objects which will suffice for certain tests, as well as test objects designed specifically for this document.

A. CYLINDRICAL UNIFORMITY PHANTOMS:

Used for noise and uniformity measurements. All are constructed as hollow right circular cylinders, at least 2 cm in depth. Phantoms may be constructed of solid acrylic or other water simulating plastic, or may be hollow and filled with distilled water. If special phantoms are constructed, hollow phantom walls should be made of 0.5-l cm thick methyl methacrylate (acrylic) or polycarbonate, with a fill space at least 2 cm in depth. Phantom construction should exclude screws, large irregularities in wall thickness, and any artifact-producing high atomic number materials. A head phantom should have a diameter of 15-21 cm and a body phantom should have a diameter of 30-32 cm, Head and body water phantoms provided with the scanner usually suffice. An optional "baby" phantom with a diameter of 8 cm is useful for systems with large pediatric caseloads. For an inexpensive alternative, thin walled, water filled disposable plastic bottles can be used for uniformity phantoms and are often available in appropriate diameters.

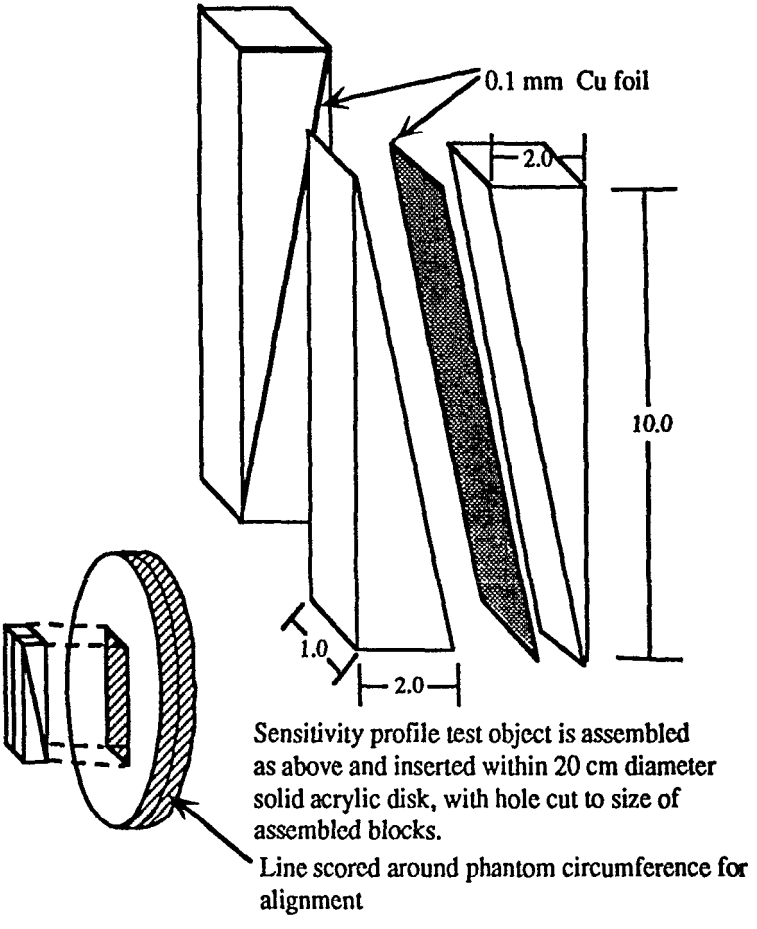


Figure B-l: Detail of sensitivity profile test object (all dimensions are in cm unless otherwise indicated).

B. SENSITIVITY PROFILE PHANTOM:

This phantom incorporates a pair of 0.1 mm copper foils, inclined at an inclination ratio of 5:1 and embedded in solid acrylic blocks. As shown in Figure B-l, these blocks can be constructed as four identical and individually machined blocks, then assembled as shown, sandwiching the copper foil between. The assembled blocks are then placed within an acrylic cylinder, with a rectangular cutout to receive them. Make an annular score mark en-

circling the outer circumference of the cylinder, and paint the edge white to facilitate alignment with scan alignment lights.

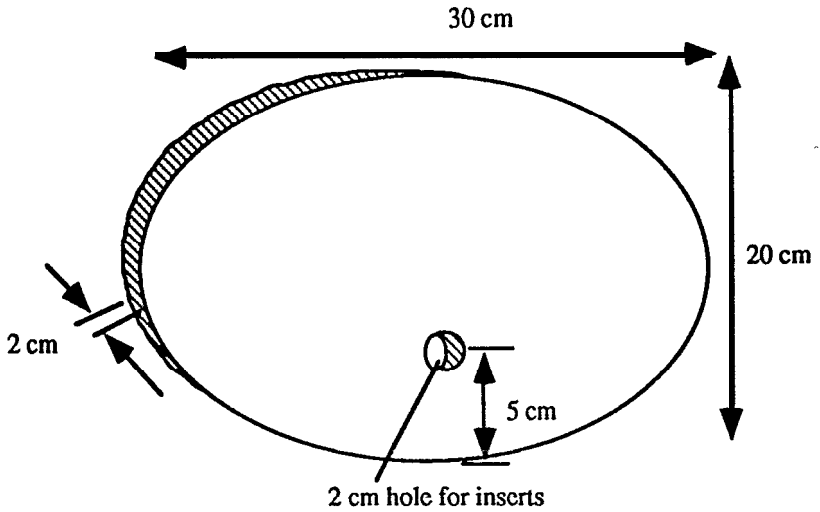


Figure B-2: Elliptical Uniformity Phantom

C. ELLIPTICAL UNIFORMITY PHANTOM:

Used for uniformity measurements. This phantom is constructed as a right elliptical cylinder, with major diameters of 20 and 30 cm, as shown in Figure B-2. Two designs are acceptable. It may be constructed as a hollow water filled design with construction similar to water phantoms, and a thickness of the water space of at least 2 cm. Alternatively, it can be constructed from a solid piece of acrylic or water equivalent material (e.g., Solid Water™ Radiation Measurements Inc., Middleton, WI) with a total thickness of at least 2 cm. Optionally, for bone mineral analysis, the phantom may have a slightly oversize ≈2 cm diameter cylindrical hole placed as shown in Figure B-2. A set of inserts should then be prepared with solutions of 0, 50, 100, and 200 mg/ml of K_2HPO_4. Alternatively the inserts could be constructed of trabecular bone equivalent plastic. An acrylic plug should be prepared to fill the hole when quantitative inserts are removed.

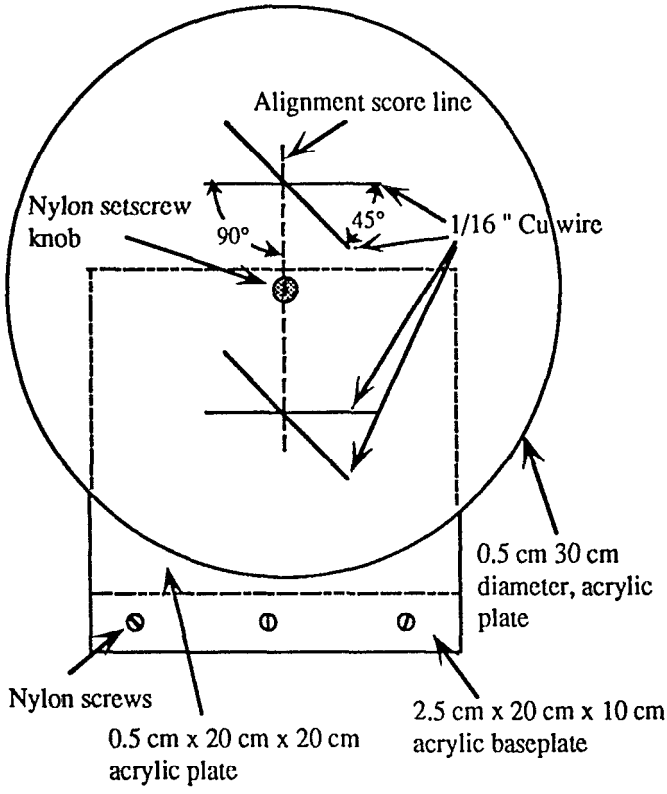


Figure B-3: Localization Image Angulation Test Object

D. LOCALIZATION IMAGE ANGLE TEST OBJECT:

This object, shown in Figure B-3, is designed with two pairs of thin crossed copper or steel wires on the surface of a 1 cm thick, 30 cm diameter acrylic disk. The disk is mounted on a central pivot so that it can be tilted to any arbitrary angle. The baseplate is also made of acrylic with nylon mounting screws. Wire length should be 8-10 cm and wires should be carefully aligned so that intersections of both pairs are on the same diameter, at a radius of 10 cm. Wire angles should be as depicted in Figure B-3, and may be placed in surface grooves in the acrylic disk to aid placemcnt.

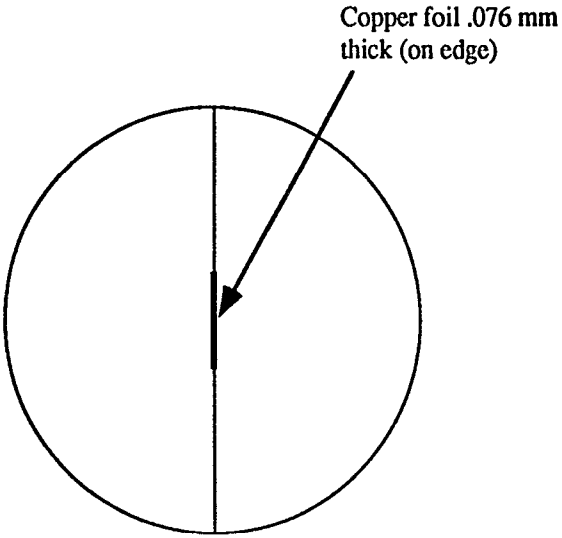


Figure B-4: LSF Test Object

E. LSF TEST OBJECT:

This test object is constructed as a disk of solid acrylic which sandwiches a .076 mm (0.003") Cu foil between the two halves so that the foil is oriented orthogonal to the scan plane. The foil should be as wide as the thickness of the disk and 3 cm long. The object is constructed from a rectangular block of acrylic, 1.5-2.5 cm thick, with sides of 20.5 cm and 20 cm. The block is cut in half, cut line parallel to the short side, and the cut surfaces are machined smooth. The copper foil is centered and glued to one of the cut surfaces of the halved block. The block is then glued back together with the foil sandwiched between. The assembly is machined into a right circular disk, 20 cm in diameter. The completed disk is shown in Figure B-4.

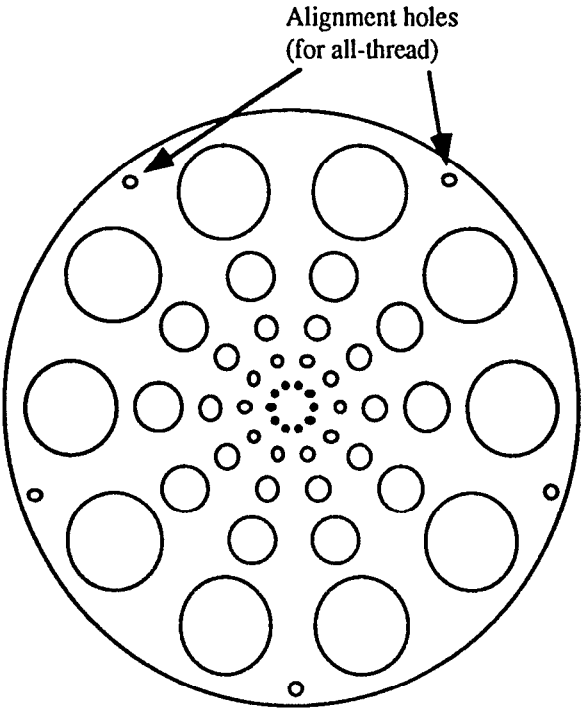


Figure B-5: Rose Phantom for CT
Detail of fifth thin sheet of acetate shown, (see text).

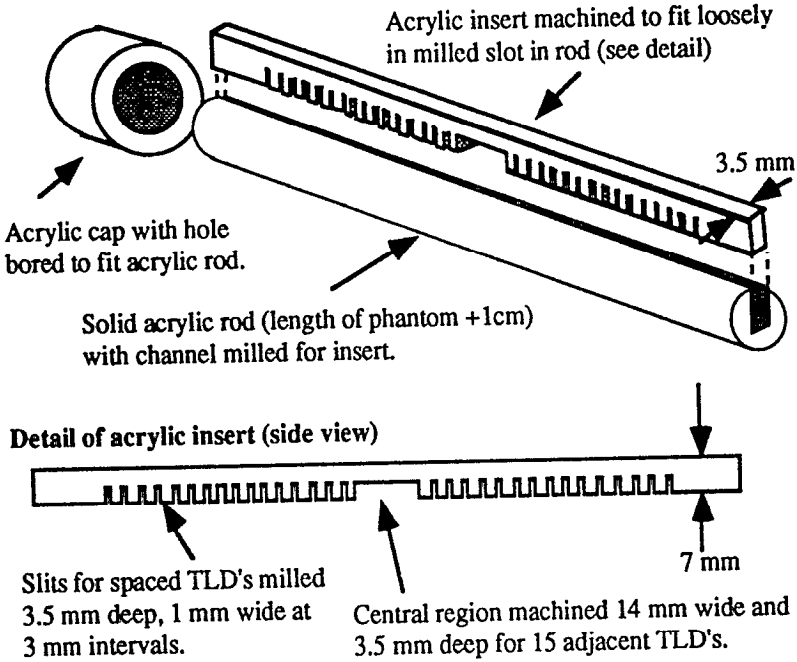
F. ROSE PHANTOM TEST OBJECT:

This test object is constructed as a stack of cellulose acetate sheets, cut into 20 cm diameter circles sandwiched between a pair of l/8", 20 cm diameter acrylic disks (to prevent the acetate from curling). Most of the test object is composed of a stack of 40 sheets of .02" (.5 mm) acetate. In the midst of this stack is a stack of 5 sheets, each .004" (0.1 mm) thick acetate. The entire stack of acetate sheets sandwiched between the acrylic disks is aligned and assembled. A series of 5 alignment holes are drilled through the stack around the periphery. The stack is disassembled and the thinner sheets are removed. Align and clamp the thin sheets between two pieces of acrylic with pins through two of the alignment holes. Mark sector hole locations, as in Figure B-5 on the acrylic sheet (becomes a template). The intent is to drill a set of live holes grouped into 10 pie shaped sectors, in a radial array

as shown in Figure B-5. The hole sizes within each sector are: 1 mm, 2 mm, 4 mm, 8 mm, 16 mm and 32 mm. One of the thin sheets has all sectors drilled, the fourth has eight drilled, the third six, the second four and the first two. This is done by putting the entire stack of 5 thin sheets in the template and two opposing sectors are drilled each with the live holes as shown. The top sheet is removed and the next pair of opposing sectors is drilled. This is repeated until all sectors are drilled in the final single sheet The thin sheets are then reassembled, with the holes carefully aligned. The thin stack is then sandwiched in the middle of the thick stack of acetate sheets (20 on each side), which is in turn placed between the acrylic plates. Put nylon all-thread through the alignment holes and fasten with nylon nuts.

G. DOSIMETRY PHANTOMS:

These phantoms are right circular cylinders of solid polymethyl methacrylate (density = 1.19 ± 0.01 g/cm^3), with lengths of 14-16 cm and diameters of 16 and 32 cm for the "head" and "body" phantoms, respectively (available from several dosimeter and radiological accessory manufacturers). A pediatric phantom is suggested here, for situations where pediatric scans predominate. This "infant" phantom is otherwise identical to the head and body phantoms, but with a diameter of 8 cm. All phantoms are drilled through their length for placement of dosimeters at different coaxial locations. All phantoms have holes at a depth of 1 cm (center to phantom edge) from the outer surface, at the 3, 6, 9 and 12 o'clock positions, and along the cylinder axis. Dosimeter holes at other radial locations may also be provided. Acrylic plugs provided for each hole location are removed when dosimeters are inserted.



Assembly:

1. Cement acrylic rod (without insert) into end cap.
2. Insert slides along channel to allow TLD's to be individually inserted and extracted.
3. Once loaded, insert is retained in slot at one end by cap, tape or machine screw can be used at opposite end.

Figure B-6: TLD Holder For Dosimetry Phantoms

1. TLD Holder

A specially designed TLD holder suitable for sampling over a 14 cm axial distance within dosimetry phantoms is shown in Figure B-6. The holder packs 15 abutted chips over the central 14 mm of the slice, towards the edges, chips are spaced 3 mm apart. The overall length of the assembled rod (less cap) should be equivalent to that of the phantom so that the abutted chips region of the holder is centered within the phantom.

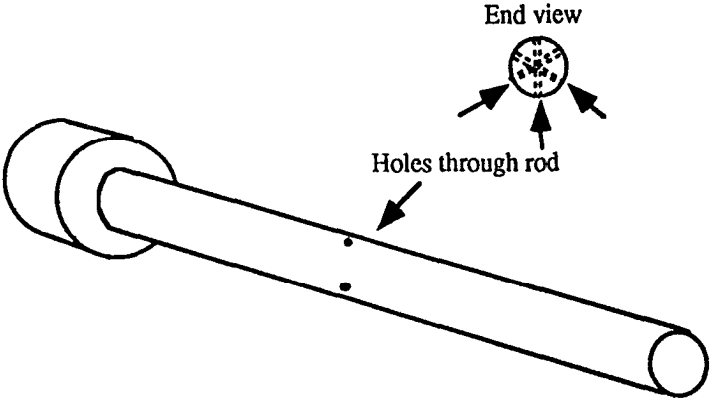


Figure B-7: TLD Alignment Rod for Dosimetry Phantoms

2. TLD Holder Centering Rod:

This rod (Figure B-7) is identical to the acrylic plugs included with the dosimetry phantoms but at mid length there are three transverse l/16" holes 120° apart. The central hole is exactly at mid-length, while the other two holes are positioned 2.5 mm (center to center) to either side.