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Standard Guide for Computed Tomography (CT)[1]

This standard is issued under the fixed designation E1441; the number immediately following the designation indicates the year of original adoption or, in the case of revision, the year of last revision. A number in parentheses indicates the year of last reapproval. A superscript epsilon (ε) indicates an editorial change since the last revision or reapproval.

This standard has been approved for use by agencies of the U.S. Department of Defense.

1. Scope*

1.1 CT is a radiographic examination technique that generates digital images in three dimensions of an object, including the interior structure. Because of the relatively good penetrability of X-rays, CT permits the nondestructive physical and, to a limited extent, chemical characterization of the internal structure of materials. Also, since the method is X-ray based, it applies equally well to metallic and non-metallic specimens, solid and fibrous materials, and smooth and irregularly surfaced objects.

1.2 This guide is intended to satisfy two general needs for users of industrial CT equipment: (1) the need for a tutorial guide addressing the general principles of X-ray CT as they apply to industrial imaging; and (2) the need for a consistent set of CT performance parameter definitions, including how these performance parameters relate to CT system specifications.

1.3 This guide does not specify CT examination techniques, such as the best selection of scan parameters, the preferred implementation of scan procedures, or the establishment of accept/reject criteria for a new object.

1.4 *Units*—No units are mentioned in this document. However, for CT, values are typically stated in SI units and are regarded as standard.

1.5 *This standard does not purport to address all of the safety concerns, if any, associated with its use. It is the responsibility of the user of this standard to establish appropriate safety, health, and environmental practices and determine the applicability of regulatory limitations prior to use.*

1.6 *This international standard was developed in accordance with internationally recognized principles on standardization established in the Decision on Principles for the Development of International Standards, Guides and Recommendations issued by the World Trade Organization Technical Barriers to Trade (TBT) Committee.*

2. Referenced Documents

2.1 *ASTM Standards:*[2]

E746 Practice for Determining Relative Image Quality Response of Industrial Radiographic Imaging Systems

E1316 Terminology for Nondestructive Examinations

E1570 Practice for Fan Beam Computed Tomographic (CT) Examination

E1695 Test Method for Measurement of Computed Tomography (CT) System Performance

E1935 Test Method for Calibrating and Measuring CT Density

E2698 Practice for Radiographic Examination Using Digital Detector Arrays

E2736 Guide for Digital Detector Array Radiography

2.2 *ISO Standards:*[3]

ISO 15708-1:2017-02 International Standard for Non-destructive Testing - Radiation Methods for Computed Tomography - Part 1: Terminology

ISO 15708-2:2017-02 International Standard for Non-destructive Testing - Radiation Methods for Computed Tomography - Part 2: Principles, Equipment and Samples

ISO 15708-3:2017-02 International Standard for Non-destructive Testing - Radiation Methods for Computed Tomography - Part 3: Operation and Interpretation

ISO 15708-4:2017-02 International Standard for Non-destructive Testing - Radiation Methods for Computed Tomography - Part 4: Qualification

3. Terminology

3.1 *Definitions*—In addition to terms defined in Terminology E1316, the following terms are specific to this standard.

3.1.1 Throughout this guide, the term "*X-ray*" is used to denote penetrating electromagnetic radiation; however, electromagnetic radiation may be either X-rays or gamma rays.

3.2 *Definitions of Terms Specific to This Standard:*

[1] This guide is under the jurisdiction of ASTM Committee E07 on Nondestructive Testing and is the direct responsibility of Subcommittee E07.01 on Radiology (X and Gamma) Method.

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[2] For referenced ASTM standards, visit the ASTM website, www.astm.org, or contact ASTM Customer Service at service@astm.org. For *Annual Book of ASTM Standards* volume information, refer to the standard's Document Summary page on the ASTM website.

[3] Available from International Organization for Standardization (ISO), ISO Central Secretariat, BIBC II, Chemin de Blandonnet 8, CP 401, 1214 Vernier, Geneva, Switzerland, http://www.iso.org.

***A Summary of Changes section appears at the end of this standard**

3.2.1 *CT detectability, n*—the extent to which the presence of a feature or indication can be reliably inferred from a tomographic examination image.

3.2.1.1 *Discussion*—CT detectability is dependent on the spatial resolution and contrast resolution of the image. Features may be detectable even if they are too small to be resolved, provided their contrast after blurring is still sufficient.

3.2.2 *CT slice, n*—a tomogram or the object cross-section corresponding to it.

3.2.2.1 *Discussion*—The slice plane is the plane, determined by the focal spot and linear array of detectors or a single line of an area array, around which each measurement of a planar tomographic scan is centered. Each such scan also has a slice thickness, which is the distance normal to the slice plane over which changes in object opacity will significantly influence the measurements; typically, an average value based on the aperture function is used to characterize this parameter. When three-dimensional CT-density maps have been reconstructed, a slice may be formed on an arbitrary plane or other surface, not just on slice planes.

3.2.3 *CT view/projection, n*—a set of X-ray opacity projection values (derived from measurements or by simulation) grouped together for processing purposes, especially for the convolution and backprojection steps of computing a tomograph.

3.2.3.1 *Discussion*—The set of line integrals resulting from a scan of an object can be grouped conceptually into subsets referred to as views. Each view corresponds to a set of ray paths through the object from a particular direction. The views are also referred to as projections or profiles, while each individual datum within a given projection is referred to as a sample or often simply a data point

3.2.4 *detector aperture function, n*—a three-dimensional function centered on the axis from the radiation source to a detector element, giving the sensitivity of the detector to the presence of attenuating material at each position.

3.2.4.1 *Discussion*—The detector aperture function gives the extent and intensity distribution of each ray around and along the length of its central line. The function is determined by the size and shape of the radiation source and of the active region of the detector, and by relative distance to the source and the detector. The average width of this function in the region of the object being examined is an important limit on the spatial resolution of a CT scan.

3.2.5 *sinogram, n*—a two-dimensional array of position within view versus view angle, which can be stacked into a volume for volumetric reconstruction techniques.

4. Summary of Guide

4.1 This guide provides a tutorial introduction to the technology and principles of CT and is divided into six main sections. Section 5 discusses the significance of CT compared to conventional radiography. Section 6 provides a brief overview of CT. Section 7 describes the basic hardware elements of CT systems. Section 8 outlines the general principles of CT imaging. Section 9 outlines CT system performance factors used in characterizing the system performance and images. Section 10 identifies some CT system processes for quantitative measurements to compare the relative performance of different CT systems and calibrate for densitometric information. Note that ISO 15708 Parts 1-4 have been updated and this guide has been harmonized with the recent publications.

5. Significance and Use

5.1 Computed tomography (CT) is a radiographic reconstruction method that provides a sensitive technique whenever the primary goal is to locate and size planar and volumetric detail in three dimensions.

5.2 CT provides quantitative volume images as a function of density and element number (attenuation coefficient) by means of computer-processed combinations of many X-ray measurements taken from different angles to produce cross-sectional images of specific areas of a scanned object, allowing the user to see inside the object without cutting. CT is considered much easier to interpret than conventional radiographic data due to the elimination of overlapping structures. The new user can learn quickly to read CT data because the images correspond more closely to the way the human mind visualizes three-dimensional structures than conventional projection radiography. Further, because CT slices and volumes are digital, they may be enhanced, analyzed, compressed, archived, input as data into performance calculations, compared with digital data from other NDE modalities, or transmitted to other locations for remote viewing.

6. Computed Tomography (CT) Overview

6.1 CT is a radiographic examination method that uses a computer to reconstruct an image of one or more cross-sectional plane (slice(s)) through an object. The result is a quantitative map of the linear X-ray attenuation coefficient, μ, at each point in the plane. The linear attenuation coefficient characterizes the local instantaneous rate at which X-ray photons are attenuated during the scan, by scatter or absorption, from the incident radiation as it propagates through the object.

6.2 One particularly important property of the total linear attenuation coefficient is that it is dependent upon atomic number and is proportional to material density, which is a fundamental physical property of all matter. The fact that CT images are proportional to density is perhaps the principal virtue of the technology and the reason that image data are often thought of as representing the distribution of material density within the object being examined; however, the linear attenuation coefficient also carries an energy dependence that is a function of material composition (atomic number). This feature of the attenuation coefficient may or may not (depending on the materials and the energies of the X-rays involved) be more important than the basic density dependence. In some instances, this effect can be detrimental, masking the density differences in a CT image; in other instances, it can be used to advantage, enhancing the contrast between different materials of similar density.

6.3 The fundamental difference between CT and conventional radiography is shown in Fig. 1. In conventional radiography, information on the slice plane "P" projects into a

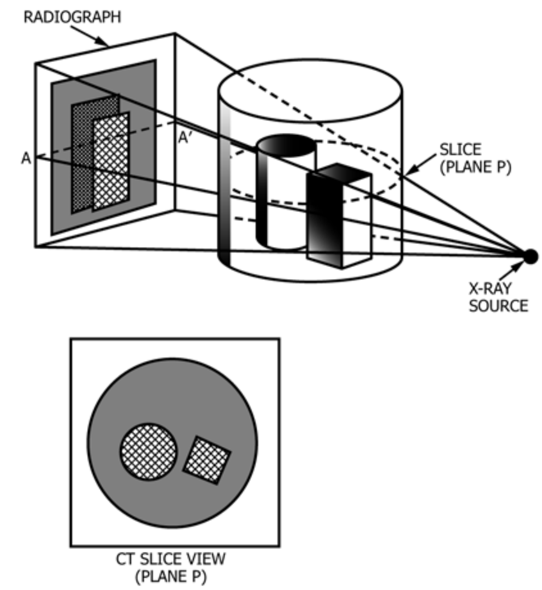

FIG. 1 A CT Image Versus a Conventional Radiograph

single line, "A-A;" whereas with the associated CT image, the full spatial resolution of the plane is preserved. CT information is derived from a large number of systematic observations at different viewing angles, and an image of the slice plane is then reconstructed with the aid of a computer. The image is generated in a matrix of voxels. The resultant map is an image of the object under examination. Thus, by using CT, one can, in effect, slice open the object under examination, examine its internal features, record the different attenuations, perform dimensional measurements, and identify any material or structural anomalies that may exist.

6.4 From Fig. 1, it can be appreciated readily that if an internal feature is detected in conventional projection radiography, its position along the line-of-sight between the source and the film is unknown. Somewhat better positional information can be determined by making additional radiographs from several viewing angles and triangulating. This triangulation is a rudimentary, manual form of tomographic reconstruction. In essence, a CT image is the result of triangulating every point in the plane from many different directions.

6.5 As with any modality, CT has its limitations. Candidate objects for examination must be small enough to be accommodated by the handling system of the CT equipment available to the user and radiometrically translucent at the X-ray energies employed by that particular system. In addition, the detector cannot be saturated in any view. This requirement can limit CT as compared to 2D radiography where one may saturate parts of the detected image in order to obtain proper signal through an area of interest where for CT, for a given view/projection image, the detector can only be saturated in an area outside the part.

6.6 *Types of CT:*

6.6.1 *Fan Beam CT (2D-CT)*—In a system which is able to perform a 2D-CT (or fan beam CT), a linear detector array (LDA) is being used. The projections needed for a reconstruction are typically collected by rotating the sample (or source and detector around the sample) at least 180° plus opening angle of the fan beam. With this kind of geometry, a single slice of object can be reconstructed. The object may be translated along the rotation axis to collect other slices.

6.6.1.1 Slice thickness is set by the X-ray optics of the system. It is a function of the object position (the magnification of the scan geometry) and the effective sizes (normal to the scan plane) of the focal spot of the source and the acceptance aperture of the detector. The effective size of the focal spot is determined by its physical size and any source-side collimation. The maximum thickness is achieved with the maximum effective focal spot size and the maximum effective acceptance aperture. The minimum thickness is achieved with the minimum focal spot size permitted and the minimum effective acceptance angle permitted. Due to the collimation (tube and detector) often used in this geometry, the influence of scattered radiation, which disturbs the reconstruction process significantly (artifacts), is minimal. Multiple slices along the axis of rotation may be stacked into a 3-D data set and rendered as a 3-D image. Two types of scan motion geometries are most common: translate-rotate motion and rotate-only motion.

6.6.1.2 *Translate-rotate Motion*—The object is translated in a direction perpendicular to the direction and in the plane of the X-ray beam. Full data sets are obtained by rotating the test object between translations by the fan angle of the beam and again translating the object until a minimum of 180° of data have been acquired. The main advantage of this design is ability to accommodate a wide range of different object sizes including objects too big to be subtended by the X-ray fan. The disadvantage is longer scan time. If the test object is larger than the prescribed field of view (FOV), either by necessity or by accident, unexpected and unpredictable artifacts or a measurable degradation of image quality can result.

6.6.1.3 *Rotate-only Motion*—A complete view is collected by the detector array during each sampling interval. A rotate-only scan has lower motion penalty than a translate-rotate scan and is attractive for industrial applications where the part to be examined fits within the fan beam and scan speed is important. If the test object is larger than the prescribed field of view (FOV), either by necessity or by accident, unexpected and unpredictable artifacts or a measurable degradation of image quality can result.

6.6.2 *Cone Beam CT (3D CT)*—In a system which is able to perform a 3D-CT (or cone beam CT), an area scan detector, typically a digital detector array (DDA), is used, where each row of the area array "acts" like a linear array. The projections needed for a reconstruction are typically collected by rotating the sample (or source and detector around the sample) at least 180° plus opening angle of the cone beam. With this kind of geometry, a volume with multiple slices of object can be reconstructed, Feldkamp reconstruction (a reconstruction process that reconstructs 3D data directly to a 3D image), with a single rotation of the object. Compared to the fan beam geometry, scattered radiation within the used cone cannot be reduced by collimation. Usually the image quality of the slices produced is worse in quality compared to the fan beam geometry. Additionally, the cone beam geometry produces another artifact, which depends on the opening angle of the cone parallel to the rotation axis. This artifact is widely known as Feldkamp artifact (see 9.7.6). The influence of the Feldkamp artifact can be reduced by minimizing the opening angle of the cone, which should be no larger than 11° in total if Feldkamp's reconstruction algorithm is used. The scanning and acquisition process for larger volumes is typically less time consuming than fan beam CT.

6.6.2.1 Mathematically only 180° plus the fan angle is required for CT reconstruction of the described method. It is, however, best practice to acquire a full 360° of data as this provides redundant information which fills in for detector defects (for example, bad pixels) and reduces artifacts. This also allows for checking the first versus the last image as a check for object motion during the scan.

6.6.2.2 *Extended Field of View/Offset Scanning*—Another method for increasing the width of the FOV while using a smaller area detector is to offset the position of the detector (or shift rotation axis, or a combination of both), collimate the beam asymmetrically, and scan the object. Each projection will be viewing a little more than half the total diameter of the field of view. For offset scanning, the center of rotation must always be within each projection (Fig. 2). This method requires 360° of data for mathematical sufficiency.

6.6.3 *Helical CT*—In a system which is able to perform Helical CT (or Spiral CT), a multi-row LDA or an area scan detector is being used. The projections needed for a reconstruction are typically collected by rotating the sample (or source and detector around the sample) combined with a linear movement along the rotation axis. With this kind of geometry, a volume with multiple slices of object can be reconstructed with multiple rotations of the object. Compared to the fan beam geometry, scattered radiation within the cone cannot be reduced by collimation. Usually the quality of the slices produced is worse in quality compared to the fan beam geometry but improved compared to the cone beam geometry. Contrary to cone beam CT, Helical CT avoids the Feldkamp cone artifact since all measured object details will pass the central plane. The acquisition process for larger volumes is less time consuming than fan beam CT. Helical CT can theoretically be used to measure infinitely long objects, like those produced in extrusion processes.

6.6.4 *Computed Laminography (Planar CT, Tomosynthesis, Coplanar Translational, Coplanar Rotational Laminography)*—Laminography is a radiographic technique in which the relative motion of the source, detector, and object show a specific plane more clearly. Some systems will utilize a reconstruction algorithm to assist in the image development. This method is especially suitable for large or flat objects that are either difficult to rotate or have geometries that make penetration from some angles impossible.

6.6.5 *Irregular (Optimized) Geometries for CT*—The quality of a CT volume is highly influenced by the geometry of the part. Due to the circular trajectories being used in most CT geometries unfavorable transmission directions are produced during the scan. Lack of penetration in these positions can lead to unwanted artifacts in the reconstruction. Collecting the projections on an arbitrary path (for example, optimized based on the object geometry) can avoid the unfavorable positions and therefore reduce artifacts. To make use of the arbitrary geometries, special reconstruction methods are necessary. In most cases, algebraic reconstruction is used, which can deal with any kind of geometry.

7. Basic Hardware Configuration

7.1 CT systems are composed of a number of subsystems, typically those shown in Fig. 3. The choice of components for these subsystems depends on the specific application for which the system was designed; however, the function served by each subsystem is common in almost all CT scanners. These subsystems are:

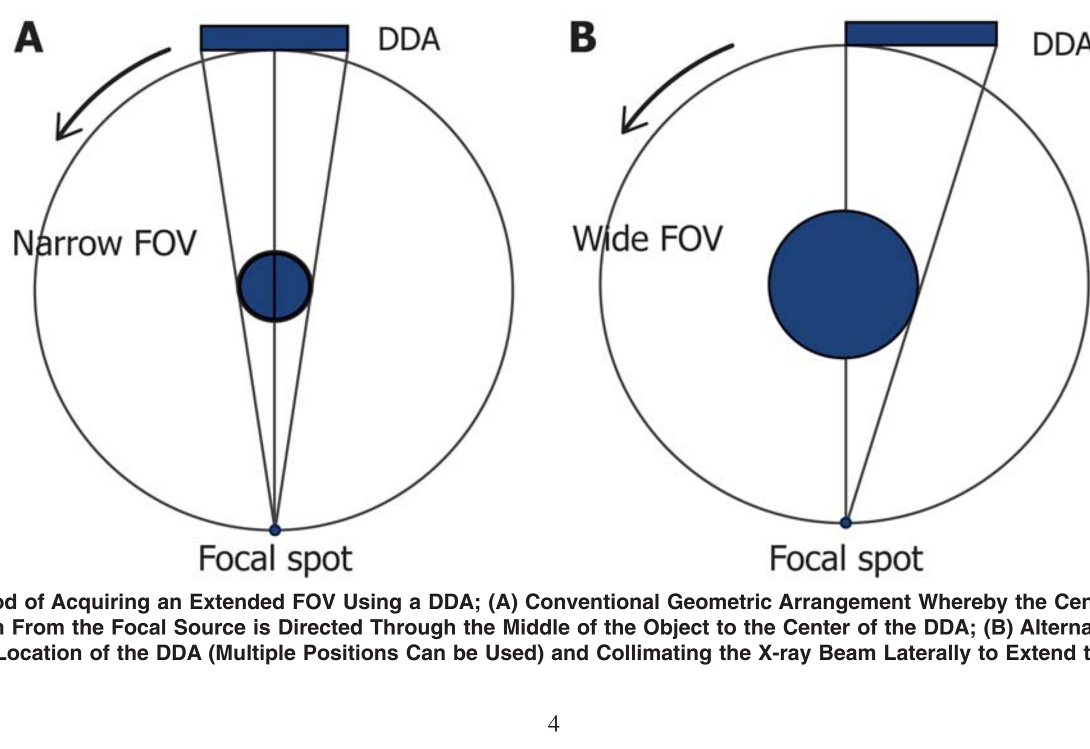

FIG. 2 Method of Acquiring an Extended FOV Using a DDA; (A) Conventional Geometric Arrangement Whereby the Central Ray of the X-ray Beam From the Focal Source is Directed Through the Middle of the Object to the Center of the DDA; (B) Alternate Method of Shifting the Location of the DDA (Multiple Positions Can be Used) and Collimating the X-ray Beam Laterally to Extend the FOV Object

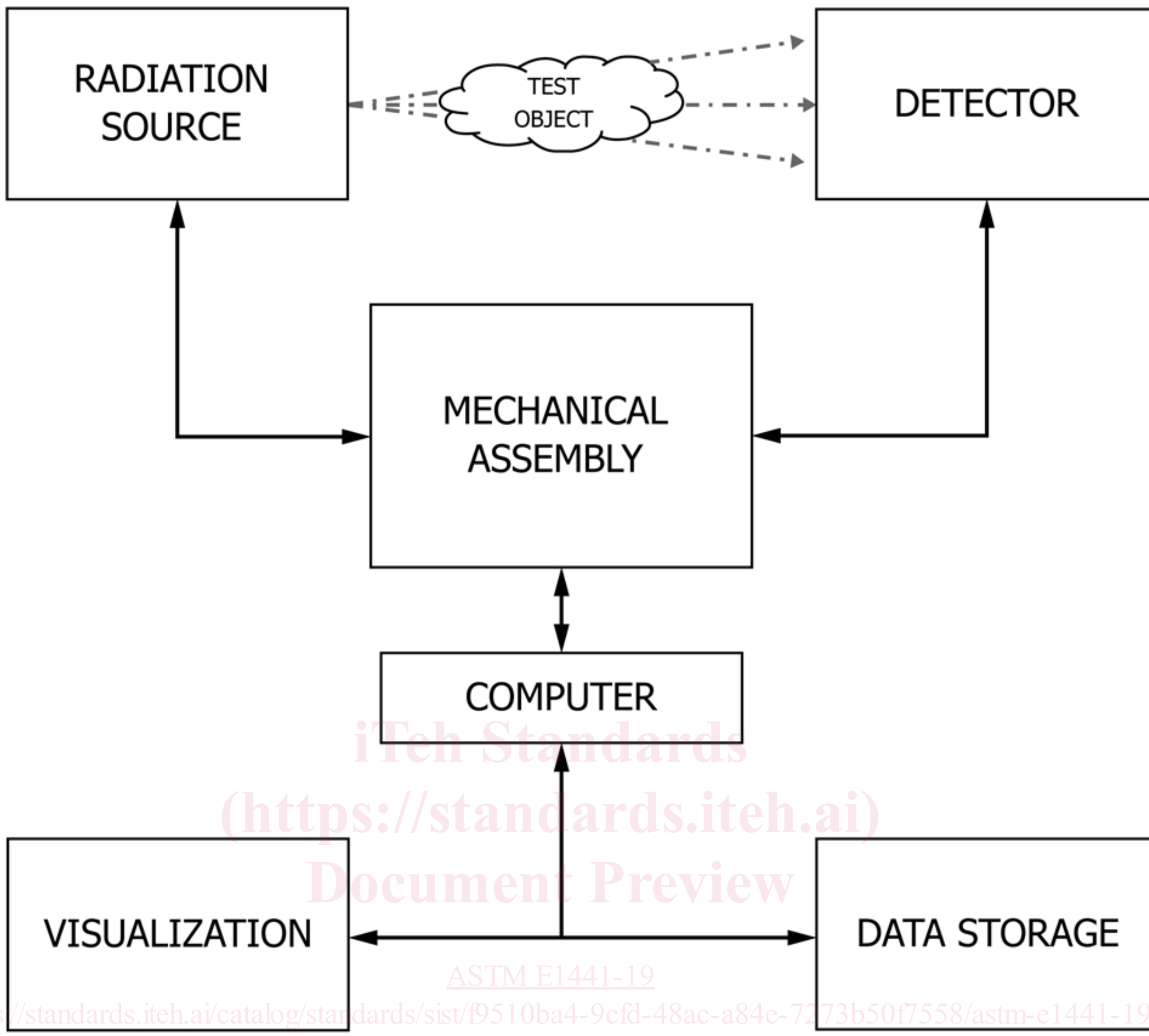

FIG. 3 Typical Components of a Computed Tomography (CT) System

7.2 *Source of Penetrating Radiation*—There are three rather broad types of radiation sources used in industrial CT scanners: (1) X-ray tubes, (2) accelerators, and (3) isotopes. The first two broad energy spectra are polychromatic (or Bremsstrahlung) electrical sources; the third is approximately monoenergetic radioactive sources. The choice of radiation source is dictated by the same rules that govern the choice of radiation source for conventional radiographic imaging applications.

7.2.1 *X-ray Sources*—While the CT systems may utilize either gamma-ray or X-ray generators, the latter is used for most applications. For a given focal spot size, X-ray generators (that is, X-ray tubes and linear accelerators) are several orders of magnitude more intense than isotope sources. Most X-ray generators are adjustable in peak energy and intensity and have the added safety feature of discontinued radiation production when switched off; however, the polychromaticity of the energy spectrum from an X-ray source causes artifacts such as beam hardening (the anomalous decreasing attenuation toward the center of a homogeneous object) in the image if uncorrected.

7.2.2 *Isotope Sources*—Isotope sources are attractive for some applications. They offer an advantage over X-ray sources in that problems associated with beam hardening are reduced or nonexistent for these monoenergetic/discrete energies type sources. They have the additional advantages, which are important in some applications, that they do not require bulky and energy-consuming power supplies, and they have an inherently more stable output intensity. The intensity of available isotopic sources, however, is limited by specific activity (decays per second per mass of 1 gram). The intensity affects signal-to-noise ratio, and, even more importantly, the specific activity determines source spot size and thus spatial resolution. Both of these factors tend to limit the industrial application of isotopic scanners. Nevertheless, they can be used in some applications in which scanning time or resolution is not critical.

7.2.3 *Source Setup*—Caution is advised against applying practices developed for projection radiography. Except at very high energies, mass attenuation differences between materials (signal contrasts) tend to decrease as the mean X-ray energy is increased; whereas, X-ray production and penetrability (signal levels) tend to increase under the same condition. Therefore, the optimum source energy for a given part is not determined by the lowest possible X-ray energy that provides adequate penetration but rather by the X-ray energy that produces the maximum signal-to-noise ratio (SNR). When a part consists of a single material or several materials with distinct physical density differences or different atomic numbers, or both, the best SNR may be obtained at a high source energy. In such cases, the decreased image noise at higher energies is more important than the increased contrast at lower energies. When chemically different components have the same or similar physical densities, the best discrimination of materials may be obtained at a low source energy. In such cases, the increased contrast at lower energies may be more important than the decreased image noise at higher energies. Use of beam filters near the source is a common way of optimizing the beam, similar to 2D radiography

7.3 *Radiation Detection Systems*—The detection system is a transducer that converts the transmitted radiation containing information about the test object into an electronic signal suitable for processing. Most systems convert X-rays to visible light in a scintillator and then detect the light with photo diodes. The detection system may consist of a single sensing element (single pixel detector), a linear detector array (LDA) of sensing elements, or a digital detector array (DDA) of sensing elements. The more detector pixels used at the same time, the faster the required scan data can be collected; but there are important tradeoffs to be considered. Single pixel detectors are used for parallel and fan-beam geometry. LDAs are used with fan-beam CT systems where the beam is collimated to a small slit to reduce scatter radiation. DDAs are used with cone-beam CT systems which can get a 3D volume image much faster than parallel and fan-beam geometry systems but are prone to scatter radiation which can be corrected by software.

7.3.1 A single pixel detector provides the least efficient method of collecting data but entails minimal complexity, eliminates detector cross talk and detector matching, and allows for collimation in two directions just to the active area. A very efficient scintillator for very high energy can be used as there is only a single pixel. This type of detector needs a mechanical movement and an exposure for each pixel of a projection, whereas an LDA needs only one exposure per projection. The CT result is a one slice image of the object.

7.3.2 Linear Detector Arrays (LDAs) have reasonable scan times at moderate complexity, acceptable cross talk and detector matching, and a flexible architecture (length of the detector and width of the collimator slit) that typically accommodates a collimation in one direction. A highly efficient scintillator with a large thickness compared to the pixel size can be used. Also, for high energies sufficient X-ray quanta are converted to light. An LDA generates one scanline per exposure. Rotating the object creates the sinogram. The CT result is a one slice image of the object. A 3D image could be created by generating multiple slices.

7.3.3 DDAs as area imaging devices can capture much more information in a single exposure than an LDA using the cone beam of the X-ray source. Collimation smaller than the active area of the DDA can be used to reduce scattered radiation and good practice is to collimate to the FOV needed for a given object. For uniform scintillator screens (Gd_2O_2S plastics, for example), the thickness of the scintillator screen will reduce the resolution when it exceeds the pixel size as the unsharpness from internal light scatter becomes bigger than the pixel size. However, a thinner scintillator converts less X-ray quanta—especially for higher energies—which results in a much lower SNR so use of a thicker scintillator or conversion screen, or both, may be desirable even though it reduces resolution. It should also be noted that in some instances a collimated 2D scintillator like CsI may mitigate this issue, particularly at lower energies. LDAs bypass this constraint by using single collimated pixels which allows for much thicker scintillators without resolution reduction. A DDA creates an area view; after rotating of the object by 360° a 3D volume image is the result. For more information about the properties of DDAs, and their constraints, refer to Guide E2736.

7.3.4 The application ranges of the three different technologies differ due to the pros and cons. For very high energies (>>MeV) where a very high collimation is required for scatter prevention, many high energy CT systems are equipped with LDA detectors. Due to the higher scintillator efficiency and physical reduction of scatter by the fan beam and collimation slit, LDA CT systems usually offer a better image quality compared to CT systems with DDAs; LDAs are commonly used in the energy range from 0.4 to 20 MeV. When possible, DDAs are preferred because they deliver 3D scans within a much shorter time. It is also becoming increasingly common to build a system with both an LDA and DDA to allow for either option.

7.4 *Mechanical Scanning Equipment*—The mechanical equipment provides the relative motion between the test object, the source, and the detectors. It makes no difference, at least in principle, whether the test object is moved systematically relative to the source and detectors, or if the source and detectors are moved relative to the test object. Physical considerations such as the weight or size of the test object should be the determining factors for the most appropriate motion to use.

7.5 *Computer Systems*—The computer system performs the tasks of Operator Interface, acquisition, reconstruction, visualization, and storage.

7.5.1 The Operator Interface is the primary control of the system and the test examination, including controls for acquisition, reconstruction, visualization, and storage.

7.5.1.1 Acquisition refers to the control of the mechanical handling system and electronic controls for the data collection for the specific examination. This includes controlling the scan motion, source operation, and data acquisition functions.

7.5.1.2 Reconstruction refers to the parameters and mathematical operations for creation of the resultant slice or volume.

7.5.1.3 Visualization includes the image display and processing of the reconstructed data. Image display and processing are subfunctions of the computer system that provide a degree of image interaction not available with conventional radiography. The mapping between the pixel linear attenuation coefficient and the displayed intensity of the pixel can be changed to accommodate the best viewing conditions for a particular feature. Image processing functions such as statistical and densitometric analyses can be performed on an image or group of images. The digital nature of the image allows major advances in the way data are processed, analyzed, and stored.

7.5.1.4 *Storage*—Storage includes the archiving requirements for the CT examination. Information such as image data, operating parameters, part identification, operator comments, slice orientation, and other data is usually archived in a computer-readable, digital format on some type of storage medium. An advantage of saving this material in computer-readable format (rather than in simple hardcopy) is that old and new data sets can be compared directly, and subsequent changes in reconstruction or analysis procedures can be reapplied to saved data or images.

8. General Principles of CT/Main CT Process Steps

8.1 *Mathematical Basis of CT*—CT is the science of recovering an estimate of the internal structure of an object from a systematic indirect measurement of a physical property, such as the linear attenuation coefficient by means of X-ray projection images. This is performed by measuring a complete set of line integrals involving the physical parameter of interest over the designated cross-section and then using an algorithm to recover an estimate of the spatial variation of the parameter over the desired slice.

8.1.1 A set of X-ray attenuation measurements is made along a set of paths projected at different locations around the periphery of the test object. The first part of Fig. 4 illustrates a set of measurements made on a test object containing two attenuating disks of different diameters. The X-ray attenuation measurement made at a particular angle, φ_1, is referred to as a single projection. It is shown as $f_{\varphi 1}(x')$, where x' denotes the linear position of the measurement. The second part of Fig. 4 shows measurements taken at several other angles $f_{\varphi i}(x')$. Each of the attenuation measurements within these projections is digitized and stored in a computer, where it is subsequently conditioned (for example, normalized and corrected) and filtered (convolved). The next step in image processing is to backproject the views, which is also shown in the second part of Fig. 4. Backprojection consists of projecting each view back along a line corresponding to the direction in which the projection data were collected. The backprojections, when enough views are employed, form a faithful reconstruction of the object. Even in this simple example, with only four projections, the concentration of backprojected rays already begins to show the relative size and position of features in the original object.

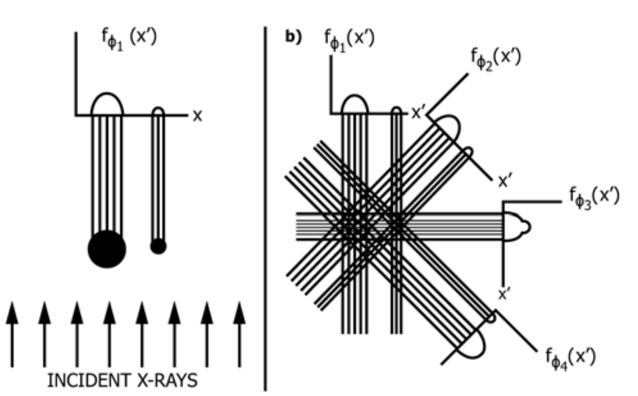

FIG. 4 Schematic Illustrations of How CT Works

8.2 *Radon Transform*—The theoretical mathematical foundation underlying CT was established in 1917 by J. Radon (**1**).[4] Radon established that if the infinite set of line integrals of a function, which is finite over some region of interest and zero outside it, is known for (parallel) ray paths through the region along all angles, then the value of the function over that region can be uniquely determined. A particular function and its associated set of line integrals form a transform pair; the set of integrals is referred to as the Radon transform of the function. Radon demonstrated the existence of an inverse transform for recovering a function from its Radon transform, providing an important existence theorem for what later came to be called CT.

8.2.1 The essential technological requirement is that a set of systematically sampled line integrals of the parameter of interest are measured over the cross-section of the object under examination and that the geometrical relationship of these measurements to one another be well known. Within this constraint, many different methods of collecting useful data exist. However, the quality of the resulting reconstruction depends on at least three major factors: (1) how finely the object is sampled, (2) how accurately the individual measurements are made, and (3) how precisely each measurement can be related to an absolute frame of reference.

8.2.2 *Sampling the Radon Transform*—For monoenergetic X-rays, attenuation in matter is governed by Lambert's law of absorption (**2**), which holds that each layer of equal thickness attenuates an equal fraction of the radiation that traverses it. Mathematically, this can be expressed as the following:

$$\frac{dI}{I} = -\mu dx \tag{1}$$

where:

I = the intensity of the incident radiation,
dI/I = the fraction of radiation removed from the flux as it traverses a small thickness, dx, of material, and
μ = the constant of proportionality.

In the physics of X-ray attenuation, μ is referred to as the linear attenuation coefficient. Eq 1 can be integrated easily to describe X-ray attenuation in the following perhaps more familiar form:

$$I = I_o e^{-\mu x} \tag{2}$$

where:

I_o = the intensity of the unattenuated radiation, and
I = the intensity of the transmitted flux after it has traversed a layer of material of thickness x.

[4] The boldface numbers in parentheses refer to a list of references at the end of this standard.

8.2.2.1 If X-rays penetrate a non-homogeneous material, Eq 2 must be rewritten in the more general form:

$$I = I_o e^{-\int \mu(s) ds} \quad (3)$$

where the line integral is taken along the direction of propagation and $\mu(s)$ is the linear absorption coefficient at each point on the ray path. In CT, the fractional transmitted intensity, I/I_o , is measured for a very large number of ray paths through the object being examined and then the logarithm is applied to obtain a set of line integrals for input to the reconstruction algorithms. Specifically, the primary measurements, I and I_o, are processed to obtain the necessary line integrals:

$$\int \mu(s) ds = -ln\left(\frac{I}{I_o}\right) \quad (4)$$

8.2.3 For the reconstruction problem the set of line integrals must represent a systematic sampling of the entire object. If the circle of reconstruction is inscribed in an M by M image matrix, this implies $(\pi/4)\ M^2$ unknowns and a need for at least $(\pi/4)\ M^2$ linearly independent measurements. Since the presence of random noise corrupts the data, and due to strong correlation between consecutive X-ray images with small angular increment, the minimum sampling requirement is greater than for noise-free data as well as to be sensitive to the algorithm employed. Typically, data set sizes are on the order of one to three times the minimal amount, depending on the system and the application. Arbitrarily complex objects require more data than objects with simple geometrical shapes or highly developed symmetries. The number of samples per view is generally more important than the number of views, and the relative proportion of views and samples should reflect this principle. In addition, each line integral must be accurately known, as well as referenced accurately to a known coordinate system. This places strict requirements on the data acquisition and mechanical handling systems.

8.2.3.1 *Sampling*—Fan-Beam reconstruction algorithms, in general require at least 180° of rotational data plus fan angle be collected by the scanner; however, there is the potential for artifacts in the data. Cone beam reconstructions require 180° plus fan angle but commonly acquire 360° of rotational data, and an uneven number of projections; the exact number of projections needed depends on the shape of the object and the reconstruction algorithm. There are reconstruction algorithms available that require only a limited set of views; however, there is greater uncertainty in the resultant slice volume and artifacts may be present in the data. Therefore, as with any technique, the user must learn to recognize and be able to discount common artifacts subjectively.

(1) For Feldkamp based reconstruction algorithms, the number of independent views (projections) should be $V \geq \frac{\pi}{2} \cdot M$ with an uneven number of projections for 360° to avoid star artifacts (see 9.7.7). For practical applications the number of views is typically reduced to $V = \frac{\pi}{4} \cdot M$, since most algorithms of filtered back projection have a smoothing effect at higher spatial frequencies

8.3 *Reconstruction: Inverting the Radon Transform*—The reconstruction task can be defined as follows: given a set of systematic transmission measurements corrupted by various known and unknown sources of error, determine the best estimate of the cross-section of the object associated with that data. There are three broad classes of reconstruction algorithms: (1) matrix inversion methods, (2) finite series-expansion methods, and (3) transform methods. Today, matrix inversion and finite series-expansion are not in typical usage; transform methods are conventionally used today in CT.

8.3.1 Transform methods are based on analytical inversion formulas. The two primary types of transform methods are (1) the convolution-backprojection algorithm or filtered backprojection algorithm and (2) the direct Fourier algorithm.

8.3.2 *Backprojection Methods*—Backprojection can be thought of as reversing the data collection process. Each sample within a given projection represents the fractional transmittance of a narrow beam of X-rays through the object, which is assumed to be sufficiently well approximated by small, discrete voxels of constant attenuation. Conceptually, backprojection can be thought of as smearing each profile back across the image in the direction of the radiation propagation. Filtered backprojection is used to reduce the "smearing" effect of each profile; a filter is applied (convolution with the data profiles) that preserves the essential response of the detector to the presence of the point object but adds a negative tail to reduce the falloff that occurs with pure back-projection. See Fig. 5. Backprojection method is the method used by virtually

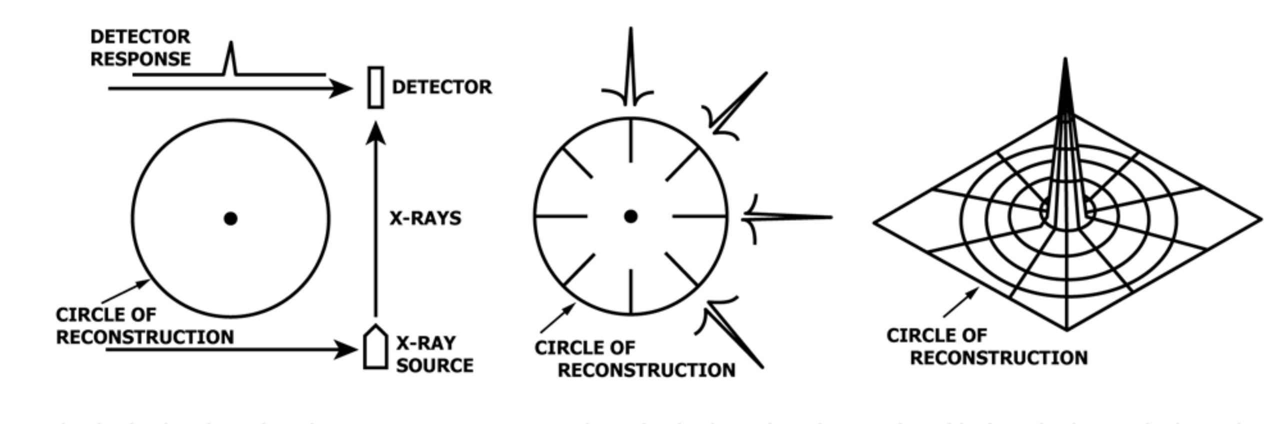

FIG. 5 Filtered Convolution Backprojection

all commercial CT systems.

8.3.3 *Direct Fourier Algorithm*—The direct Fourier algorithm is based on the underlying fact that the one-dimensional Fourier transform of a CT projection of an object corresponds to a spoke in Fourier space of the two-dimensional transform of that object (the so-called Central-Section Theorem or Projection-Slice Theorem (**3**)). Thus, in theory, all that is required in order to obtain an image by this method is to transform each projection as it is collected; place it along its proper spoke in two-dimensional Fourier space; and when all the views have been processed, take the inverse two-dimensional Fourier transform to obtain the final image. This only works for parallel beam data.

8.3.4 *Feldkamp Reconstruction Algorithm*—The Feldkamp algorithm is applied to cone-beam generated datasets. The key to the Feldkamp algorithm is that all slices in the cone beam geometry except the center one violates Tuy's (**4**) sufficiency condition that any plane through any voxel must intersect the source path. What Feldkamp observed was that if you do backprojection along a cone with shallow enough angle the reconstruction works with minimal artifacts in spite of this. This angle is usually stated as a 5° or 6° half angle (10° or 12° full angle) but slight artifacts will still be seen on some parts at the top and bottom (**5-7**). The main advantages of this reconstruction algorithm are the possibility of using partial detection coverage and high computational efficiency.

8.4 *Reconstruction Matrix Size*—The reconstruction matrix size governs the number of views and data samples in each view that must be acquired. The higher the resolution, the smaller the pixel size and the larger the pixel matrix for a given region of interest on the test object. The reconstruction matrix size affects the number of scans and length of time necessary to examine an object.

9. CT System Performance Overview

9.1 *CT System Performance*—As a CT system can never exactly duplicate the object that is scanned, the ability of a CT system to image thin cross-sectional areas of interest through an object is dictated largely by the competing influences of the spatial resolution, the statistical noise, and the artifacts of the imaging system. This section will define and derive the issues associated with a CT system performance, including Modulation Transfer Function (MTF), Contrast Discrimination Function (CDF), and Contrast-Detail-Diagram (CDD). The CDD is generated from the CDF, which represents ability to discriminate a contrasting feature from the base, and the MTF which represents the factor which reduces contrast in a CT. For the specific procedure associated with MTF and CDF calculation, see Test Method E1695.

9.2 *Modulation Transfer Function (MTF)*—The MTF describes by which factor the contrast of a periodical pattern is reduced by the total unsharpness of the CT system. The MTF describes the transfer of a modulation in an image signal (relative intensity variation) by a CT system as a function of the modulation's spatial frequency. Intentionally, it does not include noise effects as those strongly depend on scan parameters and sample materials. Noise effects are covered by the CDF.

9.2.1 Mathematically, the MTF is the modulus of the one-dimensional Fourier Transform (Magnitude) of the Line Spread Function (LSF). The LSF may be described as the one-dimensional profile across the image of a line. This profile is not accessible to a direct measurement since there is no real physical implementation of such (one-dimensional) line. However, the LSF is the first derivative of the one-dimensional profile across the image of a sharp edge, the Edge Response Function (ERF). These three steps are illustrated in Fig. 6.

9.3 *Contrast Discrimination Function (CDF)*—The ability to discriminate a contrasting feature of size *D* in mm or size *D** in voxels from the base, at a certain noise level, can be described, approximately, by a single curve, called the CDF. The CDF describes the influence of image noise on the detectability (contrast sensitivity) of a feature in an elsewhere homogeneous material neighborhood as a function of the size D* of this feature in voxels. If it is multiplied by a physiological factor c it represents the ability of human observers to recognize indications with larger than size *D* at a smaller Contrast Noise Ratio (CNR), if the unsharpness can be neglected.

9.3.1 *Calculation of CDF*—To calculate the CDF, it is necessary to determine the noise in the image. The image noise at the center of a uniform cylinder of material is characterized by measuring the standard deviation in the mean σ_m, for different areas of interest. The process for determining σ_m begins by selecting a Region of Interest (ROI). Each ROI is made of a series of tiles of squares of voxels from *1 x 1*, *2 x 2*, up to *k x k* voxels. Each tile within the ROI has the mean value of the voxel values within it. See Test Method E1695. First, a tomographic slice image of a homogenous object (cylinder) is densely covered with *m* square tiles T_i of side length *D**. Within each tile, within the ROI, the mean voxel value μ_i of all $n = D^{*2}$ voxel values $x \in T_i$ is defined by:

$$\mu_i(D^*) = \frac{1}{n} \sum_{x \in T_i} x \tag{5}$$

where:

D = length in mm;
Δv = voxel to voxel distance; and
D^* = length in number of voxels $D^* = \frac{D}{\Delta v}$

9.3.2 The ROI is then moved to an adjacent non-overlapping location and the measurement is repeated. This procedure is continued until enough independent data has been acquired to generate an accurate ensemble distribution, or histogram, of the sampling process. Once the histogram has been obtained, the standard deviation of the distribution is calculated.

$$\sigma_m(\mu_i) = \sqrt{\frac{1}{m-1} \sum_i (\mu_i - \bar{\mu})^2} \tag{6}$$

where $\bar{\mu}$ is the mean (signal or contrast to air) of all tile means μ_i. This procedure is repeated for many tile sizes or values of *D**, for example, from 1…100 voxels.

9.3.3 The resulting standard deviation of the mean is the noise in the image in a particular ROI because it represents the uncertainty associated with a measurement of the average CT value over a particular area. Generally, the whole process is

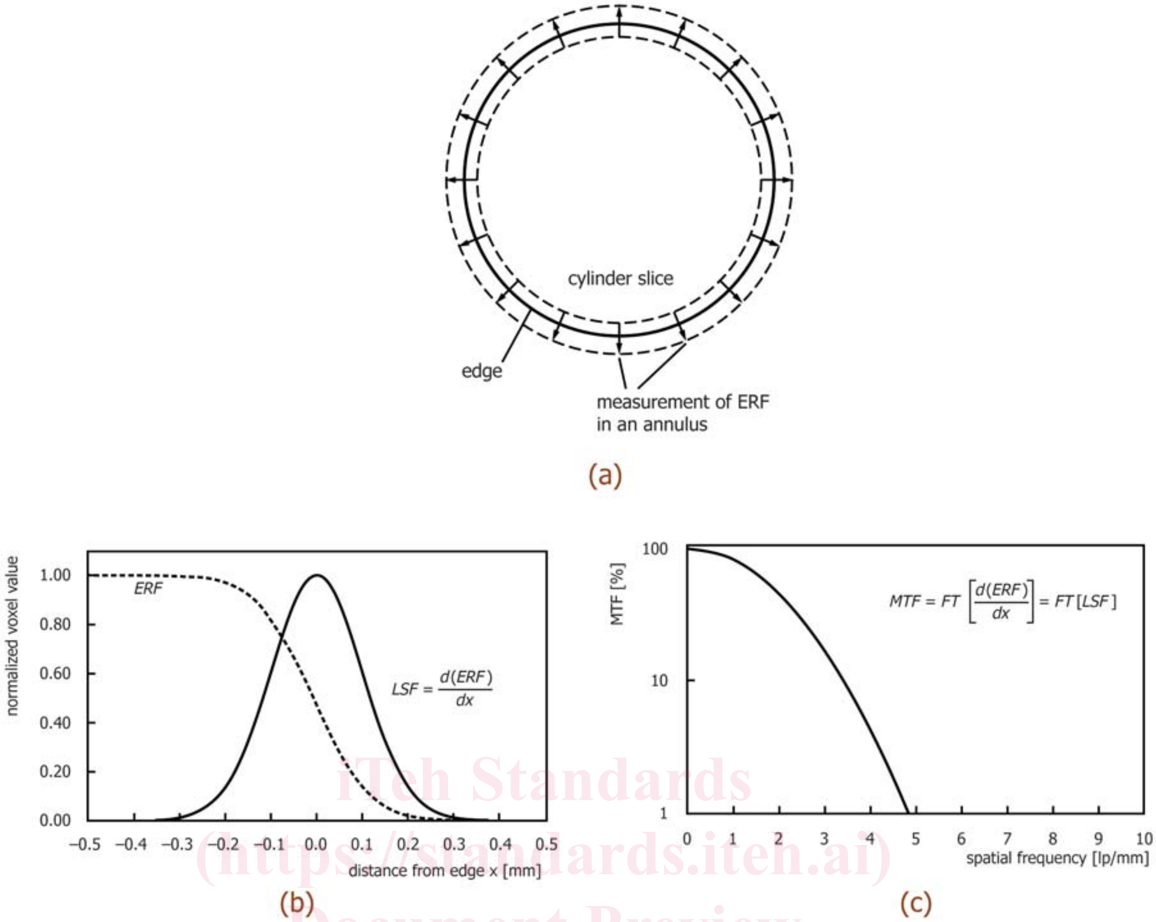

FIG. 6 (a) The ERF is Measured in the Vicinity of the Reconstructed Slice's Edge; (b) the ERF is Differentiated to Obtain the LSF; (c) the Fourier Transform (Magnitude Spectrum) of the LSF Results in the MTF

systematically implemented for different sized ROI, from an area of only one pixel to an area of 100 pixels, or more. Note that for the special case where the specified area is only a single pixel, the standard error in the mean equals the more familiar standard deviation used to compute the CNR or SNR, respectively. The subscript m is used to distinguish the error in the mean from the standard deviation, since the same symbol σ is used for both.

9.3.4 Experience has shown that the CDF conditions will be satisfied when the radius of the central region is about one-third that of the disk, in that other influences affecting the statistical nature of the noise will be negligible. For each new application, it is recommended that the validity of these requirements be verified empirically by monitoring the behavior of the CDF as the size of the region of interest is increased. The visibility of round indications for human observers requires the correction of the CDF by the MTF and a physiological factor c resulting in the CDD, if the image unsharpness cannot be neglected as described in 9.2. For the specific test associated with CDF system calculation, see Test Method E1695 which utilizes the CT image of a cylinder. CDF (D*) is calculated by:

$$CDF(D^*) \;:=\; \frac{\sigma_m(\mu_i)}{\bar{\mu}} 100\ \% \tag{7}$$

Fig. 7 shows a measured CDF curve for the central slice of a CT cylinder of 1.5 in. diameter. Thus, technically the CDF is a plot of the relative standard deviation of the mean $\sigma_m(\mu_i)$ plotted versus the feature size D*.

9.3.5 Provided the unsharpness can be neglected, as a rule, the minimum true relative contrast $\Delta\mu_f(\%)_D$* of a feature of size D* to be detected in the tomographic slice by a human observer should be:

$$\Delta\mu_f \geq c \cdot CDF(D^*) \tag{8}$$

where c is a physiological-based constant in the range of 2-5, determined by investigators as the factor for detection based on the observations of human beings (**8, 9**), see Annex A1. With the relative feature contrast $\Delta\mu_f$

$$\Delta\mu_f(\%) \;:=\; \frac{|\mu_f - \mu_b|}{\mu_b} \cdot 100\ \ \% \tag{9}$$

where, μ_f and μ_b are the mean voxel values of feature and background material voxels, respectively for $\mu_b > 0$.

9.4 *Contrast-Detail-Diagram (CDD)*—In practice, detection is not based solely on threshold criteria. Human beings use visual integration when detecting features. Thus, detection criteria should be based on the observations of human beings. Several investigators (**8, 9, 10**) have reported that the effective contrast, which human beings can detect with a 50 % probability of success, depends on the image noise and the feature diameter. The plot of the contrast required for 50 % discrimination, that is probable discrimination, of pairs of