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Standard Guide for Digital Detector Array Radiology[1]

This standard is issued under the fixed designation E2736; 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.

1. Scope

1.1 This standard is a user guide, which is intended to serve as a tutorial for selection and use of various digital detector array systems nominally composed of the detector array and an imaging system to perform digital radiography. This guide also serves as an in-detail reference for the following standards: Practices E2597, E2698, and E2737.

1.2 *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 and health practices and determine the applicability of regulatory limitations prior to use.*

2. Referenced Documents

2.1 *ASTM Standards:*[2]

E94 Guide for Radiographic Examination
E155 Reference Radiographs for Inspection of Aluminum and Magnesium Castings
E192 Reference Radiographs of Investment Steel Castings for Aerospace Applications
E747 Practice for Design, Manufacture and Material Grouping Classification of Wire Image Quality Indicators (IQI) Used for Radiology
E1000 Guide for Radioscopy
E1025 Practice for Design, Manufacture, and Material Grouping Classification of Hole-Type Image Quality Indicators (IQI) Used for Radiology
E1316 Terminology for Nondestructive Examinations
E1320 Reference Radiographs for Titanium Castings
E1742 Practice for Radiographic Examination
E1815 Test Method for Classification of Film Systems for Industrial Radiography
E1817 Practice for Controlling Quality of Radiological Examination by Using Representative Quality Indicators (RQIs)
E2002 Practice for Determining Total Image Unsharpness in Radiology
E2422 Digital Reference Images for Inspection of Aluminum Castings
E2445 Practice for Performance Evaluation and Long-Term Stability of Computed Radiography Systems
E2446 Practice for Classification of Computed Radiology Systems
E2597 Practice for Manufacturing Characterization of Digital Detector Arrays
E2660 Digital Reference Images for Investment Steel Castings for Aerospace Applications
E2669 Digital Reference Images for Titanium Castings
E2698 Practice for Radiological Examination Using Digital Detector Arrays
E2737 Practice for Digital Detector Array Performance Evaluation and Long-Term Stability

3. Terminology

3.1 *Definitions of Terms Specific to This Standard:*

3.1.1 *digital detector array (DDA) system*—an electronic device that converts ionizing or penetrating radiation into a discrete array of analog signals which are subsequently digitized and transferred to a computer for display as a digital image corresponding to the radiation energy pattern imparted upon the input region of the device. The conversion of the ionizing or penetrating radiation into an electronic signal may transpire by first converting the ionizing or penetrating radiation into visible light through the use of a scintillating material. These devices can range in speed from many minutes per image to many images per second, up to and in excess of real-time radioscopy rates (usually 30 frames per seconds).

3.1.2 *signal-to-noise ratio (SNR)*—quotient of mean value of the intensity (signal) and standard deviation of the intensity (noise). The SNR depends on the radiation dose and the DDA system properties.

3.1.3 *normalized signal-to-noise ratio (SNR_n)*—SNR normalized for basic spatial resolution (see Practice E2445).

3.1.4 *basic spatial resolution (SRb)*—basic spatial resolution indicates the smallest geometrical detail, which can be resolved using the DDA. It is similar to the effective pixel size.

[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.

3.1.5 *efficiency*—$dSNR_n$ (see 3.1.6 of Practice E2597) divided by the square root of the dose (in mGy) and is used to measure the response of the detector at different beam energies and qualities.

3.1.6 *achievable contrast sensitivity (CSa)*—optimum contrast sensitivity (see Terminology E1316 for a definition of contrast sensitivity) obtainable using a standard phantom with an X-ray technique that has little contribution from scatter.

3.1.7 *specific material thickness range (SMTR)*—material thickness range within which a given image quality is achieved.

3.1.8 *contrast-to-noise ratio (CNR)*—quotient of the difference of the mean signal levels between two image areas and the standard deviation of the signal levels. The CNR depends on the radiation dose and the DDA system properties.

3.1.9 *lag*—residual signal in the DDA that occurs shortly after the exposure is completed.

3.1.10 *burn-in*—change in gain of the scintillator or photoconductor that persists well beyond the exposure.

3.1.11 *internal scatter radiation (ISR)*—scattered radiation within the detector (from scintillator, photodiodes, electronics, shielding, or other detector hardware).

3.1.12 *bad pixel*—a bad pixel is a pixel identified with a performance outside of the specification for a pixel of a DDA as defined in Practice E2597.

3.1.13 *grooved wedge*—a wedge with one groove, that is 5 % of the base material thickness and that is used for achievable contrast sensitivity measurement in Practice E2597.

3.1.14 *phantom*—a part or item being used to quantify DDA characterization metrics.

4. Significance and Use

4.1 This standard provides a guide for the other DDA standards (see Practices E2597, E2698, and E2737). It is not intended for use with computed radiography apparatus. Figure 1 describes how this standard is interrelated with the aforementioned standards.

4.2 This guide is intended to assist the user to understand the definitions and corresponding performance parameters used in related standards as stated in 4.1 in order to make an informed decision on how a given DDA can be used in the target application.

4.3 This guide is also intended to assist cognizant engineering officers, prime manufacturers, and the general service and manufacturing customer base that may rely on DDAs to provide advanced radiological results so that these parties may set their own acceptance criteria for use of these DDAs by suppliers and shops to verify that their parts and structures are of sound integrity to enter into service.

4.4 The manufacturer characterization standard for DDA (see Practice E2597) serves as a starting point for the end user to select a DDA for the specific application at hand. DDA manufacturers and system integrators will provide DDA performance data using standardized geometry, X-ray beam spectra, and phantoms as prescribed in Practice E2597. The end user will look at these performance results and compare DDA metrics from various manufacturers and will decide on a DDA that can meet the specification required for inspection by the end user. See Sections 5 and 8 for a discussion on the characterization tests and guidelines for selection of DDAs for specific applications.

4.5 Practice E2698 is designed to assist the end user to set up the DDA with minimum requirements for radiological examinations. This standard will also help the user to get the required SNR, to set up the required magnification, and provides guidance for viewing and storage of radiographs. Discussion is also added to help the user with marking and identification of parts during radiological examinations.

4.6 Practice E2737 is designed to help the end user with a set of tests so that the stability of the performance of the DDA can be confirmed. Additional guidance is provided in this document to support this standard.

4.7 Figure 1 provides a summary of the interconnectivity of these four DDA standards.

5. DDA Technology Description

5.1 *General Discussion:*

5.1.1 DDAs are seeing increased use in industries to enhance productivity and quality of nondestructive testing. DDAs are being used for in-service nondestructive testing, as a diagnostic tool in the manufacturing process, and for inline testing on production lines. DDAs are also being used as hand held, or scanned devices for pipeline inspections, in industrial computed tomography systems, and as part of large robotic scanning systems for imaging of large or complex structures. Because of the digital nature of the data, a variety of new applications and techniques have emerged recently, enabling quantitative inspection and automatic defect recognition.

5.1.2 DDAs can be used to detect various forms of electromagnetic radiation, or particles, including gamma rays, X-rays, neutrons, or other forms of penetrating radiation. This standard focuses on X-rays and gamma rays.

5.2 *DDA architecture:*

5.2.1 A common aspect of the different forms of this technology is the use of discrete sensors (position-sensitive) where, the data from each discrete location is read out into a file structure to form pixels of a digital image file. In all its simplicity, the device has an X-ray capture material as its primary means for detecting X-rays, which is then coupled to a solid-state pixelized structure, where such a structure is similar to the imaging chips used in visible-wavelength digital photography and videography devices. Figure 2 shows a block diagram of a typical digital X-ray imaging system.

5.2.2 An important difference between X-ray imaging and visible-light imaging is the size of the read-out device. The imagers found in cameras and for visible-light are typically on the order of 1 to 2 cm^2 in area. Since X-rays are not easily focused, as is the case for visible light, the imaging medium must be the size of the object. Hence, the challenge lies in meeting the requirement of a large uniform imaging area without loss of spatial information. This in turn requires high pixel densities of the read-out device over the object under

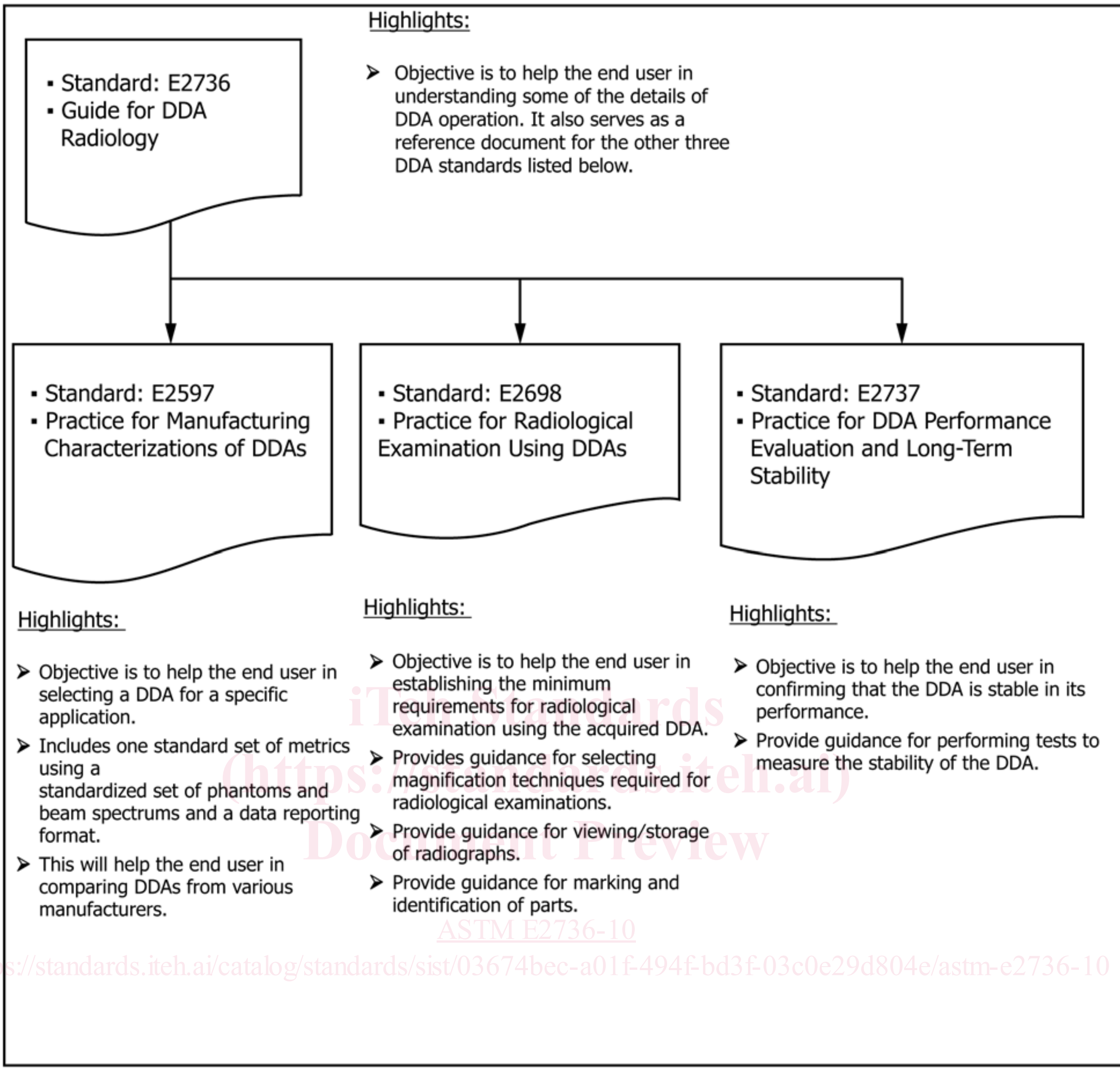

FIG. 1 Flow Diagram Representing the Connection Between the Four DDA Standards

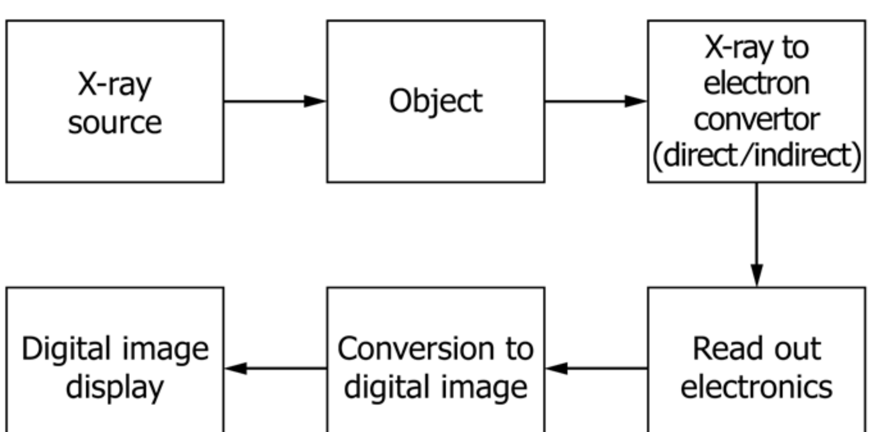

FIG. 2 Block Diagram of a Typical Digital X-Ray Imaging System

examination, as well a primary sensing medium that also retains the radiologic pattern in its structure. Therefore, each DDA consists of a primary X-ray or gamma ray capture medium followed by a pixelized read structure, with various means of transferring the above said captured pattern. For each of these elements, there are numerous options that can be selected in the creation of DDAs. For the primary X-ray conversion material, there are either luminescent materials such as scintillators or phosphors, and photoconductive materials also known as direct converter semiconductors.

5.2.3 For read-out structures, the technology consists of charge coupled detectors (CCDs), complementary metal oxide silicon (CMOS) based detectors, amorphous silicon thin film transistor diode read-out structures, and linear or area crystalline silicon pixel diode structures. Other materials and structures are also possible, but in the end, a pixelized pattern is captured and transferred to a computer for review.

5.2.4 Each primary conversion material can be coupled with the various read structures mentioned through a wide range of coupling media, devices, or circuitry. With all of these possible combinations, there are many different types of DDAs that have been produced. But all result in a digital X-ray or gamma ray image that can be used for different NDT applications.

5.2.5 Following the capture of the X-rays and conversion into an analog signal on the read-out device, this signal is typically amplified and digitized. There are numerous schemes for each of these steps, and the reader is referred to (**1**, **2**, **3**)[3] for further discussion on this topic.

5.3 *Digitization Methods:*

5.3.1 Digitization techniques typically convert the analog signal to discrete pixel values. For DDAs the digitization is typically, 8-bit (256 gray values), 12-bits (4096 gray values), 14-bits (16,384 gray levels) or 16-bits (65,536 values). The higher the bit depth, the more finely the signal is sampled.

5.3.2 The digitization does not necessarily define the gray level range of the DDA. The useful range of performance is defined by the ability of the read device to capture signal in a linear relation to the signal generated by the primary conversion device. A wide linear range warrants the use of a high bit depth digitizer. It should be noted that if digitization is not high enough to cover the information content from the read device, digitization noise might result. This can be manifested as a posterization effect, where discrete bands of gray levels are observed in the image.

5.3.3 Conversely, if digitization is selected that is significantly higher than the range of the read-out device then the added sampling may not necessarily improve performance. Secondly, if the digitization is completed well beyond the linear range of the read structure, these added gray levels would not be useable. For example, 16-bits of digitization do not necessarily indicate 65,536 levels of linear responsivity.

5.3.4 The useful range of a detector is frequently defined as the maximum usable level, without saturation in relation to the noise floor of the DDA, where again no useful differentiation can be extracted from the data. This is sometimes referred to as the detector dynamic range.

5.3.5 The dynamic range is different from the specific material thickness range (SMTR) as defined in this standard and Practice E2597. That range is a true practical range of the DDA at hand, a range significantly tighter than the DDA dynamic range.

5.3.6 The SMTR is one of the properties to consider in DDA selection, as it impacts the thickness range that can be interpreted in a single view. This is dependent on the characteristics of the read device and the digitization level. This test provides a means of determining an effective range without understanding the subtle nuances of the detector readout, and avoids erroneous parallels between bit depth and its relation to thickness range, and maximum possible signal from a device.

5.4 *Specific DDA components*—There are numerous options in each component of the imaging chain to produce a DDA. To understand the options and limitations of each category, and to best assess which technology to pursue for a given application, the underlying technology will be discussed beginning with the image capture medium. This is followed by the image read structure and then the image transfer device is discussed for the various configurations of the read-out devices. For a more detailed description of the architectures of these devices, the reader is referred to Ref. (**2**).

5.4.1 *X-ray Capture—Scintillators (phosphors)*—Scintillators are materials that convert X-ray or gamma ray photons into visible-light photons, which are then converted to a digital signal using technologies such as amorphous silicon (a-Si) arrays, CCDs or CMOS devices together with an analog-to-digital converter. This will facilitate real time acquisition of images without the need for offline processing. Since there are various stages of conversion involved in recording the digital image, it is very important to ensure that minimum information is lost during conversion in the scintillator. The properties desirable of ideal scintillators are listed below. These properties allow for high efficiency, stable and robust operation yielding ideal imaging performance:

(1) High stopping power for X-rays obtained by high atomic number and, or the use of high density materials without loss of spatial information due to scattering processes within the scintillator.

(2) High X-ray to light conversion efficiency

(3) Matched emission spectrum of the scintillator to the spectral sensitivity of the light collection device

(4) Low afterglow during and after termination of the X-ray illumination.

(5) Stable output during long or intense exposure to radiation.

(6) Temperature independence of light output.

(7) Stable mechanical and chemical properties.

5.4.1.1 The scintillator based on CsI:Tl (thallium doped cesium iodide) has shown considerable success as a scintillator because of the following reasons:

(1) Cesium iodide can be formed into needles (see Fig. 3) and coupled directly to a diode read structure or a fiber optic component to direct the light to the photodiodes without significant light loss or optical scatter. This is the most efficient

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

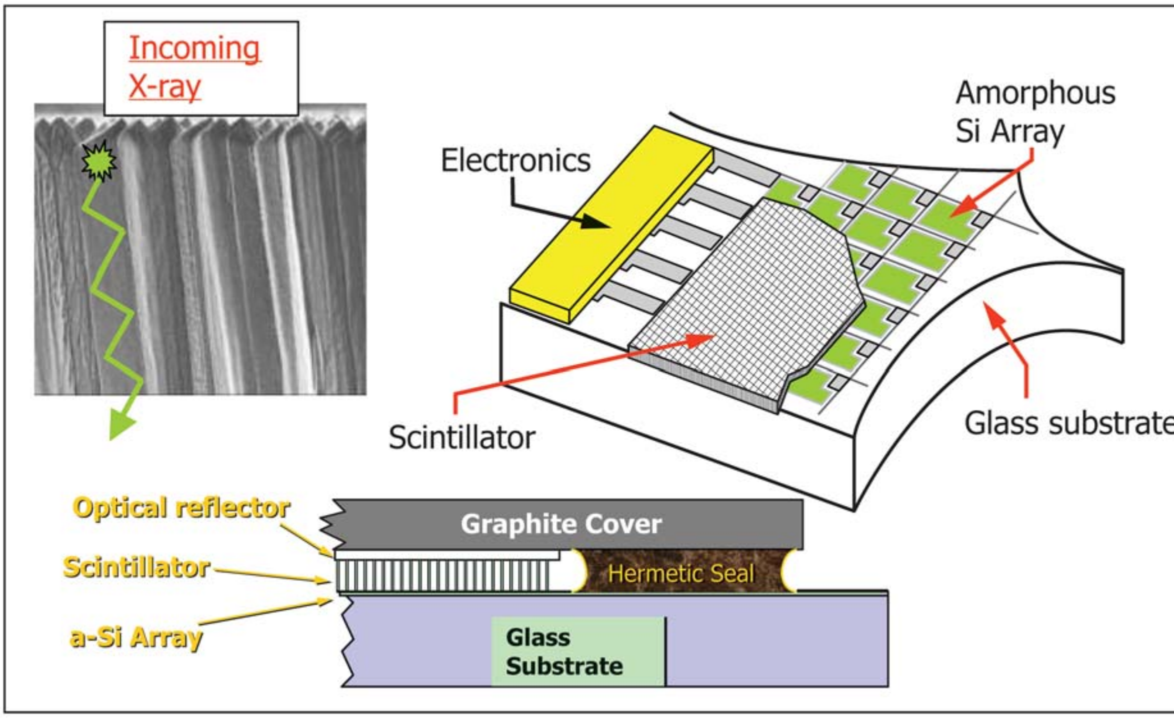

FIG. 3 Architecture of CsI:Tl needle structure demonstrating light guiding nature following x-ray conversion to light, and the amorphous silicon architecture illustrating direct contact of the scintillator with the diode thin film transistor readout matrix.

means for depositing light into photodiodes. All other scintillators lose light at the interface because of reduced optical coupling between the scintillator and the diode structure. CsI is very well matched in index of refraction to that of the entry layer of the amorphous silicon diode structure.The needle-like structure enables thick phosphor layers, which improves X-ray absorption without significant loss in spatial resolution.

(2) The cesium iodide has a high effective atomic number (Z) which also contributes to good X-ray absorption efficiency.

The drawback of CsI are:

(1) CsI:Tl has been prone to severe hysteresis effects, an effect that leads to an unstable signal under constant flow of X-rays, and this instability is non-linear with the dose rate used. This can cause residual images (ghost images) to be retained in the detector from prior scans. In some circumstances, recent preparations have significantly overcome this effect.

(2) CsI is hygroscopic and sensitive to moisture, and must be encapsulated to avoid loss in crystallinity.

(3) CsI:Tl has a primary decay time of 1 microsecond at 1/e (to ~37 % of peak signal), but has a long decay component into the millisecond range that is non-zero but well below 1 %. This needs to be taken into consideration where a very opaque object follows either an open air exposure or a very low opacity exposure, as the afterglow from the bright exposure may encroach on the signal level of the dim exposure.

5.4.1.2 Other scintillators (phosphors) such as polycrystalline Gd_2O_2S:Tb have been successfully used, but have limitations on how thick they can be made given that the powder architecture scatters the light produced from the deposited X-rays. Nevertheless, these are simple phosphors to purchase and implement, and like the CsI needles, can be optically coupled through a lens, or directly coupled to a read structure. For the latter, and as with CsI:Tl, this can be achieved via a fiber optic lens, an optical lens, or by direct coupling to the read-structure itself.

5.4.1.3 Certain scintillators such as Gd2O2S:Tb can be sintered to ceramic imaging plates with discrete cell boundaries yielding the same advantage of the CsI needle structures, but typically without the temporal drawbacks of the CsI:Tl chemistry. However, they are difficult to grow directly onto diode structures, typically require an optical couplant to improve transfer efficiency due to index mismatch, and typically are more expensive to produce and couple to large diode structures.

5.4.1.4 Certain glass scintillators based on terbium activation can be formed into fiber optic scintillating plates yielding the same advantage of the CsI needle structures. These plates tend to also have some temporal drawbacks, and are not as efficient in converting X-rays to light as any of the other scintillators already mentioned.

5.4.1.5 Other materials are under development, and the above sections are not intended to cover all possible options.

5.4.1.6 *Temporal Properties of scintillators*—When radiation impinges upon a scintillator, the atoms/molecules in the scintillator material absorb this radiation and get excited. They de-excite by emitting the energy in the form of visible light. The emitted energy is 'luminescence,' which falls broadly under two categories namely, fluorescence and phosphorescence. These manifest as a two-component exponential decay—fast (prompt) for fluorescence and slow (delayed) for phosphorescence. An ideal scintillator should essentially have only a fast decay component with a linear conversion, that is, light yield should be proportional to the deposited energy. Any phosphorescence might introduce residual latent artifacts into subsequent imagery and make interpretation difficult. Scintillator phosphorescence can lead to image lag or image burn-in as defined herein, where features from prior images contaminate new scenes.

5.4.2 *Semiconductors (Photoconductors)*—A photoconductive material converts X-rays to electron-hole pairs that then get separated by the internal bias of the device as defined by the material properties, such as the manufactured charge imbalance into the semiconductor material. As with scintillating materials, another electronic element is needed to capture the signal produced, such as an electrode structure with pixelization, possibly with additional added electron bias on one electrode to separate the electron-hole pairs. But unlike a scintillating material, there is a lower likelihood that the charges produced will have as much lateral spread as experienced optically in luminescent materials. Also since the photoconductive material converts the X-ray signal directly into electron-hole pairs, there is greater conversion efficiency than with the production of light, that first generates electron-hole pairs prior to producing the light. For X-ray applications, photoconductive materials such as amorphous selenium (a-Se), CdTe, and HgI2 have been used because of their high atomic numbers, and the ability to manufacture these materials into a monolithic structure. Other photoconductive materials are available, or may become available in the future. It should be noted that although light is not generated from these materials, lag and burn-in effects can occur due to subtle effects of sweeping the charge out of the semiconductor.

5.5 *Capture of the converted image:*

5.5.1 *Charge-coupled devices (CCDs)* are light imaging devices that are typically small in size, and have high pixel densities. They use a transparent poly-silicon gate structure for reading out the device, and because of their high pixel fill factor are very efficient in collecting the light produced from the phosphor material. Unlike amorphous silicon pixel structures, current limitations in crystal growth methods have restricted the fabrication of these devices into larger arrays. A larger field of view can be accomplished with CCDs through a lens or a fiber optic transfer device to view a phosphor or scintillator screen. The downside of the lens approach is that it has very poor light collection efficiency, while fiber optic image plates have significantly improved light collection efficiency, but are expensive and are not amenable to large fields of view. For small field of view applications, the directly coupled charge coupled device approach will provide high spatial resolution and high light collection efficiency.

5.5.2 *CMOS read structures* are based on Complementary Metal-Oxide Semiconductors, which is a dominant semiconductor circuit for microprocessors, memories and application specific integrated circuits (ASICs). CMOS technology, leveraging the multi-billion dollar semiconductor industry enables low cost production of pixelized devices. Like CCDs, they are formed with crystalline silicon, but the read structure is individually addressed. Unlike CCDs, where charge is actually transferred across active pixel regions, CMOS technology has individually addressed pixels. CMOS image sensors draw less power than CCDs. However, they are known to produce more electronic noise than CCDs. Like CCDs, they can couple to various scintillators either directly, or by lens or fiber optics.

5.5.3 *Amorphous silicon read structures*—Larger amorphous silicon based thin film transistor pixelized read structures have been made commercially available as large flat panel devices. Figure 3 provides a schematic of an amorphous silicon DDA architecture. Amorphous silicon, through large area silicon deposition and processing/etching techniques offers a solution to the size constraints of CCDs and CMOS devices. Since the phosphor or photoconductor layer is typically deposited or coupled directly onto the silicon, efficient optical or electron transfer is easily obtained. However, the readout circuitry in these devices requires a large pixel space to accommodate the thin film transistor (TFT) and data lines and scan (gate) lines required for operation, thus limiting how small a pixel this device can permit. The amorphous silicon read structure is composed of over a million pixels that include photodiodes. The diode has a sensitivity that peaks in the middle of the visible spectrum where a number of good phosphors emit. The electric charges generated within every pixel of the photodiode are read by the active matrix of TFTs in place. The TFT matrix, which is essentially a matrix of switches, is scanned progressively. At the end of each data-line is a charge- integrating amplifier, which converts the charge packet to a voltage, followed by a programmable gain stage and an Analog-to-Digital Converter (ADC), which converts the voltage to a digital number that is transferred serially to a computer, where the data is formed into an $N \times M$ (N = number of columns and M = number of rows) pixel image.

5.5.4 *Choice of Read Structure*—For small field of view applications, the directly coupled CCD or CMOS approach will provide high spatial resolution and high light collection efficiency. As mentioned, these devices have pixel pitch, as fine as 10 microns. For large field of view applications, the amorphous silicon approach offers excellent collection efficiency (no lenses), in a thin, compact, robust package. However, pixel pitch is typically on the order of 100 microns or larger, although smaller pixel pitch structures are likely to appear in the near future.

6. DDA Properties

6.1 An important prerequisite for a good digital X-ray detector system is the capability of the system to control the interplay of all its components (the entire imaging chain) and reflect the capability of the system in the final image. The technology of image capture, the representation of images as digital data, their processing, enhancing of data for a specific image display, and the nature of the display technology, form a significant part of this capability. From an image interpretation standpoint, the quality of images from the detector is an important metric for the choice of the detector and system specifications. This section introduces the image quality parameters/metrics that form the basis for selection, and monitoring performance as delineated in Practices E2597, E2698, and E2737.

6.2 The dominant contributions to a digital radiographic image, and hence the final image quality, come from two sources: (*a*) the inherent property of a detector and (*b*) the radiographic technique itself. Some of the inherent properties of the detector which influence the image quality are, (*1*) signal and noise performance for a given dose, (*2*) basic spatial resolution, (*3*) normalized signal-to-noise ratio—SNR-normalized for spatial resolution, (*4*) detection efficiency, (*5*) detector lag (residual images, ghosting), (*6*) internal scatter radiation and (*7*) bad pixels. The other metrics such as (*8*) achievable contrast sensitivity, and (*9*) specific material thickness range are dependent on both, the DDA used as well as the object under test. Another strong factor is the radiation quality of the X-ray beam used for imaging.

6.2.1 A standardized methodology has been established for evaluating the inherent detector properties of DDAs as listed in 6.2 and may be found in Practice E2597. This practice provides procedures for evaluating and recording DDA properties by manufacturers or providers so that a potential purchaser may compare devices under standardized conditions and techniques in order to make an informed decision on the purchase. The ASTM standard suggests that providers of DDAs offer a spider diagram that summarizes the performance of a detector using a numerical grading scheme listed in the standard that highlights the strengths or weaknesses of a DDA. The purchaser can easily review those diagrams and decide what is most important for the application at hand.

6.2.2 Subections 6.3 to 6.19 provide additional details into these important detector properties, and how these impact overall performance of an inspection. Section 9 provides additional guidance into the selection of a DDA based on a review of the performance metrics taken together.

6.3 *Image Quality from a DDA*—The SNR of the DDA, using a specific radiation quality, and the relative contrast sensed by the radiation beam in the object together constitute an element of the image quality that relates to the contrast sensitivity of the DDA. The higher the SNR, the better, or lower the contrast sensitivity. A high signal to noise ratio improves contrast sensitivity as noise levels are suppressed in relation to signal differences. The SNR of a DDA system can be increased significantly by capturing multiple images with identical settings and integrating in a computer (frame averaging). The ability of the imaging chain to maintain the spatial information that originally impinged onto the primary detection medium is another critical element of the resulting image quality. This is typically referred to as the basic spatial resolution, SRb.

6.4 *Signal and Noise*—The signal recorded by a DDA is the response of the DDA to a given radiation dose. The noise is the variation of the signal read using the DDA for the same amount of dose. Signal and noise characteristics of the DDA depend on the radiation quality and the DDA structure. Radiation quality which is defined as the beam spectrum used, is directly related to the efficiency of the DDA that is related to the quantum efficiency of the scintillator. The higher the quantum efficiency of the scintillator, the higher the SNR will be. The DDA structure here refers to the type of scintillator used, type of signal conversion chain employed, and the associated electronics design. In an optimized DDA system where the DDA follows Poisson statistics, the noise is proportional to the square root of the signal level captured and thus the higher the efficiency of capturing and converting the radiation to a visible, or electronic signal at the DDA, the higher the performance of the DDA. For example with higher signal levels, the noise is reduced, and lower contrast, subtle features may be discerned in an image.

6.5 The transmitted X-ray beam signal propagates through various energy conversion stages of an imaging system, as discussed in 5.2. In Fig. 4, N_0 quanta are incident on a specified area of the detector surface (stage 0). A fraction of these, given by the absorption efficiency (quantum efficiency) of the material, interact (stage 1). Here it is important that the absorption efficiency is high, or a larger X-ray dose would be needed to arrive at a desired signal level. The mean number N_1 of quanta interacting with the scintillator represents the primary quantum sink of the detector. If we assume N_1 represents a measure of the signal, then the variance σ^2 is linearly proportional to N_1. Hence, the signal-to-noise ratio (SNR) is defined as $\sqrt{N_1}$. SNR therefore increases as the square root of the number of quanta interacting with the detector. Regardless of the value of the X-ray quantum efficiency, the maximum signal-to-noise ratio of the system will occur at this point ($SNR = \sqrt{N_1}$). If the signal-to-noise ratio of the imaging system is essentially determined there, the system is said to be X-ray quantum limited in performance. For example, performance will only improve if more X-rays are captured. The phosphor layer typically creates a large gain factor at this point. Following this, any subsequent inefficiency in emitting the light and capturing it by the photodiode will result in losses and additional sources of noise. If the number of quanta falls below the primary quantum sink, then a secondary quantum sink will be formed and becomes an additional important noise source.

6.6 For most detection systems discussed here, where the phosphor is in direct contact with the diode as in the flat panel detectors, the limiting source of noise is the quantum efficiency of the X-ray conversion material. Additional discussions on SNR of digital detectors are found elsewhere (**3**).

6.7 For direct conversion systems, the photoconductor is in direct contact with the read device, and with efficient charge transfer through the photoconductor into the read device, the limiting source of noise is the quantum efficiency of the X-ray conversion material.

6.8 Since noise is related to the square root of the number of X-ray quanta absorbed, it is crucial for efficient detection systems to have a sufficient signal level to avoid quantum mottling. Quantum mottling here refers to the variation in the signal level due to quantum noise. Quantum mottling makes detection of smaller contrast features more difficult. In medical imaging, regulations allow a certain maximum dose to the patient and optimal signal levels may not be obtainable. In this scenario, it is critical to absorb as many X-ray photons as possible, and then to transfer that energy efficiently, and not introduce secondary quantum sinks. On the other hand, in nondestructive testing, it may be possible to increase signal levels by selecting any or all of the following: (*a*) a longer

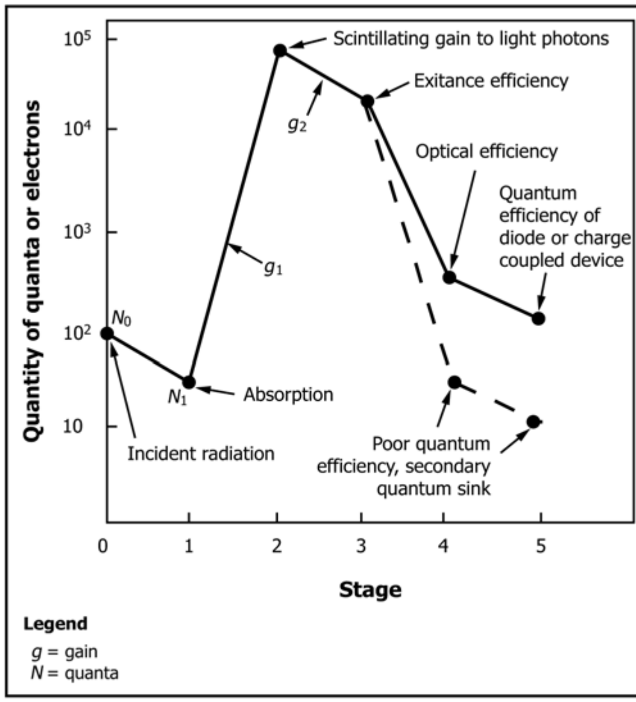

FIG. 4 Quantum Statistics of X-Ray Imager.

exposure time, (*b*) a combination of frames, either by integration or averaging, (*c*) a higher beam flux, (*d*) a higher radiation beam energy (assuming absorption is still high at those energies), (*e*) a closer working distance between source and detector, or (*f*) a different DDA with a more absorbing primary detection medium (phosphor or photoconductor). These techniques may provide improved image contrast due to higher SNR levels. Some of these techniques, however, may not meet other goals, such as throughput or allowable space needed for a specimen between the detectors and the X-ray tube. Certainly a thicker absorbing material (scintillator or photoconductor) may also impact the spatial resolution (see 6.12) possible from the DDA. Therefore, tradeoffs need to be made in selecting the appropriate DDA and technique to use for any given application.

6.9 Outside of the quantum chain discussed above, additive noise from the device in the form of fixed patterns, or other noise sources, or from the digitization process, can degrade an image even from the most efficient image chain. For a full discussion on noise sources, see (**3**). Therefore the noise of the device, as well as the coupling scheme is important in selecting the DDA for the application at hand. Appropriate calibrations (see Section 7) to remove fixed patterns within the DDA will result in drastically improved noise performance.

6.10 In a DDA system the detectability of a feature is defined in terms of contrast-to-noise ratio (CNR). Contrast in a radiographic image is mainly driven by subject contrast (see Practice E2597). DDA contrast sensitivity as mentioned above is dependent on the SNR of the device, and this contrast acts as a threshold limit for detection of subject contrast. When the subject contrast is below the DDA contrast, not enough information will be available to create a signal level in the resulting image for visual perception. Hence, CNR is related to subject contrast and noise in the imaging system.

6.10.1 Subject contrast, here referred to relative subject contrast that depends on the material properties of the object being imaged and energy of radiation used. To resolve a small change in thickness of an object (low subject contrast) and to achieve a high CNR, a high SNR of the imaging system is required. Additionally, improved detection of subject contrast can be obtained by using an optimized X-ray energy beam spectrum that best separates features in the object.

6.11 *Spatial Resolution*—The spatial resolution of the detector determines the detectability of features in the image from a pixel sampling consideration. The selection of the spatial resolution of the DDA is also important in designing or selecting a detection system. From the aspect of image contrast and spatial resolution, it is desirable to have the largest pixel that will allow detection of the features of interest in the radiographic examination. For example, it is not necessary to select a 10-µm pixel pitch if the application is for the detection of large foreign objects in an engine nacelle. Similarly, aircraft fatigue crack probability of detection will be low with a pixel pitch of 200 µm or larger, unless low unsharpness magnification techniques are used. See Fig. 5 for a discussion on selection of a DDA based on the size of the anticipated smallest defect, subject contrast, SNR, and the DDA pixel size.

6.12 *Pixel Pitch*—The predominant factor that governs the spatial resolution of a detector is the pixel pitch. Pixel pitch represents the physical dimension of the pixels. Most DDAs have square type pixels. As the pixel pitch is reduced for increasing the resolution, the total number of pixels in the image increases for a constant field of view. The file sizes for typical images run from 2 to 8 megabytes or greater. Other

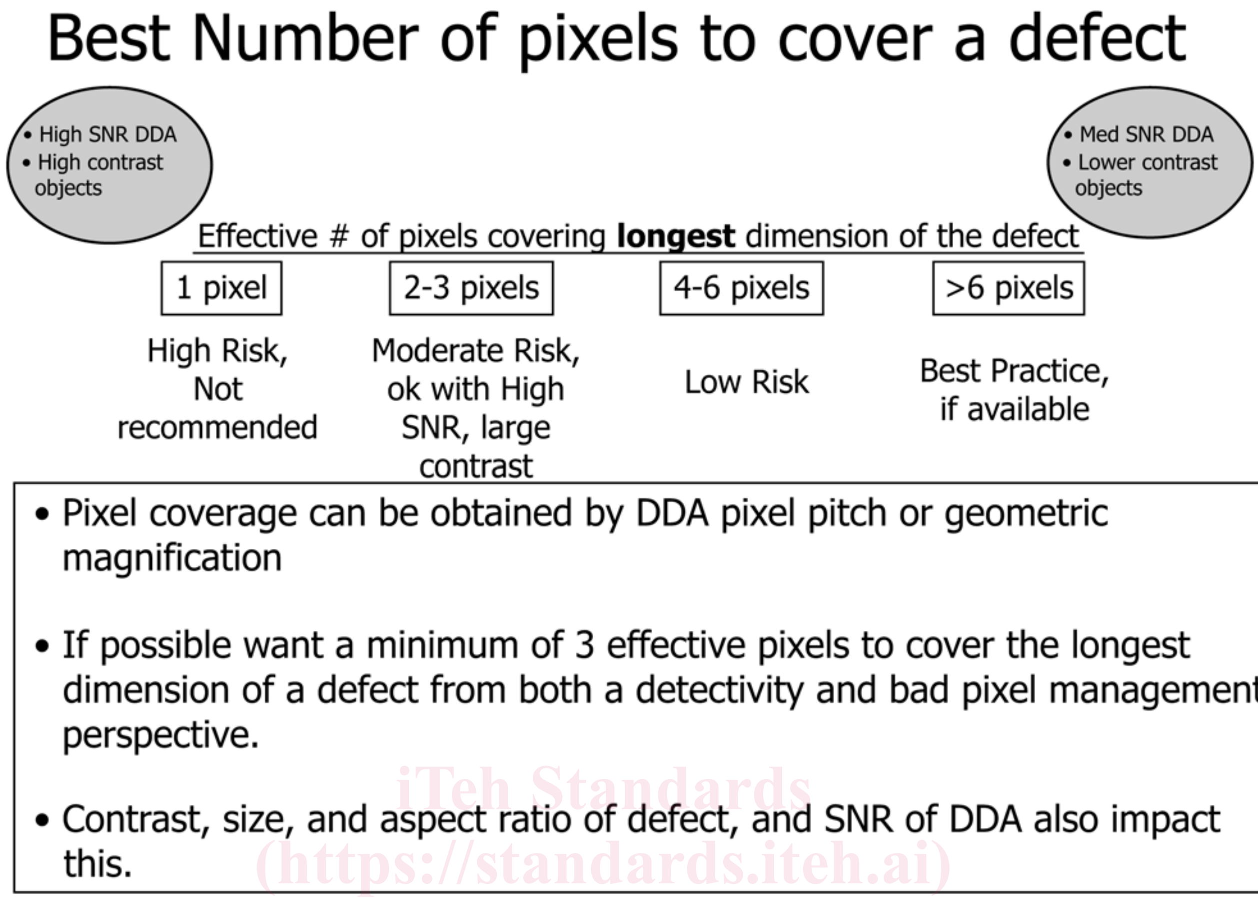

FIG. 5 Number of Effective Pixels to Cover a Defect Based on the Contrast of the Feature as Well as the SNR of the DDA. Single pixel coverage of the longest dimension of a defect is not recommended from the perspective of detection. It also may be confused for a bad pixel and missed.

factors that impact the spatial resolution of the image are (*1*) the geometric unsharpness of the inspection, (*2*) the thickness and properties of the scintillator or photoconductor material used to absorb X-rays, and (*3*) various sources of scatter that might degrade the modulation of features in an image. For a thick scintillator or photoconductive material, X-rays can scatter a greater distance depending on the X-ray energy employed and thus impact the spatial resolution. Optical spread can also occur in scintillation materials, especially thicker layers. In thick photoconductive materials, the bias levels to drive the carriers to the readout electrodes must also be high enough to avoid electron spreading that will degrade resolution. It is important to note, that the intrinsic spatial resolution of the DDA can never be higher than the pixel spacing. Magnification radiography is one means to compensate for the limitation in pixel pitch if the appropriate X-ray focal spot is available and can be used for the application at hand.

6.13 *Basic Spatial Resolution (SRb)*—The smallest geometrical detail, which can be resolved using the DDA. It is similar to the effective pixel size, and is typically expressed in µm. A means to measure the SRb is to use a duplex wire gage (see E2002), and measure the unsharpness, which in turn records the wire pair that can be seen in the image with 20 % contrast modulation. A contrast modulation of 20 % is usually assumed as a standard to determine if the the wire pair is visible. One half of the unsharpness value corresponds to the effective pixel size or the basic spatial resolution, as two pixels are typically required to resolve a wire (d) and its adjacent space (wire + space = 2d, the unsharpness). Figure 6 shows an example image of a duplex wire pair. The contrast modulation for the wire pair is the percentage dip in the signal. The SRb is calculated as the linear interpolation of the wire pair distances of the last wire pair with more than 20 % dip between the wires in the pair, and the first wire pair with less than 20 % dip between the wires (see Fig. 6). Where, D1 is the diameter of the smallest wire pair with >20 % resolution of the gap. D2 is the diameter of the largest wire pair with <20 % resolution of the gap. R1 and R2 is the modulation of the corresponding wire pair (dip %value) of D1 and D2 respectively.

6.14 *SNR and Pixel Size*—Among other factors, SNR is dependent on pixel area. A greater pixel area will typically result in higher SNR levels under identical exposure conditions. More specifically, assuming no other extraneous factors are dominant such as intra-scintillator or intra-photoconductor X-ray scatter that uniformly contaminates the signal without providing any spatial information, or a spatial frequency dependent fixed pattern noise, the SNR will increase by the square root of the pixel area if the X-ray conditions are held constant. A means to determine if these extraneous factors are present is to measure the SNR as a function of binning pixels.

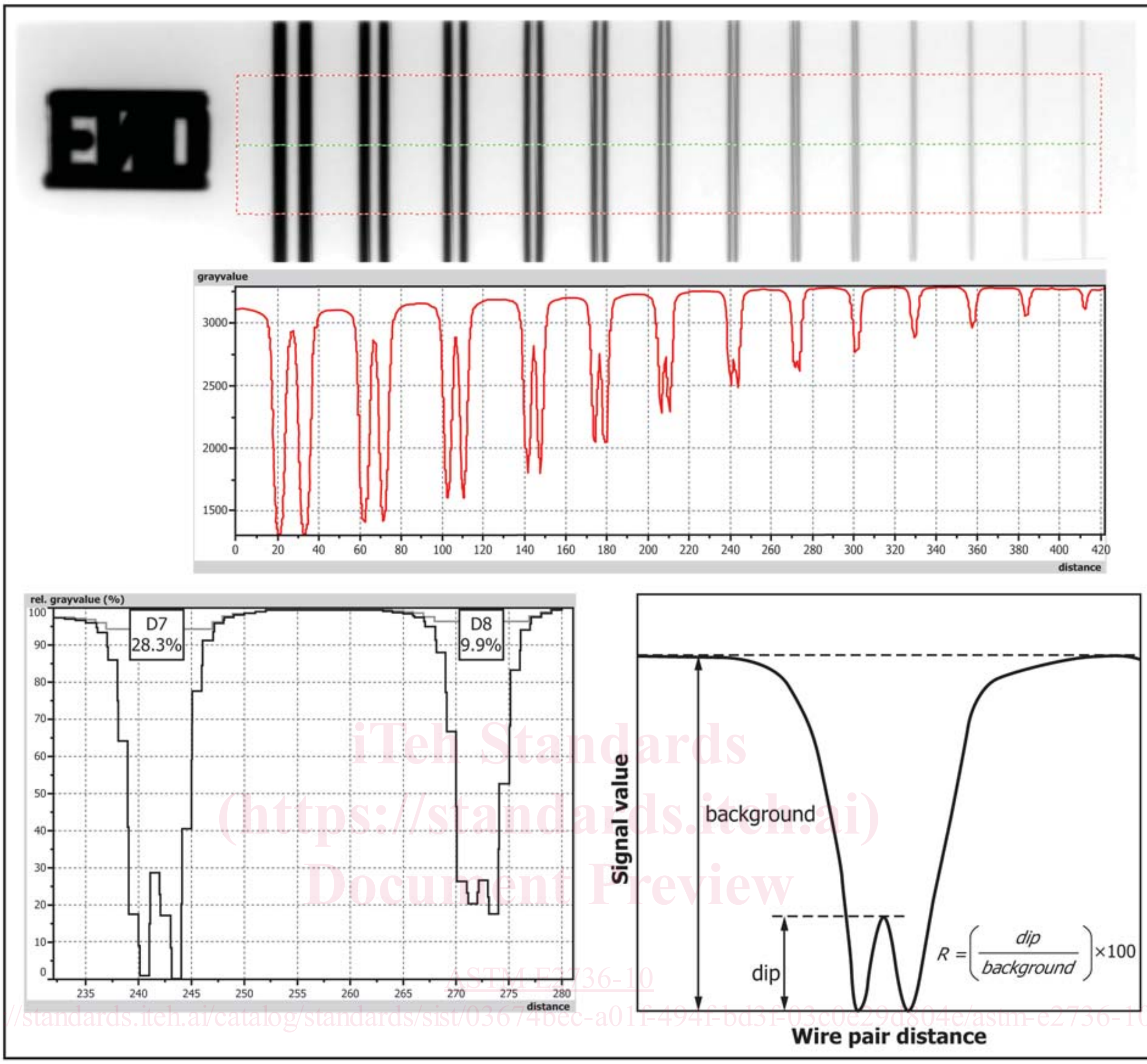

FIG. 6 Wire-pair Image Analysis for Calculation of Basic Spatial Resolution. Schematic of the measurement is shown at lower right.

If doubling the pixel size (quadruple the area) does not double the SNR (square root of the increased area), then some of these extraneous factors are present in the DDA.

6.15 *Normalized SNR*—To compare DDAs with different pixel architectures a first approximation can be made to normalize the SNR by the basic spatial resolution of the detector (SRb). Note: It is to be understood that this comparison might breakdown if the extraneous factors listed above are dominant (as the SNR and, or SRb values may be altered differently by those extraneous effects). For the normalization, 88.6 micron factor is used as the baseline value taken from the film normalization procedures in (see Test Method E1815). The circular aperture area for film densitometry is the same as the area of a digital square sampling box with 88.6 micron sides. Thus the DDA square pixel can be compared on a 1:1 basis to film. Hence the normalized SNR is computed as:

$$SNR_{norm} = SNR \times \left(\frac{88.6\ microns}{SRb} \right) \quad (1)$$

This same SNR_{norm} is also defined in the CR standards (see Practices E2445 and E2446), and is now in the DDA standards (Practices E2597 and E2698).

6.16 *Efficiency*—Efficiency of a DDA represents its speed to get to an SNR value. Typically this is expressed as a graph representing the dependency of SNR on incident dose to the DDA. A good measure of efficiency is the relationship between normalized SNR and the square root of dose incident on the DDA surface. This relationship should be linear. When the dose is set to 1 mGy, the normalized SNR at that point is the slope of the curve and represents an efficiency value for the beam quality employed. Figure 7 shows an example of efficiency of a DDA with various beam spectra. Each DDA has a peak efficiency, typically related to the thickness and absorptivity of the primary X-ray capture medium.

6.17 *Detector Lag*—Detector lag is a phenomenon where residual signal in the DDA is observed shortly after an exposure is completed and a “ghost” image is obtained. Lag in