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An Exposure Indicator for Digital Radiography

Report of AAPM Task Group 116

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An Exposure Indicator for Digital Radiography

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Abstract

Digital radiographic imaging systems, such as those using photostimulable storage phosphor (PSP), amorphous selenium, amorphous silicon, charge-coupled device (CCD), and metal oxide semiconductor-field effect transistor (MOSFET) technology, can produce adequate image quality over a much broader range of exposure levels than that of screen/film imaging systems. In screen/film imaging, the final image brightness and contrast are indicative of over- and underexposure. In digital imaging, brightness and contrast are often determined entirely by digital post-processing of the acquired image data. Over- and underexposures are not readily recognizable. As a result, patient dose has a tendency to increase gradually over time after a department converts from screen/film-based imaging to digital radiographic imaging. The purpose of this report is to recommend a standard indicator which reflects the radiation exposure that is incident on a detector after every exposure event and that reflects the noise levels present in the image data. The intent is to facilitate the production of consistent, high-quality digital radiographic images at acceptable patient doses. This should be based not on image optical density or brightness, but on feedback regarding the detector exposure provided and actively monitored by the imaging system. A standard beam calibration condition is recommended that is based on RQA5, but uses filtration materials that are commonly available and simple to use. Recommendations on clinical implementation of the indices to control image quality and patient dose are derived from historical tolerance limits and presented as guidelines.

Key words: digital radiography, direct digital radiography, indirect digital radiography, computed radiography, photostimulable storage phosphor, image quality, image noise, exposure index, quality control, quality assurance, acceptance testing.

I. Introduction

The charge of Task Group 116 (TG-116), as approved by the Science Council of the American Association of Physicists in Medicine (AAPM) was to identify a method of providing feedback, in the form of a standard index, to operators of digital radiographic systems, which reflects the adequacy of the exposure that has reached the detector after every exposure event. This report is the answer to that charge and will cover all digital radiographic image detector systems, leaving digital fluorography or fluoroscopy (radioscopy) for future consideration.

Unlike screen/film imaging, image display in digital radiography is independent of image acquisition. Inadequate or excessive exposure is manifested as higher or lower image noise levels instead of as a light or a dark image. The final brightness of the image is controlled not by the exposure to the detector, but by post-processing applied to the acquired image data. Consequently, overexposed images may not necessarily be dark, and underexposed images may not appear light. This may be a new and confusing concept for operators of digital radiographic systems who are accustomed to screen/film imaging.

For more than a decade, the phenomenon of "exposure creep" in photostimulable storage phosphor (PSP) imaging has been reported.[1,2,3] This is attributed to the fact that digital imaging systems can produce adequate image contrast over a much broader range of exposure levels than screen/film imaging systems. This broad dynamic range is one of the benefits of digital detectors. However, if the detector is underexposed, higher noise levels may obscure the presence of subtle details in the

image. Excessive detector exposures may produce high-quality images with improved noise characteristics but at the expense of increased patient dose. In extreme cases, excessive detector exposures may result in artifacts. As a result, most radiologists tend to complain about underexposed images but remain silent when images are acquired at higher dose levels unless apparent saturation has occurred. Technologists quickly learn that they can produce images of better quality if they increase their exposure techniques, resulting in less noisy images and avoiding radiologist complaints about noisy images. Average exposure levels tend to creep up over time if a clear indicator of exposure is not provided and routinely monitored.

Techniques required to achieve optimal radiographic imaging in Digital Radiography (DR) may be different from those used for screen/film imaging. In addition, different DR detectors may require different technique factors due to differences in the energy dependence of the detector materials in use (see Figure 1).[4] These differences in technique among DR systems may cause confusion and suboptimal image quality at sites where more than one type of system is in use. Operators need a clear set of rules to produce consistent, high-quality digital radiographic imaging, based not on image density but on feedback regarding the detector exposure provided.

Several manufacturers currently use an exposure indicator that parallels the concept of "speed" or "speed class" used by film manufacturers. In addition, many manufacturers and users have become accustomed to characterizing their systems as functioning within a given speed class. This has created some misunderstandings and scientific inaccuracies, which have been discussed in the literature.[5]

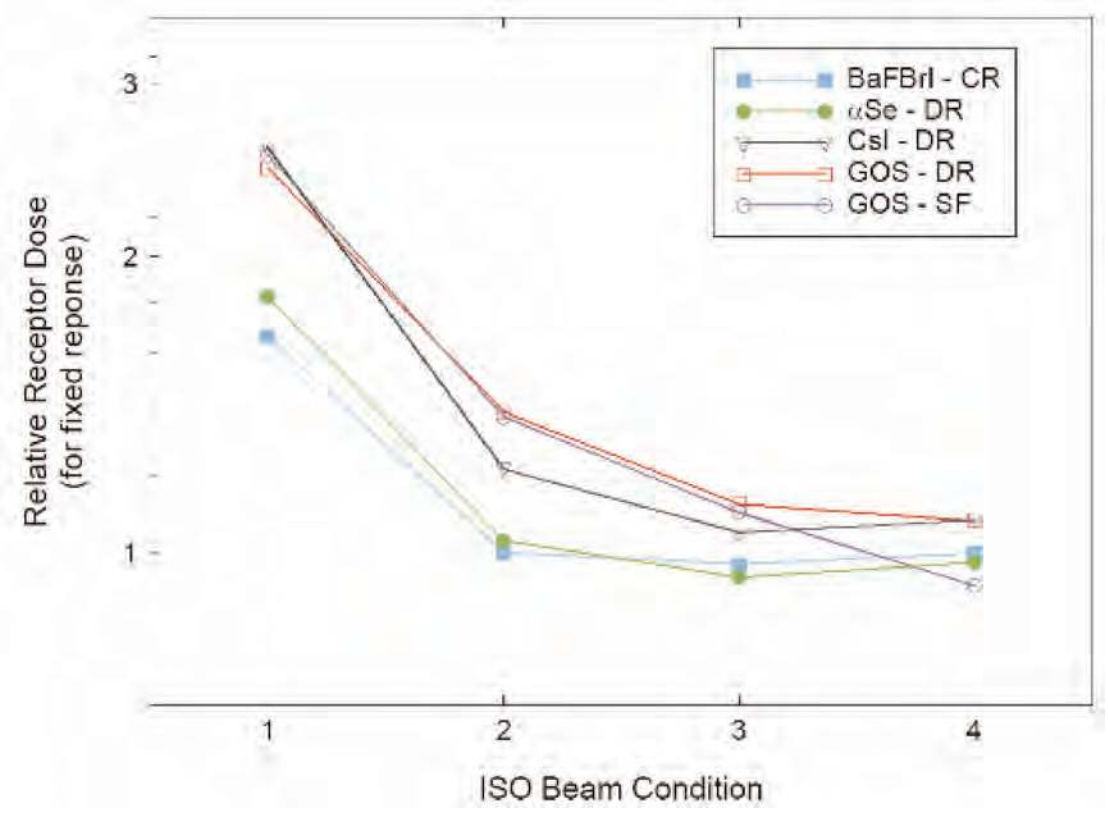

Figure 1. Energy dependence of common detector materials.

TG-116 recommends avoiding the concept of "speed class" when referring to DR system performance. The definition of radiographic speed according to ISO 9236-1[6] is based on the radiation exposure required to achieve a net optical density of 1.0 on the developed film. With DR there is no fixed relationship between the radiation exposure and the resultant density in the image. With screen/film detectors a change in speed may also result in a change in the spatial resolution properties of the detector. This same relationship does not hold true with digital detectors since sharpness is independent of the amount of exposure used to acquire the digital image.

The characterization of a digital radiographic system as being in a given speed class may give the false indication that it should always be operated at a specific exposure level. It may also give the false impression that the resulting digital image will have the same noise and resolution characteristics as that acquired with an equal speed class screen/film system. The digital system in reality can be operated over a broad range of sensitivity since the amount of radiation exposure determines only the level of quantum mottle and not the brightness of the image. From this context the level of radiation exposure, and thus the so-called "speed class," should be dependent upon the imaging task and upon the observer's tolerance of image noise. As a general rule, the As Low As Reasonably Achievable (ALARA) concept should prevail in that the minimum amount of exposure, and hence the maximum tolerable noise content, should be used to achieve the necessary diagnostic information.[7] Using the speed class characterization for given digital imaging systems may increase the possibility that ALARA is violated for some imaging tasks.

An index of detector exposure is appropriate because it is reflective of the noise content, and thus the signal-to-noise ratio (SNR) in the image. For DR systems, the appropriate incident exposure is variable based on the desired SNR rather than on the resulting optical density of a radiograph. Different digital detectors may require more or less radiation exposure to achieve the same noise content depending upon the detective quantum efficiency (DQE) of the detector technology in use. For a given system, the image noise content will track inversely with the detector exposure. As radiation exposure to the detector increases, image noise will decrease and SNR will increase.

A standardized indicator of the exposure incident on a DR detector that is consistent from manufacturer to manufacturer and model to model is needed. This could be used to monitor differences in exposure between DR systems at a given institution, to compare techniques between institutions, or to estimate the quality of images from a given radiographic system. It could also provide quality control (QC) data if software is provided to record and retrospectively analyze exposure data from all systems.

A standard indicator that reflects the radiation exposure that is incident on a detector after every exposure event is appropriate. The detector exposure indicator is intended to reflect the noise levels present in image data. An adequate exposure is one that results in an appropriate noise level in the image as determined by the clinic where the system is in use. This report does not make recommendations on exposure adequacy, nor does the indicator represent exposure to the patient.

TG-116 considers the recommendations in this report to be achievable and important. It recognizes that a parallel standard was recently completed within the International Electrotechnical Commission (IEC), designated IEC 62494-1 Ed. 1: Medical electrical equipment - Exposure index of digital x-ray imaging systems - Part 1: Definitions and requirements for general radiography. This IEC standard specifies the definitions and calibration conditions for the detector exposure indices of DR systems. The leadership of TG-116 participated in this IEC effort since its inception and served as U.S. National Committee experts in IEC Working Group 43. The concepts and calibration conditions

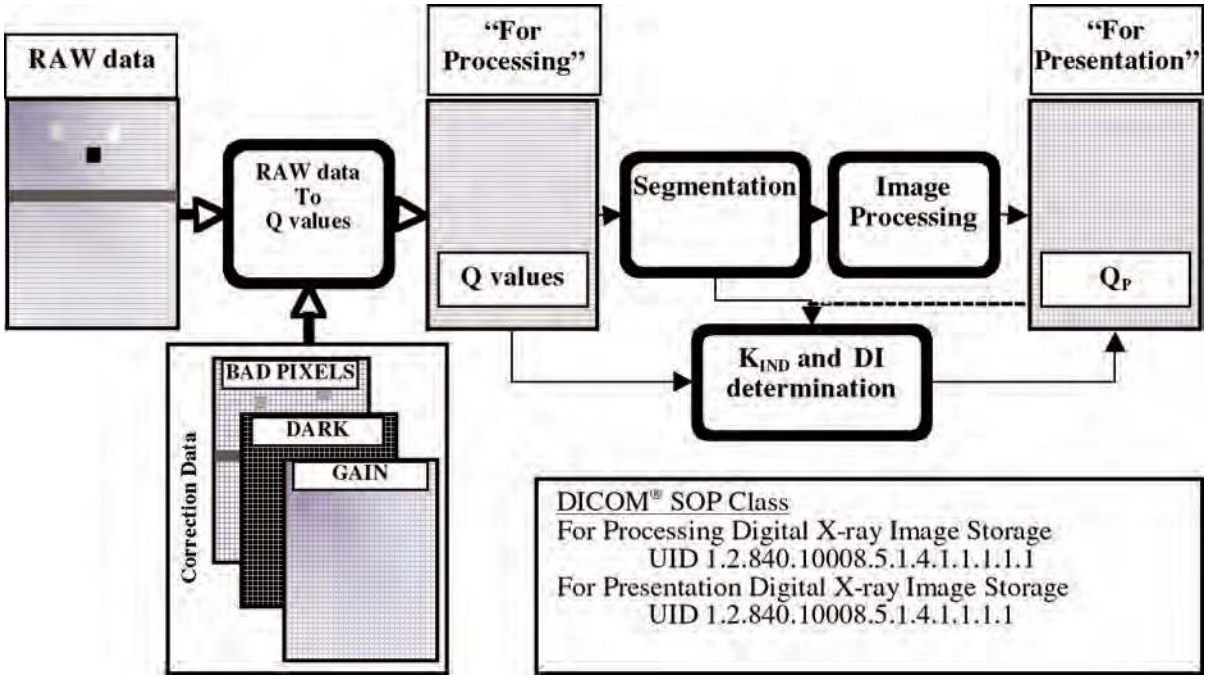

Figure 2. Essential processes in the acquisition of a digital radiograph. K_{IND} and DI are computed from Q values using segmentation information.

in the IEC working draft are consistent with those in this report. While the terminology and definitions in the IEC standard may differ in scale and nomenclature from those in this report, the IEC standard is completely consistent with this report. Absolute adherence to the nomenclature, symbols, and multiplicative factors of scale in this report are inconsequential to achieving the ultimate benefit of these recommendations as long as all manufacturers adhere to the IEC standard definitions. Users should be able to rely on a manufacturer's claim of conformance to the IEC standard to identify equipment offering a standard exposure index as described in this report.

II. Definition of Terms

DR systems utilize a series of computational processes to transform the raw data of the detector into an image intended for presentation. These processes include those used to assess the average response of the detector and its relation to the incident x-ray exposure.

The image formation process (Figure 2) begins with the extraction of raw data from the detector immediately following an exposure event. That data must be corrected for imperfections in the detector array such as the presence of bad pixel elements, dark current corrections, and gain corrections that may be applied on a pixel-by-pixel basis. After these corrections have been applied, the resulting pixel values are ready to be processed by the system and are referred to as "for-processing" pixel or "Q" values. The system then attempts to identify which of these pixels contain information that is of interest to the user, typically those that contain information relevant to the anatomy being examined. This process is called segmentation. It is from the segmented image values that TG-116

proposes the exposure indicators be determined. The final image for display results from grayscale transformations, broad area equalization, edge restoration, noise reduction, or other image-related processes that are performed on the "for-processing" Q values resulting in "for-presentation" Q_P values. Q_P values are typically stored in a Picture Archiving and Communications System (PACS) and transmitted to a printer or workstation for display. The remainder of this section defines the terms used in this document that relate to the DR processes just described.

Digital Radiography

Radiographic imaging technology producing digital projection images such as those using photostimulable storage phosphor (computed radiography, or CR), amorphous selenium, amorphous silicon, CCD, or MOSFET technology.

Standardized Radiation Exposure (K_{STD})

The air kerma at the detector of a DR system produced by a uniform field radiation exposure using a nominal radiographic kV_P and specific added filtration resulting in a specific beam half-value layer (HVL). (See section IV. Standard Radiation Exposure Conditions.)

For-Processing Pixel Values (Q)

For-processing pixel values are the image pixel values produced by a DR system after necessary corrections have been applied to the initially recorded raw data to compensate for these types of effects [see IEC 62220-1 Ed.1 for a complete description of appropriate correction methods].[7] The following corrections may be applied:

1. Defective pixels may be replaced by appropriate data.
2. Flat-field correction.
3. Correction for the gain and offset of single pixels.
4. Geometric distortion.

The relationship between Q and K_{STD} may vary for different DR systems. Manufacturers are expected to provide access to Q data and to provide information on this relationship as a part of normal system documentation. Images with Q values would typically be processed by the DR system in order to produce images for presentation.

Normalized For-Processing Pixel Values (Q_K)

Normalized for-processing pixel values, Q_K, are for-processing pixel values, Q, which have been converted to have a specific relation to a standardized radiation exposure (K_{STD}). Q values are converted to Q_K using the DR system's relationship between Q and K_{STD}. After conversion of Q to Q_K, the relationship between air kerma at the input surface of the detector and the Q_K value is

$$Q_K = 1000 * \log_{10}\left[\frac{K_{STD}}{K_0}\right] \quad (1)$$

where K_{STD} is in microgray units, $K_0 = 0.001\ \mu$Gy, and $K_{STD} \geq K_0$.

For-Processing Image Values (Q_P)

For-processing pixel values (Q) are typically modified by image processing to produce an image with values suitable for display (Q_P). This processing generally determines the useful values for display and applies a grayscale transformation. The processing may also provide broad area equalization, edge restoration, or noise reduction.

Indicated Equivalent Air Kerma (K_{IND})

An indicator of the quantity of radiation that was incident on regions of the detector for each exposure made. The value reported may be computed from the median for-processing pixel values in defined regions of an image that correlate with an exposure to the detector. The median value is then converted to the air kerma, K_{STD}, from a standardized radiation exposure that would produce the same detector response, i.e., result in the same median for-processing signal value, Q, in a predefined region of interest (ROI). The regions from which the median is determined may be defined in different ways (see section V. Assessment of Indicated Equivalent Air Kerma, K_{IND}). The value should be reported in microgray units with three significant figures of precision.

Image Values of Interest (VOI)

Pixel values in the original image (Q) that correspond to the region in the recorded image area for a particular body part and anatomical view. K_{IND} may be calculated from a subset of pixels within the VOI.

Not all pixel values in an image are associated with objects that are of interest to the viewer for the purposes of diagnosis. Those that are of interest are referred to as the "values of interest" (or VOI). The pixels that are associated with the VOI are typically identified based on their physical location and their relative signal strength characteristics. This identification process is referred to as segmentation. Specification of a standard method to be used for segmentation is not within the charge of this task group. Further, to recommend one standard method above all others may impede the development of more sophisticated methods that yield more stable results in the future.

Detector values suitable for presentation (Q_P) are typically sent to display devices (printers or workstations) or image archives. Digital Imaging and Communications in Medicine (DICOM) standards, including DICOM PS3.14, define these as presentation values, or P-values.[8]

Target Equivalent Air Kerma Value (K_{TGT})

The optimum K_{IND} value that should result from any image when the detector is properly exposed. K_{TGT} values will typically be established by the user and/or DR system manufacturer and stored as a table within the DR system. The table is referred to in this document as $K_{TGT}(b,v)$ where b and v are table indices for specific body parts and views.

Deviation Index (DI)

An indicator as to whether the detector response for a specific image, K_{IND}, agrees with $K_{TGT}(b,v)$. Relative exposure indices are to be reported as:

$$DI = 10 * \log_{10}\left[\frac{K_{IND}}{K_{TGT(b,v)}}\right] \quad (2)$$

with one significant decimal of precision (i.e., 0.0, 0.6, −1.3, etc.). DI is intended as an indicator for radiographers and radiologists for whether or not the technique used to acquire a radiograph was correct. This definition results in a DI of 0.0 when the reported K_{IND} equals K_{TGT} (a perfect exposure). The index changes by ±1.0 for each +25%/−20% change of the reported K_{IND}.

III. Recomendations

This report makes the following specific recommendations regarding the indicator of exposure for DR systems:

- It is recommended that all DR systems (regardless of detector design) provide an indicator of the x-ray beam air kerma (expressed in μGy) that is incident on the digital detector and used to create the radiographic image. This indicator shall be called the Indicated Equivalent Air Kerma (K_{IND}). It is further recommended that the DICOM standard incorporate a new element for DR that is specifically defined as the Indicated Equivalent Air Kerma. The indicator value shall be included in the DICOM header of every image as a floating point value with three significant figures.
- In addition to the Indicated Equivalent Air Kerma, it is recommended that the relative deviation from the value targeted by the system for a particular body part and view be reported. This index, the Deviation Index (DI), should be prominently displayed to the operator of the DR system immediately after every exposure and immediately after any modification of the detected image VOI. It is further recommended that the DICOM standard incorporate a new element for DR that is specifically defined as **the Deviation Index**. This indicator value shall be included in the DICOM header of every image as a signed decimal string value between −9.9 and +9.9 with one significant digit after the decimal.
- The Indicated Equivalent Air Kerma and the Deviation Index are determined from the segmentation image pixels (see section V. Assessment of Indicated Equivalent Air Kerma, K_{IND}). It is recommended that systems provide display functions to delineate the segmented image pixels optionally as an overlay on the recorded image that is otherwise normally presented for approval by the operator. Additionally, this overlay region can be incorporated in any images exported for archive or viewing using DICOM services. DICOM Segmentation Storage SOP Class (Supplement 111) forms the basis for achieving this functionality.[9] Alternatively, this could be accomplished with overlay and annotations that are part of Gray Scale Presentation State Storage objects described in DICOM Supplement 33.[10]
- Vendors should provide appropriate analytical tools (see section XI. Recommended Optional Features) and allow for-processing image data (Q values), or exposure values normalized to the standard beam conditions (Q_K), to be displayed and analyzed on the system console. It is also important that the vendor allow these data to be exported in DICOM format for off-line analysis. To accomplish this, all DR systems should provide access to images

containing for-processing pixel values, Q. This can be provided by support for DICOM export services of DX for-processing images containing normalized for-processing values, Q_K. Alternatively, images of either Q_K or Q can be made available in DICOM PS3.10[11] format on a media storage device.

- The relationship between Q_K values and the standardized radiation exposure incident to the DR detector is required for tests of system performance. It is recommended that this relationship be provided by the system manufacturer over the full range of radiation exposures that the system is capable of recording.
- For tests of system performance, it is useful to view and analyze the for-processing image values of acquired test radiographs. It is recommended that systems provide functions to display images without image processing (i.e., Q values) and to report the mean, median, mode, standard deviation, and pixel count of values within graphically defined regions. Interactively drawn circular or rectangular regions are appropriate for this purpose.
- For testing of systems, it is recommended that manufacturers provide methods to remove the anti-scatter grid without otherwise changing the detectors response or provide grid attenuation factors to be used in calibration.

IV. Standard Radiation Exposure Conditions

A. Standard Beam Spectrum

A uniform field radiation exposure made to the detector of a DR system is used to assess the relation between for-processing image values recorded by the detector (Q) and the quantity of radiation incident on the detector (K_{STD}). The radiographic technique used to make the exposure is intended to provide a beam quality typical of that for most examinations for which the system is used. This is done by using additional filtration to emulate the beam hardening of a patient. This section recommends standardized radiation conditions to be used for this purpose and addresses only general radiographic systems.

The IEC and ISO have previously made recommendations for standard radiation conditions for use in testing medical diagnostic x-ray systems.[6,12,13] A variety of conditions with different beam quality are recommended and labeled with "RQA" prefixes. However, these conditions require thick filters composed of 99.9% aluminum (Al), which is impractical for field measurements. Alloy 1190 falls into the category of scientific grade (also called ultra pure aluminum) and is available only through specialty metals companies for a high price, in small quantities, and limited form. Alloy 1100 is a 99.0% pure Al alloy that is widely available on the market.

The use of copper (Cu) as a component of the added filtration is recommended in order to reduce the overall thickness of added material. In prior publications, 0.5 mm of Cu was found to minimize the variability in the response of a CR system as kV_P was varied within 80 kV_P±10%.[14,15] The addition of Al material allows a HVL near the desired nominal to be achieved while keeping the Cu thickness at a value that is readily available from metal foil suppliers. The added Al material should be on the beam exit surface of the Cu so that any Cu characteristic radiation is absorbed in the aluminum.

Typically, clinical tubes in use at modern facilities contain enough inherent+added filtration to exceed the IEC open beam HVL specification of 2.5 mm Al at 70 kV_P (RQR5). If this is the case, the filtration to be added to the beam should be reduced to satisfy RQA5 by removal of all or part of the alu-

minum. If the open beam HVL falls below the specification of 2.5 mm Al at 70 kV_P, the filtration to be added to the beam should be increased to satisfy RQA5 by addition of up to 4 mm Al. If the unfiltered beam is generated using exotic added filtration, it is recommended that added filtration be replaced with enough aluminum to meet the requirements of RQR5. The kV_P may also be adjusted, if necessary.

Added filtration with 0.5 mm Cu plus 3–4 mm Al is suitable for x-ray tubes with modest intrinsic filtration. For an unfiltered x-ray tube spectra with HVL of 2.58 mm Al at 70 kV_P (RQR5), computational simulations (see appendices A and B) indicate that a similar beam quality with HVL = 6.8 mm Al is obtained using added filtration of either 21 mm of pure aluminum as specified for RQA5, 0.5 mm Cu plus 3 mm Al (type 1100), or 24 cm of muscle.

The following types of brass (30%–50% zinc with traces of tin) are also considered acceptable to be used in place of the copper: admiralty brass (30% zinc and 1% tin), alpha brass (less than 35% zinc), alpha-beta brass or duplex brass (35%–45% zinc), arsenical brass or DZR brass, beta brass (45%–50% zinc), cartridge brass (30% zinc), common brass or rivet brass (37% zinc), high brass (35% zinc), low brass (20% zinc), naval brass (40% zinc brass and 1% tin), red brass (gunmetal), or yellow brass (33% zinc). Use of leaded brass, aluminum brass, white brass, or gilding metal is not recommended.

For the first edition of IEC 61267, kV_P was to be adjusted to achieve a desired beam HVL.[11] For the second edition, more stringent constraints were placed on the beam quality before added filtration rather than allowing kV_P adjustments. As a consequence, the conditions recommended in the second edition are applicable only to laboratory facilities. It has also been reported that CR system response varies by approximately 10% over a limited a range of kV.[14] Measurements made by task group members on a variety of clinical systems (both CR and DR) indicate that system response is very consistent within the recommended ranges of HVL, kV, and added filtration (see appendix B). System response in terms of for-processing signal per unit exposure varied by less than or equal to 6% with kV_P/filter combinations as low as 67 kV_P with 0 mm Al (5.93 mm Al HVL) to 73 kV_P with 8 mm Al (8.62 mm Al HVL). Similar results have been found by others and reported in the literature (see Figure 1).[4] For this limited range of kV, 5% tolerance seems reasonable.

While use of the RQA5 beam is acceptable for calibration, TG-116 recommends standard beam conditions using copper foil and highly available type 1100 aluminum with a specified kV_P range whose accuracy has been independently verified to be within 3% of the indicated value (see Table 1). The target HVL is intended to be reasonably close to RQA5 (6.8±0.25 mm Al). Minor adjustments in indicated kV_P and added filtration are permitted to achieve the target beam quality. While not required, it is acceptable to vary the kV_P by up to ±5% and the amount of added aluminum within the listed range to achieve a beam quality that is as close as possible to the listed target HVL. The generator used for the standard beam must be capable of maintaining a constant kV throughout the entire exposure.

B. Standard Beam Geometry and Calibration Verification Procedure

This section describes the measurement geometry to be used to determine K_{STD} under the standard radiation exposure conditions shown in Figure 3. The steps to use when making these measurements are summarized below.

1. Prior to any measurements verify that the x-ray source has acceptable exposure reproducibility (coefficient of variation < 0.03) and kV accuracy (±3%) at the standardized

Table 1. Standard Beam Radiographic Conditions

kV_P	*Added Filtration*	*Nominal HVL*	*IEC Surrogate*
66-74	0.5 mm Cu + (0-4) mm Al* or 21 mm pure Al	6.8 mm Al*	RQA5

*Type 1100

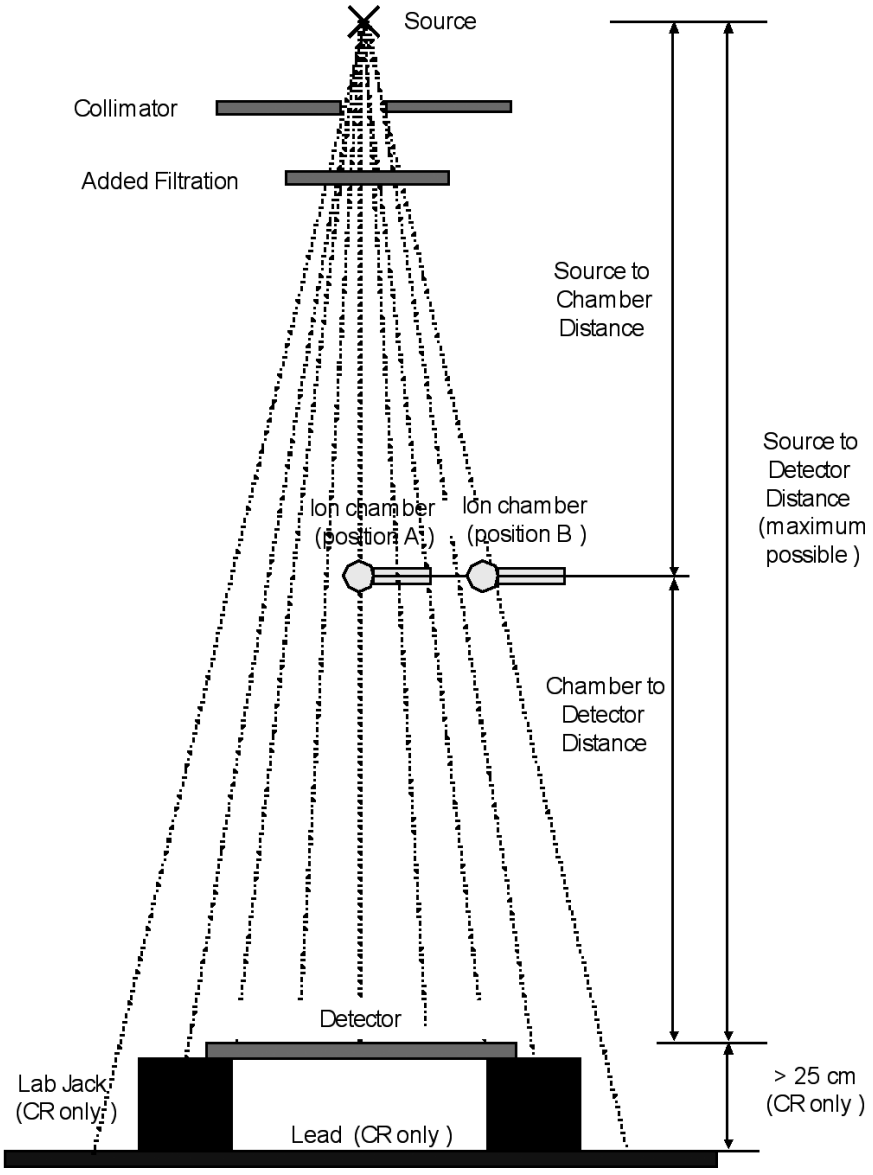

Figure 3. Standard beam geometry.

condition. During any exposure for which an image is not acquired, the image receptor should be protected from exposure to the primary beam.

2. The detector should be placed as far from the x-ray source as is practical, at least 100 cm.
   a. If the detector is a CR plate, the cassette should be separated from any surface that may increase backscatter from that surface entering the cassette (see Figure 3), as recommended in AAPM Report No. 93.[15]
   b. If present, and if possible, remove the anti-scatter grid and any other system components present between the ion chamber and the image detector without otherwise modifying the response of the detector. If any components cannot be removed, obtain the attenuation factors from the DR system or component vendor.
   c. If the detector is not square, the long axis of the detector should be perpendicular to the x-ray tube anode-cathode axis.
3. Place a calibrated ion chamber at the center of the beam approximately midway between the source and detector (see position A in Figure 3). The distances should be measured to the center of the chamber and to the surface to the detector. The distance from the x-ray source to the center of the ion chamber and the surface of the detector must be known. If the distance from the detector housing surface to the detector is not labeled, consult the manufacturer for this measurement.
4. Collimate the x-ray beam to only cover the ion chamber with no more than 1 in. margins.
5. Add 0.5 mm Cu filtration at the face of the collimator.
6. Cover the detector with a lead apron or similar barrier to preclude exposure to the detector. Measure the HVL of the filtered beam and adjust the kVp and/or Al filtration within the limits specified in Table 1 to obtain a HVL as close as possible to 6.8 mm Al.
7. Make an exposure and determine the air kerma at the detector (K_{STD}) using an inverse square correction and applying the grid attenuation factor, if appropriate. Repeat the exposure at a new mAs setting that will deliver the desired air kerma to the detector. In general, the desired air kerma will produce a value of K_{STD} that is in the middle of the detector response range.
8. Move the ion chamber perpendicular to the tube axis such that it is outside the detector field of view (see position B in Figure 3).
9. Remove the filtration and open the collimator so the x-ray beam will cover the entire detector and includes a margin large enough to cover the ion chamber. If the system does not allow the collimator to be opened beyond the detector size, open the collimator as large as possible and place the ion chamber as close to the edge of the x-ray beam as possible within the field of view of the detector. Ensure that the entire ion chamber is in the radiation beam and is not shadowed by a collimator blade. Replace the filtration.
10. Remove the protective lead apron and make an exposure using the mAs found in step 7 above. Determine the ratio of the air kerma at position A to that at position B.

When making standardized radiation exposures using this geometry, the air kerma recorded by the ion chamber is converted to K_{STD} for each exposure using the K_A/K_B ratio determined and the inverse square correction.

Some manufacturers have specified other requirements in addition to beam quality, such as readout time delay after exposure with CR systems. These requirements should be adhered to as long as the standard beam conditions specified in this part are not affected.

V. Assessment of Indicated Equivalent Air Kerma (K_{IND})

It is expected that manufacturers of DR systems will establish the relationship between for-processing image values (Q) and standardized radiation exposure (K_{STD}), i.e., Q as a function of K_{STD}. This relationship should be specified over the full range of exposures for which the system is designed to respond. If individual systems vary in response, information provided with the system should include the acceptable variation specific to a particular system. As a part of acceptance testing, physicists may wish to verify this relationship by recording images of a uniform field obtained using standard beam exposures made with an appropriate set of mAs values.

For validating Q as a function of K_{STD}, the incident air kerma should be measured using the methods described in section IV.B for which K_{STD} reflects the radiation incident to the central region of the detector. For each image recorded, either the for-processing image pixel values, Q, or the normalized for-processing image pixel values, Q_K, should be analyzed to determine the median value from a central region of interest ($<Q>$ and $<Q_K>$).

For determining the indicated equivalent air kerma from an individual clinical image, K_{IND} is computed as the K_{STD} corresponding to the $<Q>$ value in a defined region of a recorded image (see Figure 4). The median of image values is computed and transformed using the known relationship between for-processing image pixel values (Q) and exposure. The median is recommended rather than mean or mode because it is less affected by data extremes and outliers. Rectangular or circular regions having an area greater than or equal to about 4% of the active detector area should be used. The median value can be determined using vendor-supplied analysis tools designed specifically for evaluating the test image or by exporting the test image as a DICOM object to an external workstation for evaluation. The median Q value and the median K_{STD} value are thus the same as long as the transformation is monotonic. Additionally, the median value can be determined from the histogram of values within the defined region.

The region used to compute K_{IND} should be defined such that the indicated equivalent air kerma reflects the median exposure within the segmentation image pixels in the recorded image. The segmentation image pixels will vary depending on the purpose of the radiograph. For example, the primary anatomic region of interest in a chest radiograph is the lung parenchyma, whereas the mediastinal and subdiaphragmatic portions of the image would be secondary regions. However, the mediastinum would be a primary region for a thoracic spine radiograph. Hence the segmentation image pixels for the PA Chest exam and for the PA T-Spine, even if collimated identically, would not comprise the same set of image pixels.

For some existing systems the segmentation image pixels are defined by the portions of the image for which body tissue has attenuated the beam. Unattenuated regions of direct exposure are excluded along with regions outside of the collimated primary beam that receive only scattered and off-focus radiation. Other systems have used geometric regions (circles, rectangles, etc.) positioned in the general area of the primary anatomic region. These can be systematically placed in the field such as for the position of a central Automatic Exposure Control (AEC) cell.

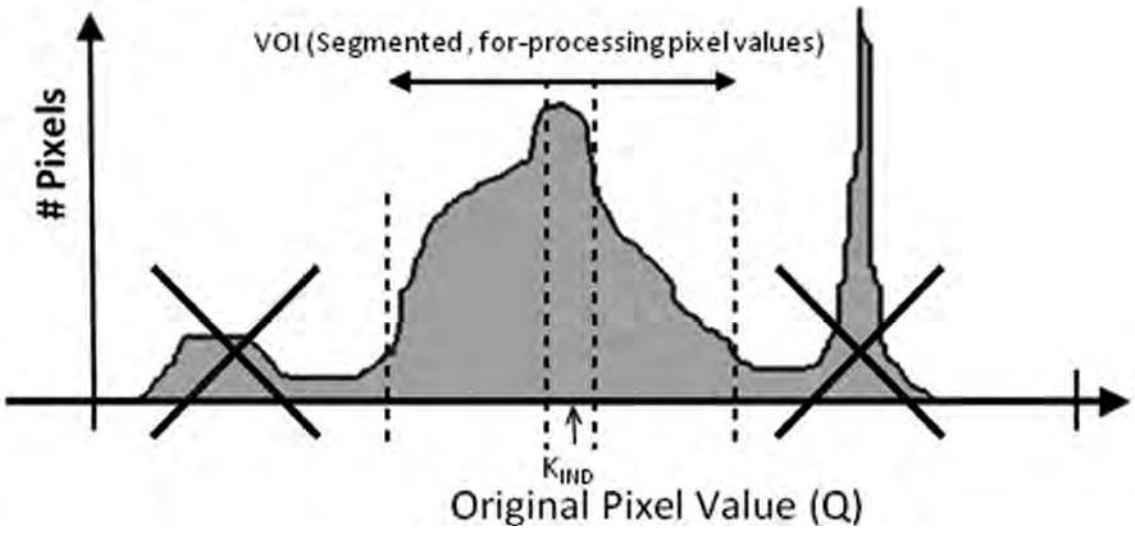

Figure 4. Example of a method to determine K_{IND} from an analysis of the image histogram.

More robust scene recognition algorithms may be used to identify the segmentation image pixels. Robust region definition methods typically require advanced image segmentation algorithms that have generally not been fully disclosed by manufacturers. In most cases, these methods occasionally fail under certain clinical conditions. To aid users in identifying recordings for which the segmentation may have failed, it is recommended that systems provide functions to display an overlay of the segmentation image pixels. If provided, this overlay should be accurately registered to the image pixels. Additionally, methods to modify the selection of segmentation of the image pixels after the automated segmentation algorithm is performed should be provided.

For many systems, region definition is used to identify that portion of the image that should be rendered in the mid-portion of the grayscale transformation. In recent reports, it has been suggested that the "for-presentation" image pixel values (Q_P) be used to define the region for computation of K_{IND}.[16] In this work, pixels of the for-presentation image (Q_P) within a fixed range of presentation values are used to define the region for computation of K_{IND} (see Figure 5). For example, presentation values from 45% to 55% of the full range of values are in the mid-gray regions of the image, which normally correspond to the anatomic regions of highest interest to be rendered with maximum contrast. The value of K_{IND} is computed from pixels in the for-processing image that correspond to this range. Regardless of the method used to define the region that is used to compute K_{IND}, its value should reflect any changes to the segmentation image pixels that are made by the operator.

This report does not make recommendations as to how the segmentation image pixels are to be defined. Rather, the scope of recommendations is restricted to recommendations directed at standardizing the terminology and beam conditions associated with reporting indices of exposure. It is expected that conformance in these areas can be achieved in the near future. The task group recognizes that the defined region from which K_{IND} is computed has strong influence on the result. A completely satisfactory solution for this difficult engineering problem remains unidentified. In light of this, the consensus of the task group is that it would be inappropriate to limit further development of

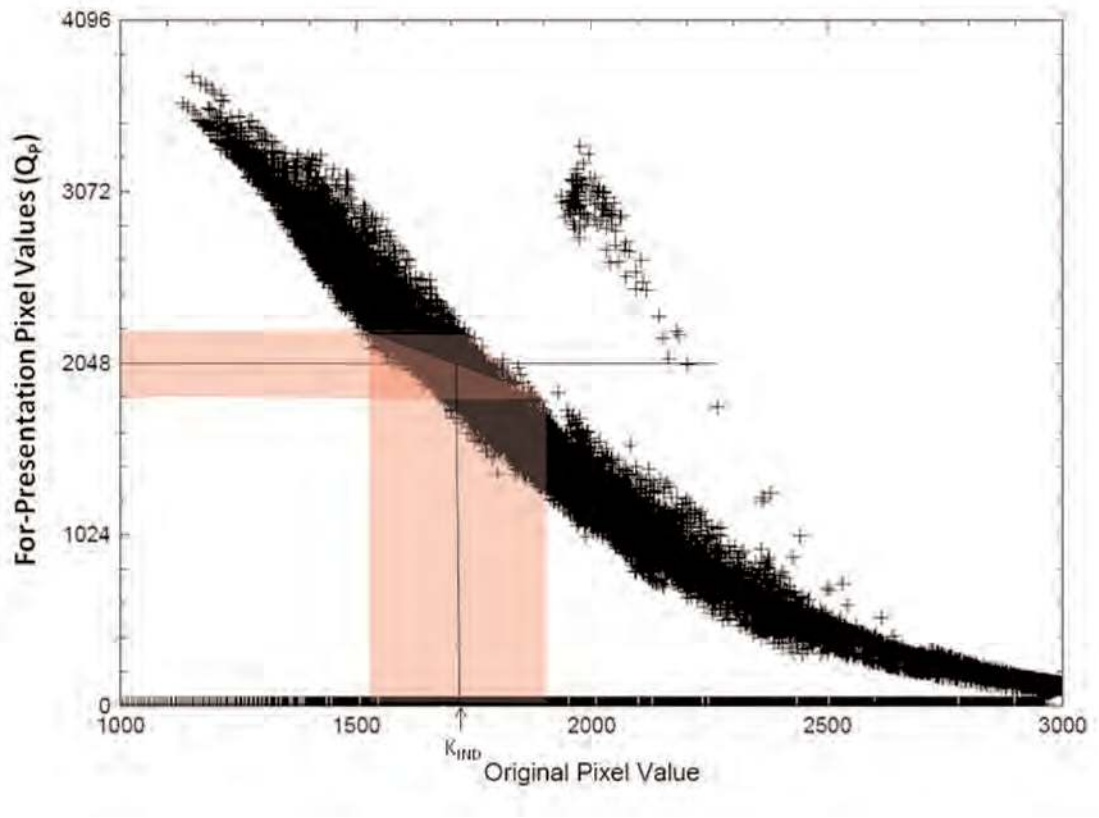

Figure 5. Example of a method to determine K_{IND} from Q_P of a processed image.[16]

more robust segmentation algorithms by establishing a standard method for defining segmentation pixels at this time.

The Indicated Equivalent Air Kerma, K_{IND}, is an indicator of the detector response in regions where anatomically important tissues have been recorded by a DR detector. K_{IND} is not equal to the incident exposure for the radiograph recorded. Rather, it is associated with the incident exposure from a standard reference beam that would have produced the same detector response. For this reason, it is referred to as an "equivalent" air kerma. For the general radiography standard beam conditions, the incident exposure required for the same receptor response in a typical DR detector varies modestly for peak tube potential values in the range from 55 to 90 kV_P.[4] In general, the actual incident exposure required to produce the same detector response will change with varying tube potential for radiographs of a specific object.

VI. Reporting Deviation Index (DI)

For radiographs of different body parts and/or views, the value of K_{IND} required to obtain acceptable image quality may vary. Additionally, the purpose and clinical diagnostic indications expected for a particular procedure may influence what is considered acceptable. For this reason, it is recommended that manufacturers automatically reference the appropriate standard beam condition (based on body

part and anatomical view) when determining K_{IND}, and deduce the recorded relative exposure from the appropriate K_{IND} in relation to that targeted for the body part and view of the radiograph.

The DI is intended to indicate the acceptability of SNR conditions in the segmentation image (as compared to high or low) to persons performing or interpreting radiographic examinations. How this index is calculated and the information displayed to these groups has an influence on how it is interpreted. Several options were considered by the task group for the nature of this index. Some were of the opinion that an index that varies linearly with $K_{IND}/K_{TGT}(b,v)$ would be more understandable to both radiologists and technologists. However, this approach suffers from the fact that such an index would asymptotically approach 0 as exposures decreased to 0, thus minimizing the apparent impact that underexposure has on image quality. Another consideration is the fact that image noise is logarithmically related to exposure. For underexposed images, use of a linear indicator would not reflect the magnitude of the change necessary to bring about a corresponding improvement in noise. It was decided that a logarithmic scale in base 10 would provide appropriate information in terms of both direction (over- or underexposure indicated by a positive or negative value, respectively) and magnitude (+1 is approximately 125% of the intended exposure, –1 is 80% of the intended exposure) on needed technique corrections. An exposure resulting in a DI value of +1 would require an adjustment of –1 step on the density or mAs control of a properly calibrated modern radiographic system.

Tables of targeted values may be provided by manufacturers with values reflecting typically acceptable K_{IND} values for the detector technology being used. Typically, these will be lower for detector technology that has higher DQE. Provisions must be available for imaging centers to adjust the K_{TGT} values based on an individual facility's criterion for image quality. Systems should provide a mechanism to export and import tables in a consistent format so that tables could be shared between imaging facilities using the same DR system. A process for updating the tables of all systems within a facility that is managed via a network would be extremely valuable so that changes in K_{TGT} values can be readily disseminated to distributed systems. DR systems should also provide the means to save the associated $K_{TGT}(b,v)$ for any body part and view as the default target value automatically when invoked by an appropriately privileged operator.

VII. Clinical Use of the Deviation Index (DI)

The clinical use of the DI is essentially the same as that of film optical density: it serves as an indicator of proper radiographic exposure. For screen/film images, the optical density of the image itself is used to indicate proper exposure according to the clinical preferences of the facility. By de-linking image appearance (in terms of brightness or contrast) from the amount of radiation exposure used to produce it, digital imaging alleviates the dynamic range limitation suffered by film. The drawback is that the direct visual feedback as to proper exposure is also severed. As has been noted before, the result can be widely varying clinical techniques, with consequences to both image quality and patient radiation exposure. The primary concern with DR image quality as it relates to detector exposure is with image noise (quantum mottle). DR post-processing and "QC" workstations generally utilize displays of lower resolution (1024×1024 or less), lower brightness, and capable of rendering fewer gray levels than those to be used for primary diagnostic interpretation. These "secondary"[17] workstations are also rarely calibrated to DICOM PS3.14. As a result, it is often the case that image noise is not well appreciated on such displays. What might appear acceptable on the QC workstation

may be diagnostically unacceptable to the radiologist reading on a display calibrated to DICOM PS3.14. The DI can be used clinically to ensure that the amount of radiation delivered to the detector and, hence, the noise content in the image, is appropriate for a given imaging task.

The reader should be cautioned that possible system errors in the segmentation may cause the DI to incorrectly reflect the adequacy of the exposure. Causes of errors in segmentation will vary by vendor—some examples include the presence of prostheses or gonadal shielding, failure in identification of collimated area due to scatter or off-focus radiation, and unexpected positioning of a body part in the field of view. Poorly defined collimation may lower the DI, depending on the exam and projection. Undercollimation that causes unusually large amounts of the image to be unattenuated by the patient may increase the DI. Both of these interferences can result in a false DI. If any of these is the case and the reported DI is out of acceptable range, a repeated exposure may not be necessary. The technologist should closely scrutinize every image for noise content, using zoom and pan utilities if available. DI values should be treated more like a guide instead of an absolute measure of image quality. Until segmentation becomes more robust, the technologist should judge image quality on noise content and not depend solely on the exposure index value.

VIII. Exposure Indicator and Radiographic Techniques

The K_{IND} indicator serves as a means of establishing appropriate radiographic techniques that might otherwise drift widely from desired levels. Adhering to target ranges for the particular DI values can be a valuable tool for standardization and stabilization of manual techniques. For departments involved in clinical aspects of radiologic technology training programs, DI can also be used as an aid to instruct students in proper manual technique selection and for evaluation of trainee performance in this regard.

DI values are determined for each body part and anatomical view being imaged on an exposure by exposure basis by comparing the K_{IND} value for a given exposure to the target $K_{TGT}(b,v)$ values stored on the system. These $K_{TGT}(b,v)$ values are the optimal exposure values determined either by the vendor or by the site system administrator for each body part and anatomical view being imaged. The $K_{TGT}(b,v)$ values should be set according to clinical preferences and specific exam needs. This

Table 2. Exposure Indicator DI Control Limits for Clinical Images

DI	*Range Action*
> +3.0	Excessive patient radiation exposure Repeat only if relevant anatomy is clipped or "burned out" Require immediate management follow-up.
+1 to +3.0	Overexposure: Repeat only if relevant anatomy is clipped or "burned out"
–0.5 to +0.5	Target range
Less than –1.0	Underexposed: Consult radiologist for repeat
Less than –3.0	Repeat

approach is consistent with maintaining exposures in a range that is as low as reasonably achievable (ALARA). The $K_{TGT}(b,v)$ values associated with each body part and view, if present, should automatically be invoked when the body part and view for post-processing are selected by the operator at the processing system console.

Once $K_{TGT}(b,v)$ levels are set, it is useful to identify several types of "control limits" on DI: a target range, a "management trigger" range, or a "repeat" range (see Table 2). The reason for this is that unlike filmed images, in which inadequate or excessive image optical density is a determinant of when a repeated film is needed, the reason for repeating a digital image is primarily noise related. What would be an underexposed film image may be of adequate diagnostic value in digital form. Similarly, it is never appropriate to repeat overexposed digital images unless analog-to-digital converter saturation has occurred, which may cause relevant parts of the image to be "burned out" or "clipped" (that is, all pixels in the affected region are forced to the maximum digital value and thus containing no information) or contrast to be affected in excessively exposed regions of the image. Since this judgment depends upon the diagnostic task, it is appropriate to seek consultation with a radiologist for certain ranges of DI-indicated under- and overexposure prior to repeating.

A properly functioning AEC system will produce optical densities of ±0.15 OD under varying combinations of kV_P and phantom thickness (adjusting mA to result in exposure times greater than 10 ms).[18] For a screen/film combination, optical density in the straight-line portion of the H&D curve is related to detector exposure (or K_{IND}) as follows:

$$\Delta OD = \gamma \log_{10}\left(\frac{K_{IND_2}}{K_{IND_1}}\right). \tag{3}$$

Combining equations (2) and (3), the range of deviation indices corresponding to a given OD range would be:

$$\Delta DI = \left(\frac{10}{\gamma}\right) \times \Delta OD. \tag{4}$$

For a screen/film combination with a gamma of 2.85 (a fairly common gamma in clinical use), the acceptable DI range would be:

$$\begin{aligned} \Delta DI &= \left(\frac{10}{2.85}\right) \times 0.3 \\ &= 1.05 \\ &= \pm 0.5. \end{aligned} \tag{4a}$$

The task group recommends the action levels shown in Table 2.

To be effective, care must be taken to assure that appropriate targets and limits are posted and the radiographers are educated and periodically re-educated as to their meaning.A substantial deviation from the established target range should require management oversight to determine the cause for the deviation and to implement appropriate corrective action such as retraining, recalibration of the equipment, or reassessment of the target value. Operators should be instructed that high DI values are associated with excessive radiation dose but have good image quality with respect to noise. Tighter limits on DI may be difficult to achieve in practice due to variations and drifts in CR reader

calibration (especially with multiple readers), variations between detectors, as well as traditional differences between x-ray rooms (generator design, calibration, and tube filtration).

IX. K_{IND} and Automatic Exposure Control (AEC) Systems

In regard to maintaining appropriate image quality and patient exposures, it is clear that AEC systems are just as important to digital imaging as for screen/film imaging, despite the wide dynamic range of DR. Regardless of detector type, AEC systems are designed to (and must be appropriately calibrated to) terminate an x-ray exposure once a predetermined radiation exposure is recorded at the detector. Like screen/film systems, digital detectors have energy dependence, which in general differs from that of the AEC sensors (see Figure 1).[4] Depending on design and calibration of the AEC, the result can be digital image levels that vary from the desired level.

A well-designed AEC should be capable of modifying required detector exposures based on exposure conditions (typically selected kV_P and mA) to compensate for energy dependence and exposure rate and thereby maintain a consistent image SNR.[19] Assuming that AEC performance is evaluated under clinically relevant conditions which can be simulated by various thicknesses of acrylic and kV_P ranging from 60 to 120,[19] the K_{IND} can serve as the indicator of image signal level for this purpose, just as optical density did for film. Adequate AEC performance with screen/film is considered to be a density variation of ±0.15 OD on a system using film with a gamma of 2.5. For equivalent performance on a DR system using AEC, one would expect K_{IND} to remain constant with varying thickness and kV_P (adjusting mA to result in exposure times greater than 10 ms) within ±7%.

In using K_{IND}'s during AEC performance evaluation, several caveats must be noted. First, the K_{IND} may be associated with a different image region from that used by AEC sensors; second, the size of the area used by K_{IND} determination may introduce different field size and energy-related effects from those affecting the AEC; and third, many of the conventional radiographic systems used with DR were designed to compensate for screen/film energy dependencies, and may not be capable of providing constant response for DR.

Many radiographic systems in use today incorporate AEC systems designed for use with screen/film systems and may allow for energy compensation appropriate for screen/film. Such compensation may be hard-wired and unalterable, or may have insufficient ability to compensate appropriately for DR. In particular, it is often the case that K_{IND}'s tend to be higher for AEC-based exposures at lower kV_P's, because the AEC compensation intended for rare-earth screen/film systems overcorrects for lower kV_P's.[20] If this is the case, $K_{TGT}(b,v)$ values for DI may need to be adjusted upward to reflect this energy dependence appropriately.

Appropriate $K_{TGT}(b,v)$ ranges for AEC performance evaluation must therefore take into account the age and pedigree of the radiographic system. Derived $K_{TGT}(b,v)$ limits for AEC testing are equivalent to those that are used for film (for example, ±0.15 optical density units).[21] Certainly, the much narrower latitude of screen/film calls for fairly tight AEC performance limits for reliable clinical results. Although desirable for DR as well, this may not be achievable in practice at this time.

X. Inappropriate Clinical Use of DI

A final note regarding DIs and clinical techniques: even if images being produced clinically have corresponding DIs well with the target range, the clinical techniques used may still not be appropriate. One can just as readily achieve an acceptable DI for an AP L-Spine view with 65 kV_P as with 85 kV_P; evidence of underpenetration and concomitant excess patient exposure with the lower kV_P may be clear from the contrast and underexposure of the spine regions, but may be windowed and leveled out in a digital image. Similarly, poor collimation, unusual patient body habitus, the presence of prosthetic devices, or the presence of gonadal shielding in the image may raise or lower DIs (depending on the exam and projection) and perhaps hide an inappropriate technique.[22,23,24,25] It is essential that all aspects of good clinical technique be adhered to and an appropriate DI value should not be interpreted as proof of good work.

XI. Recommended Optional Features

In addition to implementation of this standardized exposure indicator, there are opportunities for other useful tools to facilitate presentation of image processing-related information and to improve the overall quality of the imaging operation.

For instance, section III calls for an overlay that graphically illustrates the pixels in a given image that have been used to calculate Q_K. This is intended to provide a very quick method of determining that the automated segmentation software performed correctly for any image. A similar feature would be to create a pop-up display of the Q histogram with the locations of the segmentation image pixels min and max overlaid on it showing the minimum and maximum Q values used for Q_K determination. Finally, there are many clever ways to indicate the DI for every image using a sliding bar or color-coded tool with position and/or color linked to the magnitude of DI.

Other highly desirable features are logs of the DI values and logs of the reasons for rejected and repeated images stored on the system. The anatomical view selection and technique factor information for every rejected image as well as the images themselves should also be stored on the system. Software to analyze these logs to assist with process improvement by identifying potential problematic exams, problems with equipment, and technologists in need of continuing education is also invaluable to the user community.

As already mentioned in section III, systems should provide a mechanism to export and import tables in a consistent format so that tables could be shared between imaging facilities using the same DR system. A process for updating the tables of all systems within a facility that is managed via a network would be extremely valuable so that changes in K_{TGT} values can be readily disseminated to distributed systems.

For each clinical mode of operation other than general radiography for which a system is designed, the calibration conditions should be specified by the manufacturer (see section IX). The manufacturer should also provide the following:

- filter(s) to be utilized for establishing K_{STD} for general radiography as well as for other modes of operation and should specify which body parts and views are associated with each filter (see section XII),

- a means to readily mount/dismount the filter(s) on an x-ray collimator,
- mA, time/mAs, source-to-image distance (SID), grid or no grid, and system protocol settings for calibration conditions,
- the grid transmission factor associated with each beam condition (if present),
- exact specification of how to configure the system to obtain the "for-processing" image data,
- written, step-by-step calibration protocol and recommended frequency of calibration,
- full disclosure of the K_{TGT} values associated with each body part and view,
- specification of the accuracy and reproducibility of the K_{STD} for the standard beam condition and the expected change in K_{IND} for other beam conditions.

The task group strongly recommends implementation of all of these ideas and anticipates the creation of many more once the efforts of the equipment manufacturing community are brought to bear on these issues.

XII. Application to Dedicated Chest, Mammography, Veterinary, and Dental Radiography

K_{IND} is intended to be used as a measure of image quality with respect to image noise. For low-energy x-rays, more incident radiation is required to create the same detector response as for high-energy x-rays. The variation in detector response for kV_P values between 55 and 90 is sufficiently small to make K_{IND} an effective indicator of image quality with respect to the recorded noise in the image.[4] For higher energies, this may not be the case. To maintain a consistent relationship between image noise and the indicator, it is possible that two standard beam conditions could be defined: one for imaging of the chest at tube potential settings above 100 kV_P and one for all other radiographic images. This higher energy standard beam should be reasonably close to RQA9 (see Table 3).

For a tube with HVL of 5.00 at 120 kV_P (RQR9), similar beam quality with HVL = 11.6 mm is obtained with 40 mm of pure aluminum as specified for RQA9 or with 1.0 mm Cu plus 4 mm Al (type 1100). If 11.6 mm Al HVL cannot be achieved at 120 kV_P with the recommended filtration, the additional aluminum filtration may be reduced and the kV_P adjusted to achieve the required HVL.

Digital mammography, veterinary, and dental radiography can all potentially benefit from a universal exposure indicator for the same reasons discussed in this report. DR in these fields suffers the same problems with manufacturer-specific exposure indices from which DR suffers. Application to these areas would require modification of the calibration beam conditions to reflect the differences in typical beam attenuation and beam energies in clinical use.

Table 3. A Standard Beam Condition for Dedicated Chest Imaging Systems

Application	*kV_P*	*Added Filtration*	*Target HVL*	*IEC Surrogate*
Dedicated Chest	114–126	1.0 mm Cu + (0–4.0) mm Al* or 40 mm pure Al	11.6 ± 0.3 mm Al*	RQA9

* Type 1100

The conditions for general radiographic systems differ substantially from those for mammography systems. Developing a universal exposure indicator for mammography would be useful for providing technologists feedback about exposure adequacy, especially for institutions with digital mammography units from different manufacturers.

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Appendix A

RQA5 vs. TG-116 Standard Beam Conditions, XSPECT 3.5b Computation Simulation

M. Flynn, 15 July 2007

Simulation of Standard Beam Conditions

Computational simulations using the XSPECT toolkit were used to simulate:

- the x-ray spectrum emitted by typical x-ray tube and collimator assemblies
- the attenuation of the spectrum by various amounts of added filtration
- the half-value of the x-ray spectrum
  - o incident on the added filtration
  - o incident on a detector
- the signal and noise of a digital radiography detector.

X-ray Source Conditions

Three x-ray sources were considered based on their HVL at 70 kV_P:

A. IEC RQR5 - 2.58 mm Al, 70 kV_P, Al intrinsic filtration
B. Henry Ford Health System (HFHS) Rm 3 - 3.02 mm Al, 70 kV_P, Al/Cu intrinsic filtration
C. HVL 4 - 4.00 mm Al, 70 kV_P, Al/Cu intrinsic filtration

Source A is a lightly filtered source based on the IEC RQR5 specifications. Source B is based on a clinical system at HFHS for which the HVL has been experimentally measured. Source C is a hypothetical system with relatively heavy intrinsic filtration.

Added Filtration

Three added filtration configurations were considered:

A. 21 mm aluminum (99.9%, RQA5).
B. 0.5 mm copper and 2.8 mm aluminum (99.0% Al).
C. 24.5 cm muscle (National Bureau of Standards (NBS) muscle composition).[A1]

Added filtration A is based on the RQA5 standard and filtration B on the TG-116 standard. For filtration B, the aluminum thickness is specified as 2.8 mm, based on results indicating that the HVL matches at the same kV_P. The added aluminum was iteratively changed to determine this value. Additionally, the relation between added aluminum and the kV_P needed to obtain a HVL of 6.8 was determined. Similarly, 24.5 cm of muscle is chosen to match the HVL at the same kV_P.

The results are summarized as

- The three added filtration conditions produce nearly identical spectra at the same kV_P when the kV_P is adjusted to obtain a HVL of 6.8 (see Figure A1)
- The HVL increases/decreases by about 0.1 for a 1 kV_P change (see Figure A2)
- The kV_P for which a HVL of 6.8 is produced is related to the HVL of the source with a slope of –2.5 kV/mm (73.6 kV_P at 2.58, 72.6 at 3.02, 70.1 at 4.00 mm)
- The detected signal and SNR in relation to kV_P is nearly identical for added filtrations A and B above (see Figure A3).

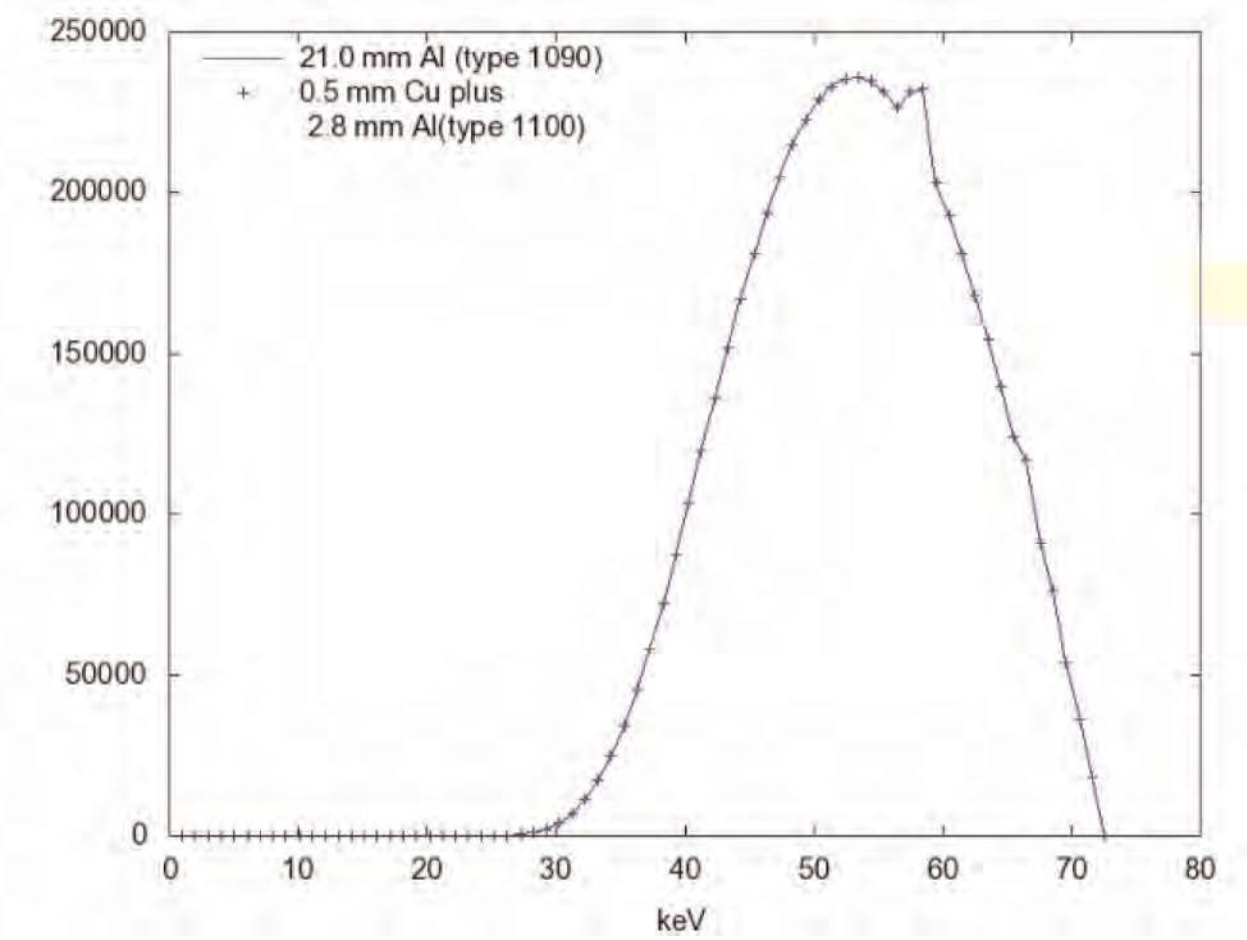

Figure A1a. Using a source with HVL=3.02 mm Al, the spectra for alternative additive filtrations are shown. Both spectra are for kV=72.59 and both have HVL=6.80. The points have been adjusted as indicated to produce the same exposure (mR) as the line. The spectral units are in #/mAs-cm^2-keV. The spectra for 21 mm added aluminum (type 1090, 99.9%) is shown as a solid curve.

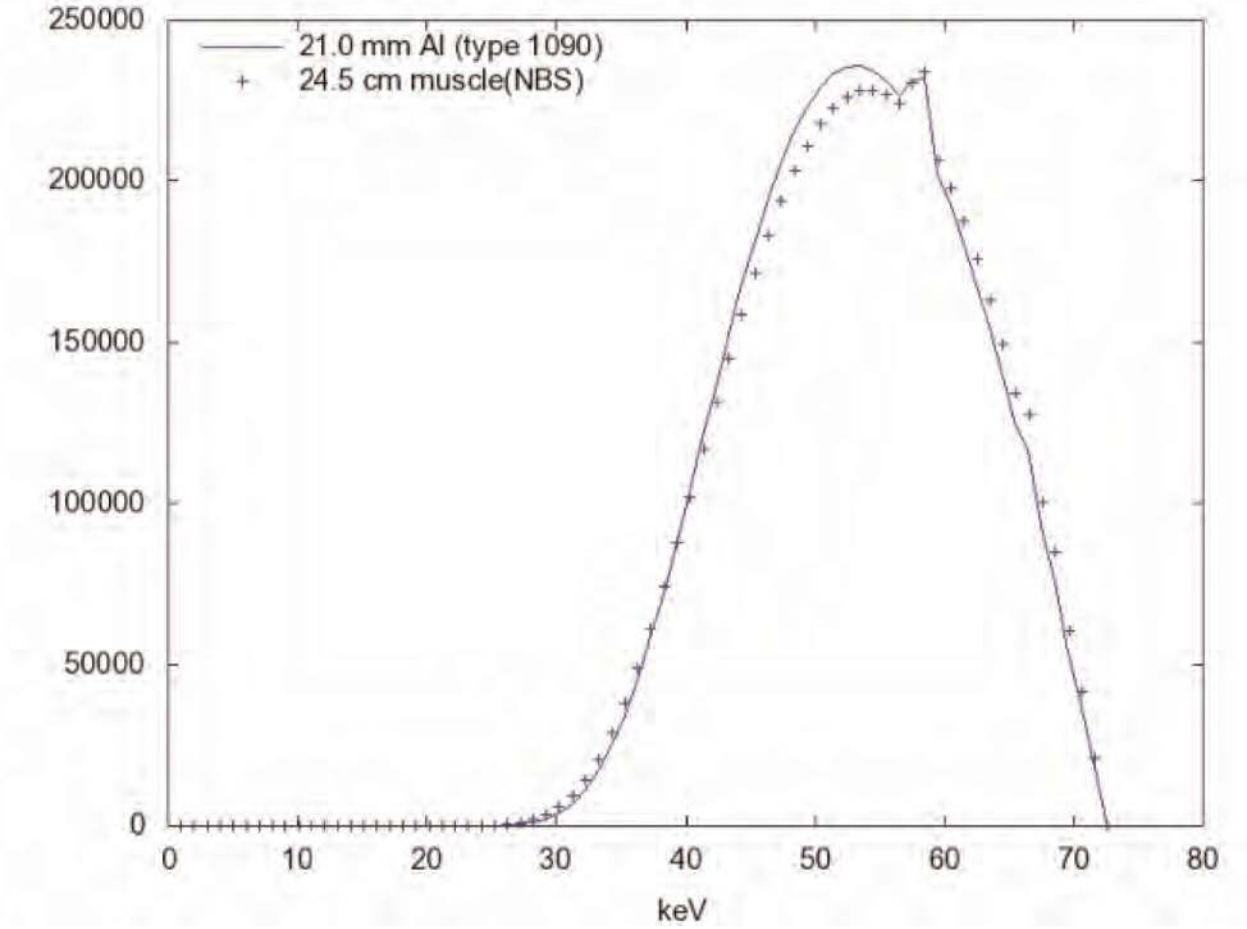

Figure A1b. The spectrum with added filtration of 24.5 cm of muscle (NBS) is shown as points. These values have been increased by a factor of 43.7 to normalize the exposure.

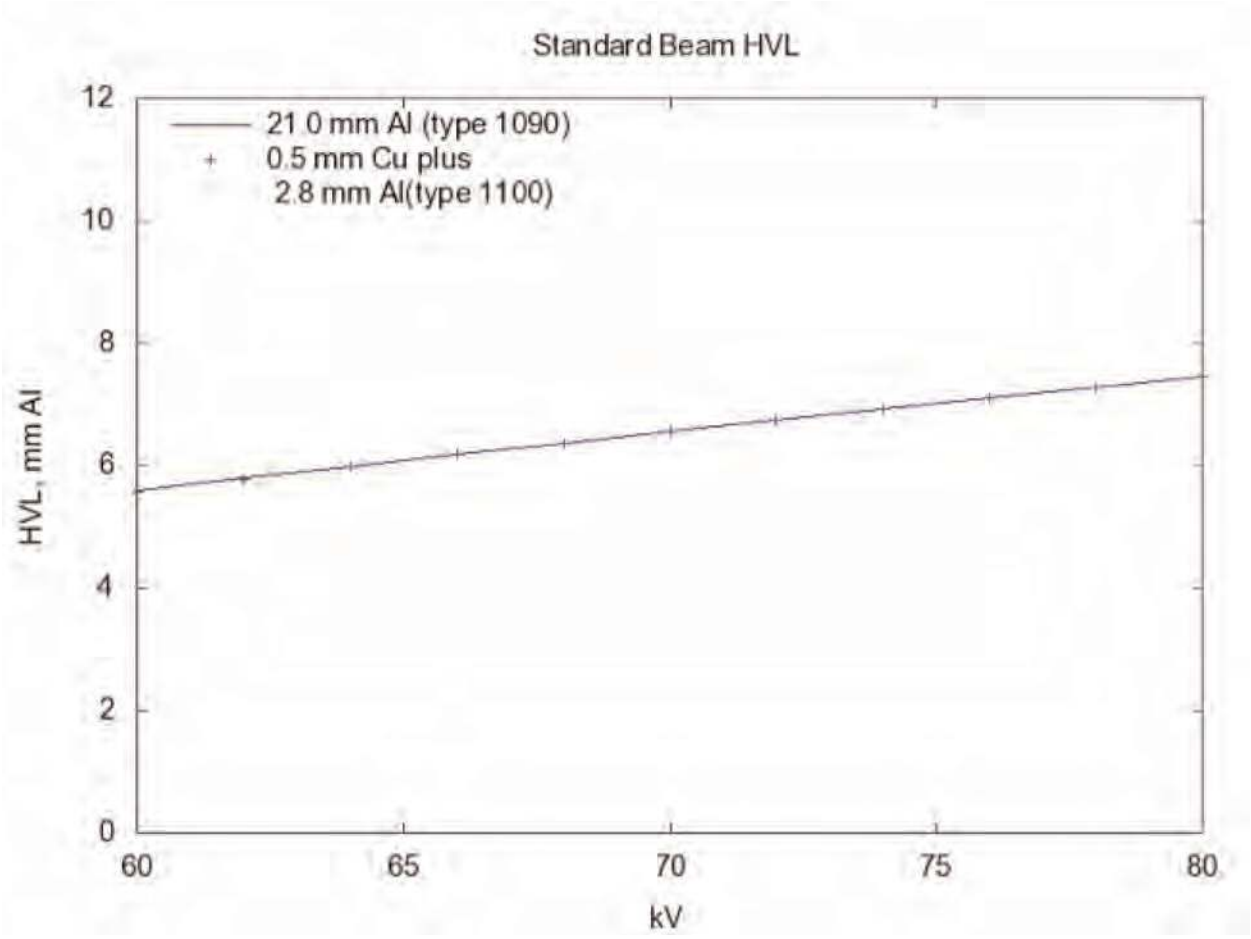

Figure A2a. The points are for 0.5 mm Cu plus 2.8 mm Al (type 1100) added filtration. The HVL of both is 6.8 at a kV of 72.59. For both, the slope of the relation with kV is 0.094 mm/kV. As a rule of thumb, a 1 kV increase will produce a 0.1 increase in HVL.

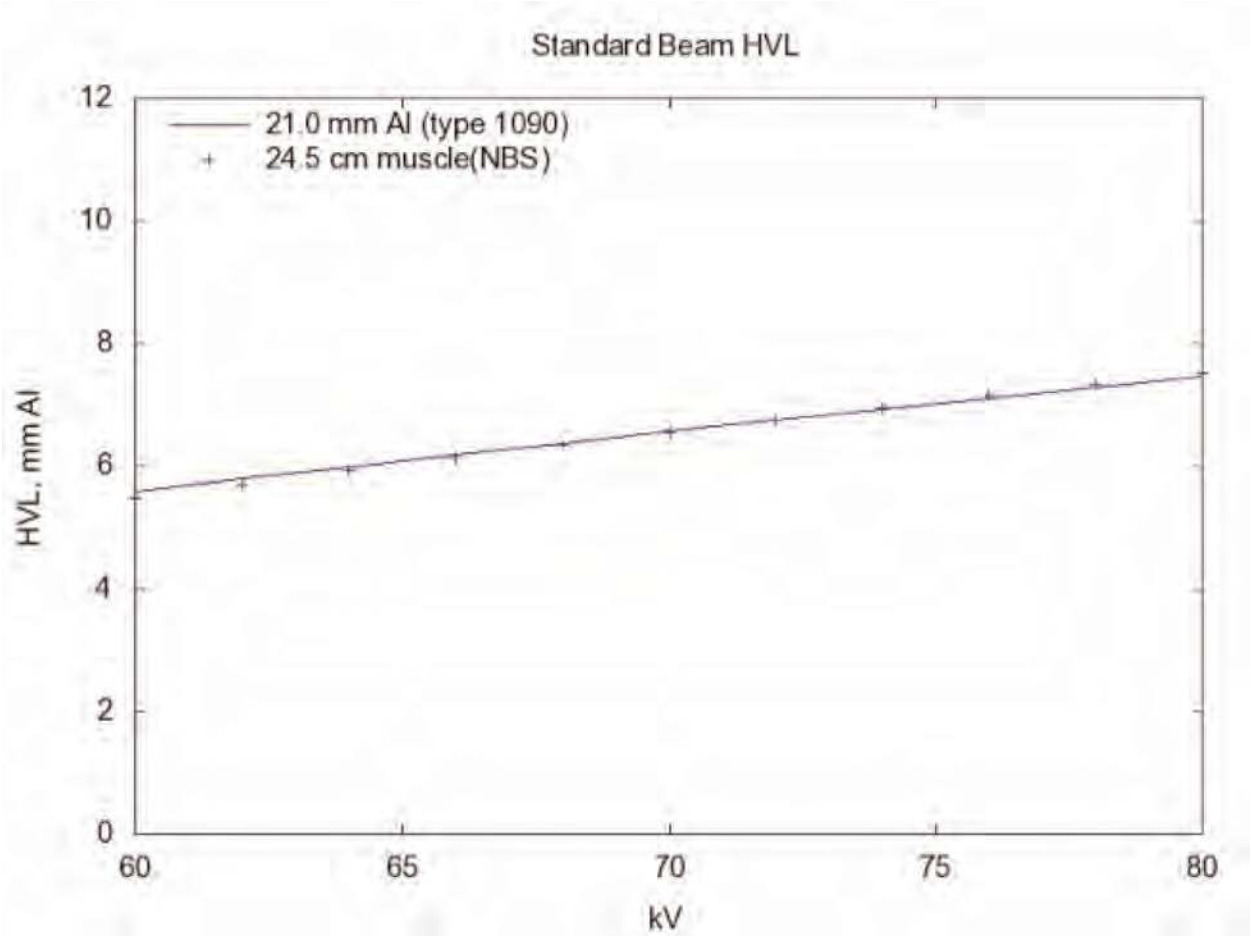

Figure A2b. The points are for 24.5 cm muscle (NBS) added filtration. The sensitivity of HVL in relation to kV is slightly more compared with that for 21 mm Al added filtration.

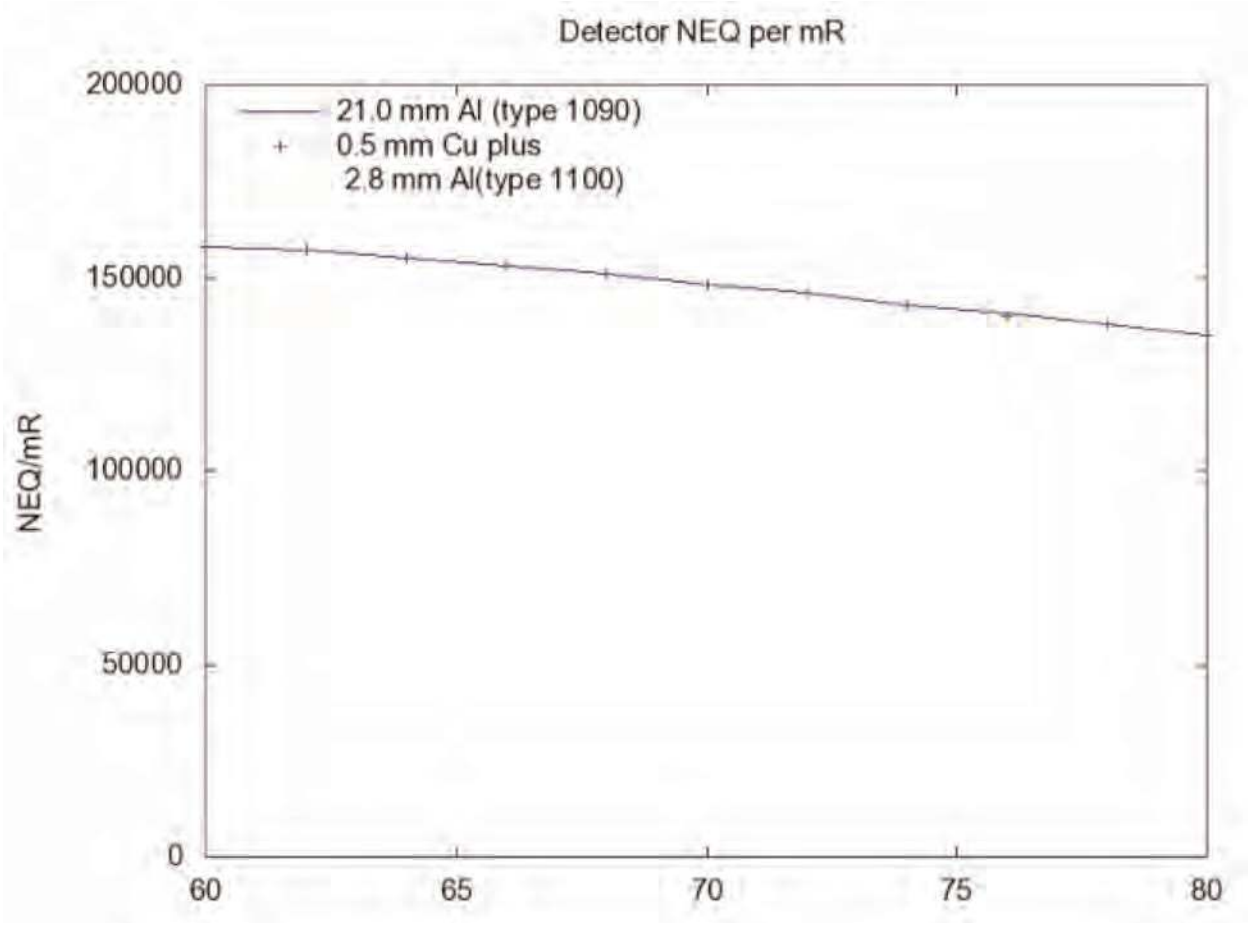

Figure A3a. Using a source with HVL=3.02 mm Al, the detected NEQ (i.e., SNR2) for alternative additive filtrations are shown in relation to kV. The response for 21 mm added aluminum (type 1090, 99.9%) is shown as a solid curve. The points are for 0.5 mm Cu plus 2.8 mm Al (1100) added filtration. The two conditions produce nearly identical response. As expected, there is a slow decrease as kV is increased.

XSPECT Computational Tools

Half-Value Layer (HVL, aluminum) was computed using the XSPECT toolkit for modeling x-ray transmission imaging. This toolkit contains programs for generating and modifying x-ray spectra stored in a common format. Utility programs modify the spectra to account for primary beam attenuation, unit conversion, and distance effects. Other programs compute the exposure in mR, the HVL of the x-ray spectrum, and the response of a detector in terms of the recorded signal and noise in units of electrons. The toolkit is presently being used in a graduate laboratory course on radiation imaging (NERS 580, University of Michigan, http://www.engin.umich.edu/class/ners580).

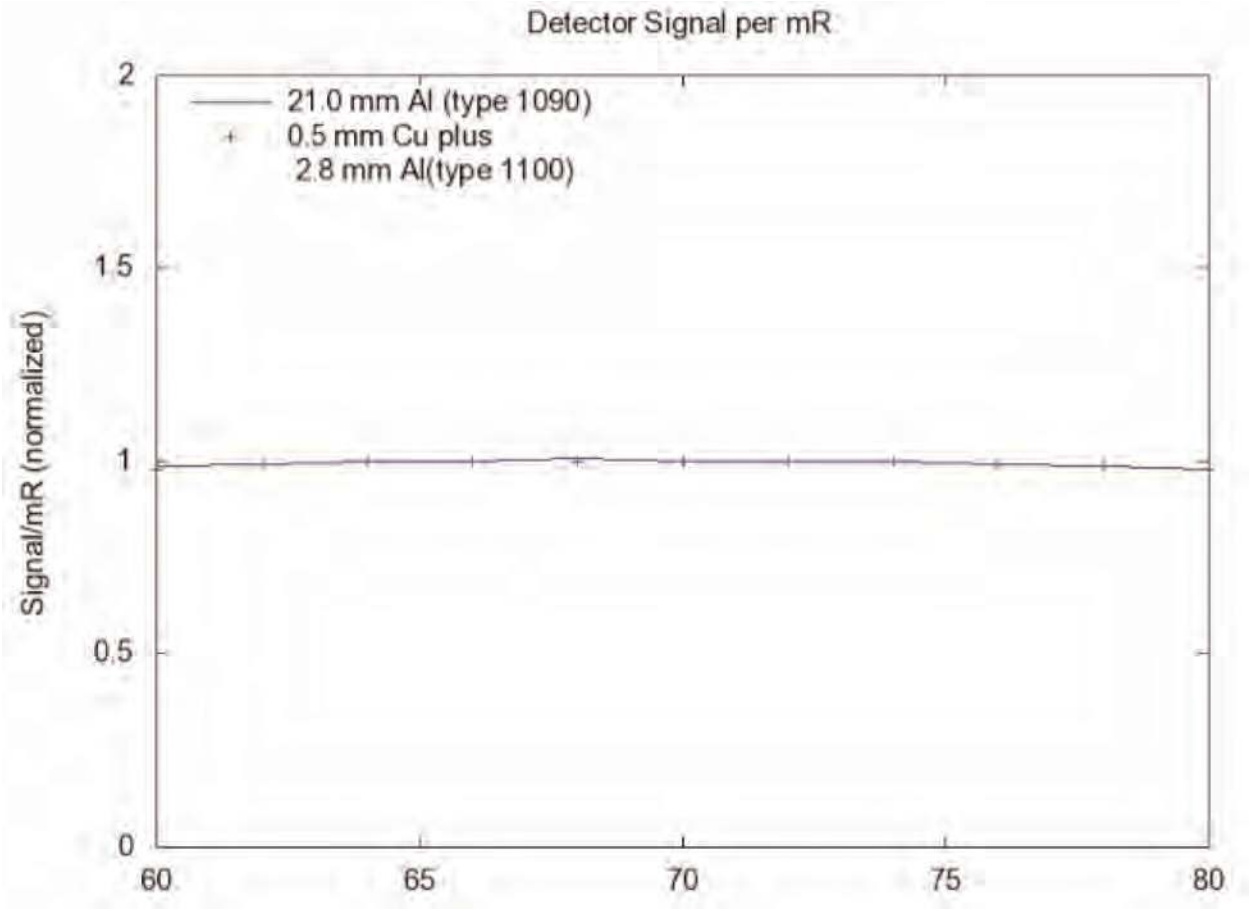

Figure A3b. With similar conditions as for Figure A3a, the detected signal (electrons) per incident mR (scatter free) is seen to vary slowly with kV.

X-Ray Emission Spectrum (SPECT_GEN)

X-ray emission spectra in units of photons/(keV-sr-mA-s) are obtained from the SPECT_GEN module. In generating a spectrum, the user defines the kV_P, atomic number, tube angle, voltage waveform, and spectral increment. The program then generates a spectrum of bremsstrahlung and characteristic radiations and corrects them for self-absorption. The spectral calculations are based on equations from Storm[A2,A3] and contain empirical corrections for tube angle and voltage ripple. The results from XSPECT have been crosschecked for molybdenum and tungsten tubes against Fewell's data published by the U.S. Food and Drug Administration (FDA) through the Bureau of Radiological Health.[A4].

Primary Beam Attenuation (ATTEN)

Attenuation of the emission spectrum by various layers of materials is done using the program ATTEN that applies Beer's law to the primary beam based on the material composition of the layer. Material compositions are stored in material files having a standardized format and maintained in a material library. Photoelectric cross sections for the elements were obtained from the work of Biggs.[A5] The coherent and incoherent cross sections were taken from the work of McMaster.[A6] For aluminum materials, type 1100 was used for 99.0% purity material and type 1090 for 99.9% purity material. Typical impurities were taken from matweb.[A7]

Entrance Exposure (mR)

After the beam is attenuated, a simple inverse square law is applied to yield the spectrum at the entrance point to the patient. The mR program then computes the exposure at this point. The computation of exposure involved summing the product of energy fluence (ergs/cm^2/keV) and the air energy absorption coefficient over the range of spectral energies.[A6] A conversion factor of 87.643 (ergs/gm)/roentgen is used along with the air ionization factor of 33.97 joules/coulomb.

Detector Response (DETECT)

The DETECT routine is used to compute detector signal by a weighted numeric integral of the x-ray energy spectrum incident on the detector assembly, $\phi_d(E)$ (photons/{keV-cm^2-mA-s}):

$$S = A_p \int_0^{kV_P} \eta_S(E) E \phi_d(E) dE \qquad \sigma_S^2 = A_p \int_0^{kV_P} (\eta_\sigma(E))^2 E^2 \phi_d(E) dE,$$

where A_p is the pixel area in cm^2 and S has units of electrons/(pixel-mA-s). The signal transfer efficiency, $\eta_s(E)$ (electrons/keV), represents the average number of electrons produced in the detector by an incident x-ray with energy E. If $\eta_s(E)$ equals 1.0 for all energies, the signal computed is that for an ideal energy integrating detector, S_i.

The noise variance of the detector signal was computed using a similar integration (see equation above). The noise variance, σ_S^2 , has units of electrons^2/(pixel-mA-s). The noise transfer efficiency, $(\eta_s(E))^2$ (electrons^2/keV)^2, represents the average contribution to the signal variance of a pixel by an incident x-ray with energy E. The noise computed for an ideal detector is similarly obtained by setting $\eta_s(E)$=1. The signal-to-noise ratio, S/σ_S , is seen to be proportional to $A_p^{1/2}$ as expected.

The accuracy of the signal and noise estimate computed by the DETECT routine depends on an accurate knowledge of the transfer efficiencies $\eta_s(E)$ and σ_S^2 . For this work, both were determined by using a Monte Carlo analysis to estimate the detector's signal probability distribution function, $p(q,E)dq$, that describes the probability of collecting q electrons when an x-ray of energy E is incident on the detector. The detector tables for a 500-micron thick selenium direct DR detector were used in this comparison.

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A4. Fewell, T.R., and R.E. Shuping. Handbook of Mammographic X-ray Spectra. HEW publication (FDA) 79-8071, 1978 (Revised data tables obtained from BRH).

A5. Biggs, F., and R. Lighthill. "Analytical Approximations for X-Ray Cross Sections." Part II, SC-RR-710507. Sandia Laboratories, Albuquerque, New Mexico, 1971.

A6. McMaster, W.H., N.K. Del Grande, J.H. Mallett, and J.H. Hubbell. "Compilation of X-Ray Cross Sections." Sec. 2, Rev. 1, University of California, Livermore, U. S. Atomic Energy Commission Report UCRL-50174, 1969.

A7. www.matweb.com accessed 25 June 2008.

Appendix B

Comparison of Pure Aluminum vs. Commercially Available Types 1100 and 1190 Aluminum and a Copper/Aluminum Alternative for RQA5

David L. Leong, Wei Zhao, Patrick C. Brennan
November 25, 2007

Introduction

Task Group 116 has recommended 0.5 mm copper (Cu) with 0 to 4 mm of Alloy 1100 aluminum (Al) as an alternative x-ray beam hardener to obtain the same RQA5 spectrum as specified in IEC 61267:2005.[B1] IEC 61267 requires the use of 99.9% pure aluminum for the RQA5 radiation qualities.

Alloy 1190 is a 99.9% pure Al alloy that meets the requirements of IEC 61267. In attempts to purchase Alloy 1190, the authors were unsuccessful in finding an off-the-shelf source; alternatively Alloy 1100 is a 99.0% pure Al alloy that is widely available on the market. Alloy 1190 is registered with 99.9% purity, which is higher than what is considered the highest purity commercial grade Al with 99.45%.[B2] Alloy 1190 falls into the category of scientific grade (also called ultra pure aluminum) and is available only through specialty metals companies for a high price, in small quantities and limited form.

What is currently unknown is the impact of using the widely available Alloy 1100 compared with the specified Alloy 1190.

Materials and Methods

Since aluminum composition will vary from batch to batch and from the source of the raw materials, the first goal using simulation is to determine mixture compositions that reflect the maximum, minimum, and median attenuation possible for Alloys 1100 and 1190. Since it is not possible to simulate all the possible combinations of elements in a specific batch, mixture modeling theory was used to select a reasonable set of mixtures as input to the simulation model to produce a prediction formula.

To perform the mixture modeling, the custom design platform in JMP 6 software (SAS Institute, Cary, NC) was used. A total of 1395 mixtures for each alloy using 30 elements were made in order to develop an accurate prediction formula.

Once the mixtures were selected in JMP, radiation quality simulation was performed using IDL 6.4 (IIT Visual Solution, Boulder, CO). The initial beam spectrum was calculated using the method developed by Boone and Seibert,[B3] and to this beam different filters were applied depending on the results desired. As the filters were applied, the attenuation was calculated with 1 keV interval from 1 to 70 keV.

The mass attenuation coefficients used in the simulation were taken from Cullen et al.[B4], with the exception of vanadium which was from the National Institute of Standards and Technology (NIST)[B5] mass attenuation coefficients dataset. For a composite material like Alloy 1100, the linear attenuation coefficient μ was calculated according to the weight fraction of each element.

Table B1. Simulated HVL Results from the Various Alloy Batches

	Pure Al	*Alloy 1190*	*Alloy 1100*	*0.5Cu/1Al*	*0.5Cu/2Al*	*0.5Cu/3Al*
Mean	6.7638	6.7885	6.8803	6.6567	6.7539	6.8460
Std Dev	--	0.0006293	0.0141664	0.0007604	0.0014424	0.0020562
Mean-Target	–0.0362	–0.0115	0.0803	–0.1433	–0.0461	0.046

To determine the half-value layer (HVL) for each Al Alloy mixture model, the initial x-ray spectrum was hardened to RQR5 with pure Al to achieve a HVL of 2.58 mm Al. Then the composite filter was added and a new beam spectrum calculated. Last, the HVL was determined for the new beam using 100% pure Al. For the calculation of Cu/Al filters, pure copper was used first and then the 1100 Alloy.

Once simulated values were determined for each mixture, JMP 7 software (SAS Institute, Cary, NC) was used to develop a HVL prediction formula. Next, the HVL for 5000 simulated batches of each Alloy was generated using the HVL prediction formula.

Results and Discussion

The numeric details of the distribution of the 5000 simulated samples of both Alloys 1100 and 1190 are listed in Table B1. It is important to note that the standard deviation of Alloy 1100 is 22.5 times greater than that of Alloy 1190, but the ±3 standard deviation is still only ±0.0425 mm Al. The last line in the table shows the deviation of the mean from the RQA5 target of 6.8 mm Al.

From the simulation results, we see that the HVL using pure Al is slightly lower than the target of 6.8, but the rounding to one decimal point is 6.8. Out of all alloys tested, 1190 provides a HVL that is the closest to the target with a mean difference of just 0.01 mm Al. The standard deviation is negligible, so it can be expected that any batch of Alloy 1190 would produce a HVL value rounded to 6.79 mm Al. HVL for Alloy 1100 had a noticeable shift from the target of 6.8, which may be attributed to the addition of Cu and iron (Fe) to the alloy. With a range of 6.8803±0.0283 mm Al covering 95% of all batches, we have a range of 6.8525 to 6.9081 mm Al. The 0.1 mm Al error margin is acceptable for clinical use and well within the ±0.25 mm Al proposed by TG-116.

Figure B1 shows the comparison between the mean beam spectrum for Alloy 1100 and that of pure Al. Alloy 1100 absorbs slightly less radiation, and the beam quality is shifted slightly harder. This can also be seen in Table B1 with the mean having a difference of 0.1165 mm Al. A normalized plot of the two spectra makes it easier to see the difference in beam hardness.

Figure B2 compares the spectrum for 21 mm pure Al to that for Cu/Al combination—there is a large difference in the height of the beam spectra. Using the 0.5 mm Cu and 2 mm Al filter combination, there are over two times more photons. Figures B3a and B3b show the spectra for different Cu/Al filters normalized to the output of pure Al. It can be seen that there is a left shift with 2 mm Al, and with 3 mm Al, the spectra are nearly identical. The HVL values in Table B1 show that 2.5 mm Al with 0.5 mm Cu would have essentially the same HVL as 21 mm pure Al.

In addition to the investigation by the authors, several TG-116 members performed testing of x-ray systems at their institutions to determine if the target HVL of 6.8 mm Al can be achieved using 0.5 mm Cu filter in combination with an Al filter ranging in thickness from 0 to 4 mm. A total of 25 systems

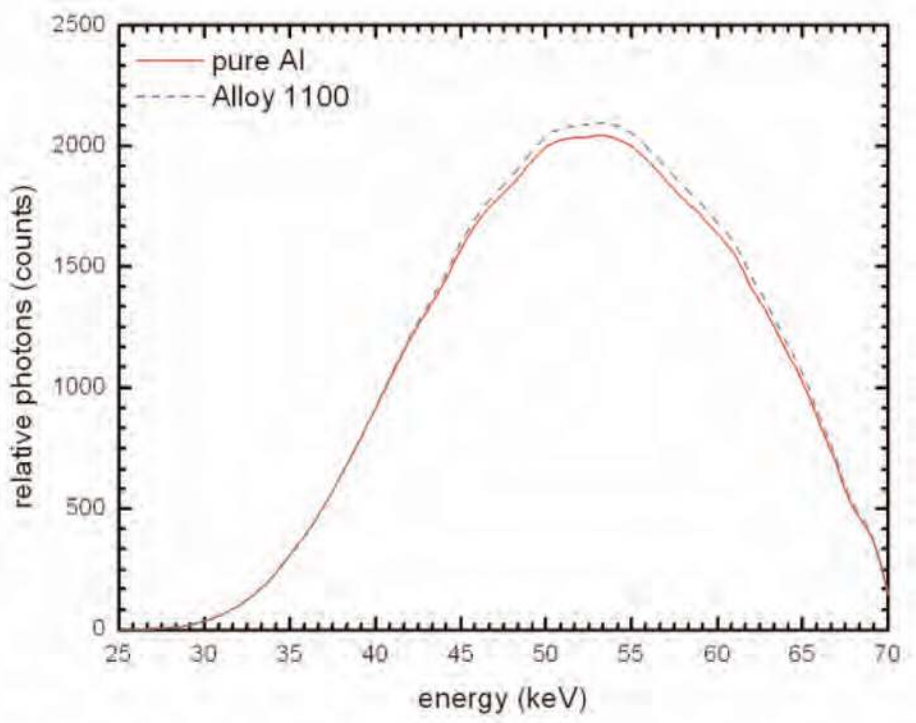

Figure B1a. Simulated beam spectrum comparison of 21 mm 100% Al and the mean of 21 mm Alloy 1100 at 70 kV_P.

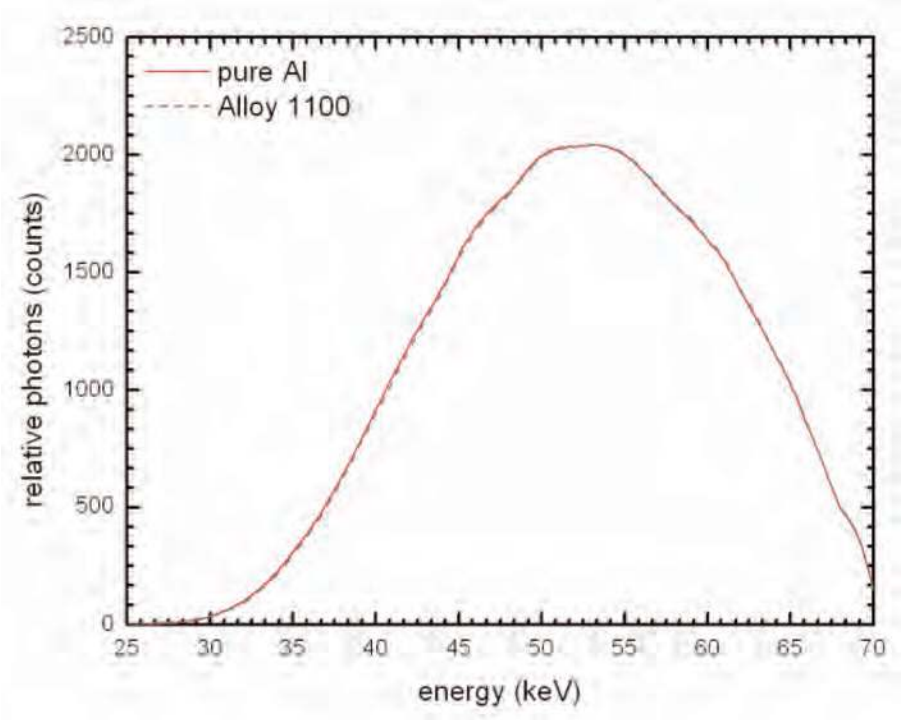

Figure B1b. Simulated beam spectrum comparison of 21 mm 100% Al and the mean of 21 mm Alloy 1100 at 70 kV_P normalized to the 100% Al spectrum to allow for better visualization of the differences.

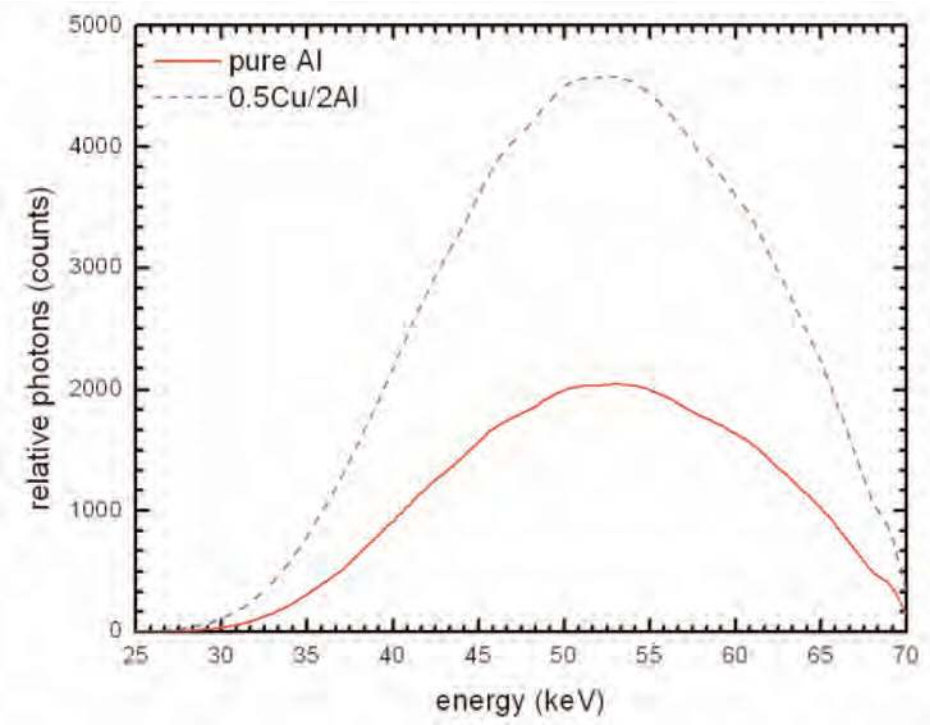

Figure B2. Simulated beam spectrum comparison of 0.5 mm Cu & 2.0 mm Alloy 1100 and 21 mm 100% Al at 70 kV_P.

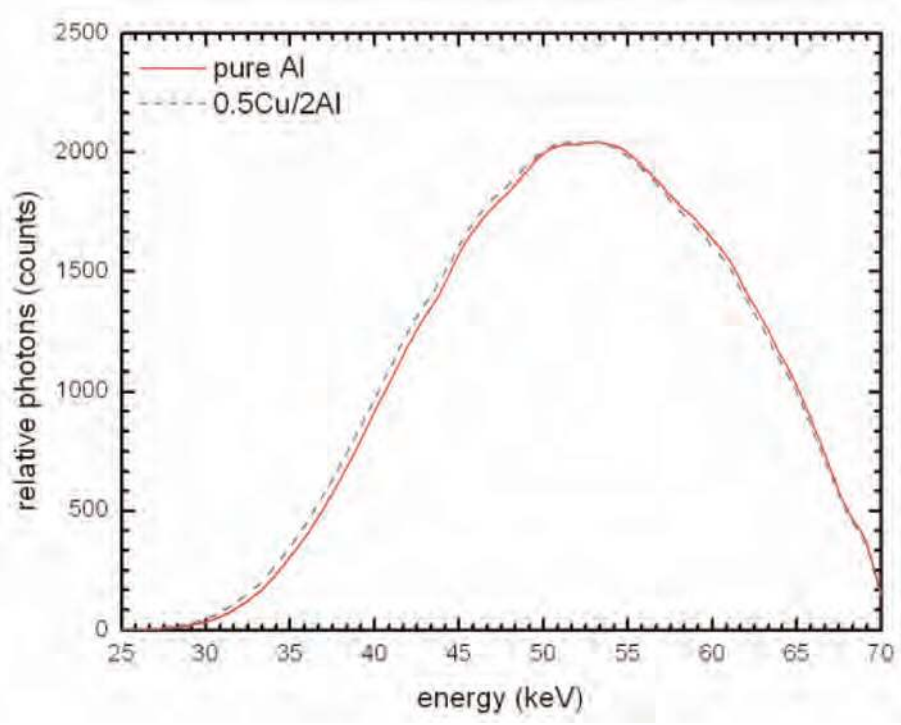

Figure B3a. Simulated beam spectrum comparison of 0.5 mm Cu & 2.0 mm Alloy 1100 and 21 mm 100% Al at 70 kV_P normalized to the 100% Al spectrum to allow for better visualization of the differences.

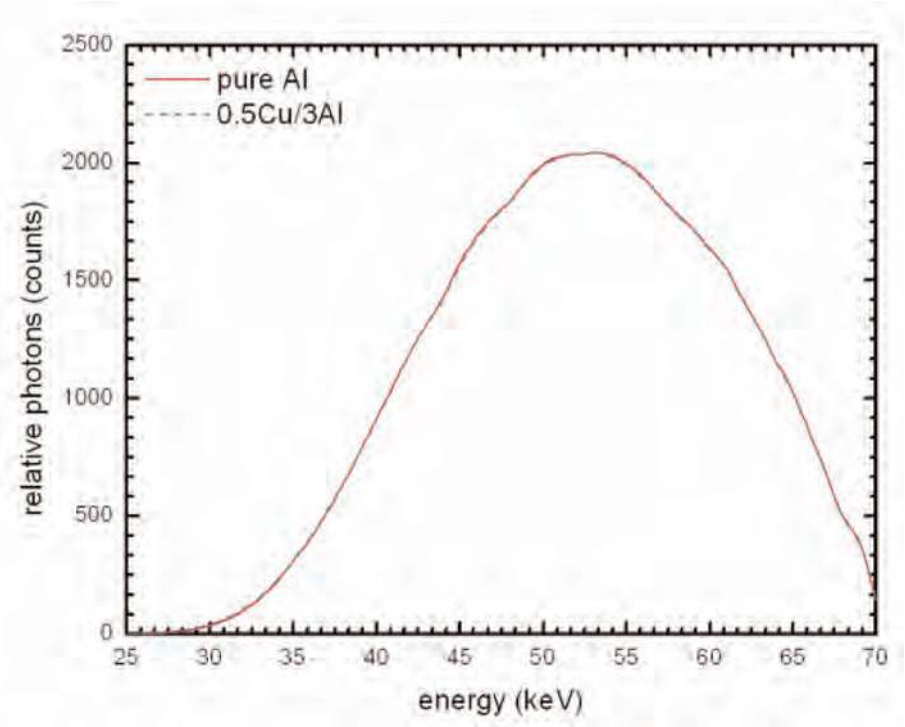

Figure B3b. Simulated beam spectrum comparison of 0.5 mm Cu & 3.0 mm Alloy 1100 and 21 mm 100% Al at 70 kV_P normalized to the 100% Al spectrum to allow for better visualization of the differences.

were tested. The initial filtration was measured for all 25 systems and the results are plotted in Figure B4.[B6] It is important to note that out of the 25 systems only 6 had initial HVLs of less than the maximum of 2.58 mm Al specified in IEC 61267.

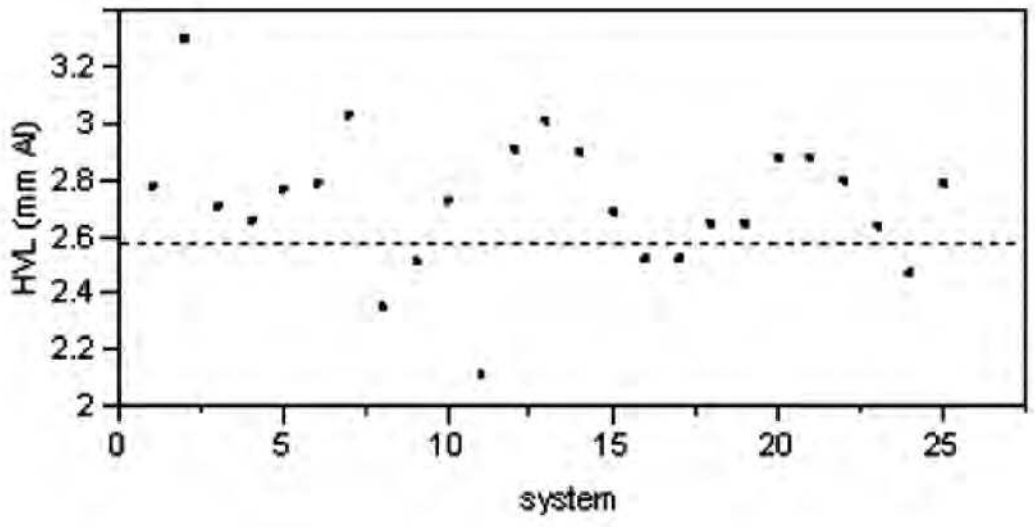

Figure B4. Initial x-ray system HVL at 70 kV_P.

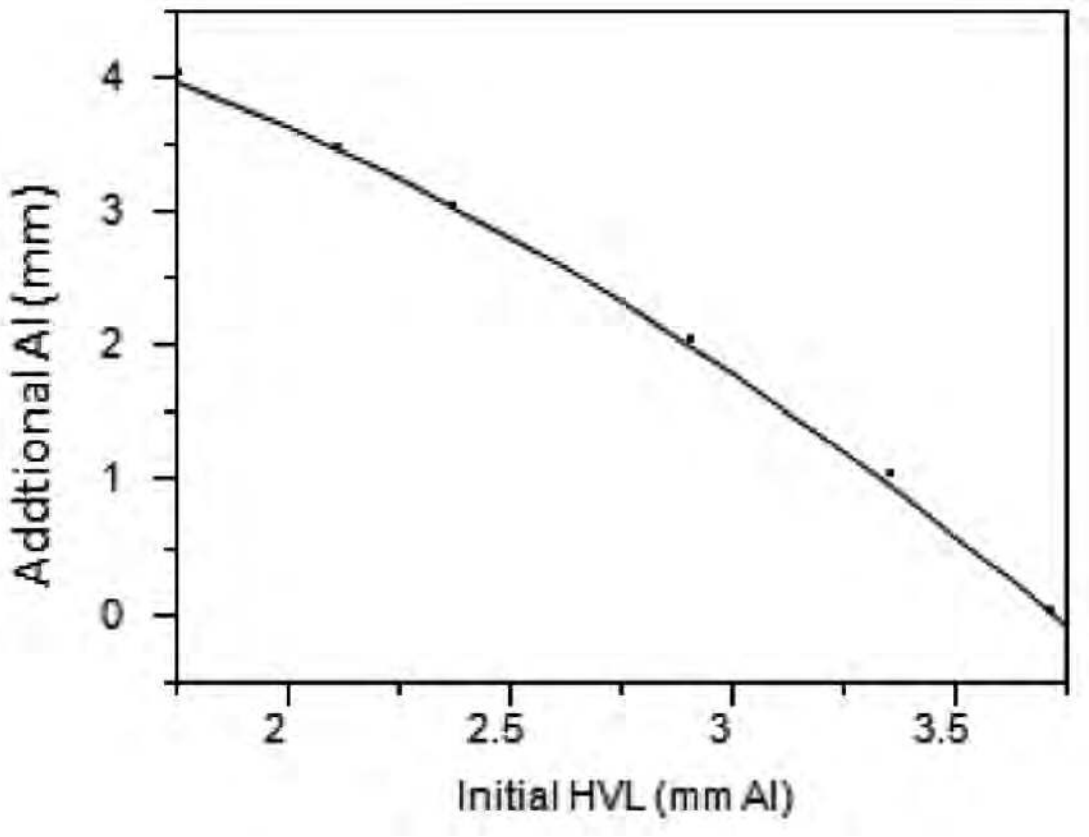

Figure B5. Approximate Al to add to 0.5 mm Cu for RQA5 at 70 kV_p.

Since it is usually difficult to remove filtration in a clinical setting, it is much more practical to adjust the added filtration to reach the target HVL of 6.8 mm Al. The AAPM recommendation of varying the amount of Al in the Cu/Al filter combination is an appropriate method. Figure B5 plots the amount of additional Al needed to achieve the target HVL of 6.8 mm for different initial HVL values.

Conclusion

We conclude that the IEC 61267 requirement to use 99.9% pure Al is unrealistic and is unnecessary. While the use of 21 mm of Alloy 1100 Al is a close substitute, a slightly thinner filter would be appropriate to achieve the target HVL of 6.8 mm Al.

The use of a Cu/Al combination filter is an adequate alternative. Our simulation results show that 0.5 mm Cu and 3.0 mm Al provide the same HVL as 21 mm of pure Al. A side benefit of using Cu/Al combination is that it requires half of the exposure to achieve the same exposure as a 21 mm pure Al filter.

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Appendix C

Current Status of Exposure Indices

A variety of exposure indicators have been provided by manufacturers of DR systems. Some of these are summarized in Table C1, which illustrates the wide variation in terms, units, mathematical forms, and calibration conditions of these exposure indicators. Inconsistency among manufacturers is the primary drawback for clinical use of exposure indices at present. Inconsistency creates confusion for practitioners who work with systems from more than one vendor, or those who have been trained on one system, but practice using another.

Appendices D–O present detailed descriptions of the exposure indices provided by some of the digital radiography vendors.

The use of exposure indicators began with the cassette-based computed radiography (CR) systems. Because of the extremely wide dynamic range of the CR detectors and the relatively narrow dynamic range of exposures in the radiographic projection, the first exposure indicators were developed to estimate the exposure to the detector in order to modify the gain for harvesting the latent image. Later cassette-based systems employed the same sort of estimates to re-scale the digitized data to increase contrast and to compensate for variations in exposure factor. Although not originally intended by the manufacturers to be used for quality control purposes, practitioners soon recognized that the exposure indicator was a useful means to evaluate the adequacy of radiation exposure to the detector and, thus indirectly, the appropriateness of selected technique factors. Not only was this useful to the technologists when setting technique factors, but, from a more global perspective, it allowed hospitals and clinics to analyze overall exposure trends.[C1] QC programs based on exposure indicator monitoring have been shown to moderate exposure in actual clinical practice.[C2] This practice has matured to the point where some manufacturers now offer automated tools to log and report exposure indicator statistical information for the purposes of QC analysis.

Exposure indicators for cassette-based CR systems from several manufacturers are presented in appendices D through G, K, and M.

Fujifilm's so-called "Sensitivity" or "S-number" is the oldest exposure indicator. It closely mimics the concept of "speed class" that is familiar to technologists. That is, when operated in Automatic or Semi-automatic Exposure Data Recognizer (EDR) mode, the index value increases with a decrease in exposure to the detector and vice versa. In an absolute sense, the numerical value of the indicator does not correspond exactly with the ISO 9236-1 definition of speed, so there is some confusion with the nomenclature.[C3] Accurate interpretation of the S-number is limited without knowledge of the value of "Latitude" or "L-number" for the particular image.[C4] Approximately 2.5 times as much exposure is required to produce the same S-number on a high resolution (HR) cassette as with a standard resolution (ST) cassette. The QC value of this indicator is compromised in the vendor's most recent software in that the user can retrospectively modify the S-number value. This feature creates uncertainty in the validity of the S-number in representing exposure trends.

Kodak CR uses an exposure indicator known as the "Exposure Index," or "EI," which represents the average pixel value of the clinical region of interest. Because of the characteristic function of the digitized image, a change of 300 in the value of EI indicates a change of a factor of 2 in exposure to the

Table C1. DR Exposure Indicators, Units, and Calibration Conditions (Adapted from reference [C5])

Manufacturer	Indicator Name	Symbol	Units	Exposure Dependence	Calibration Conditions
Fujifilm	S Value	S	Unitless	200/S X (mR)	1 mR at 80 kV_P 3mm Al (Total) => S=200[a]
Kodak	Exposure Index	EI	mbels	EI + 300 = 2X	1 mR at 80 kV_P + 1.0 mm Al and 0.5 mm Cu => EI=2000
Agfa	Log of Median of histogram	lgM	bels	lgM + 0.3 = 2X	2.5 μGy at 75 kV_P + 1.5 mm Cu =>lgM=1.96 at 400 Speed Class
Konica	Sensitivity Number	S value	Unitless	for QR=k, 200/S α X (mR)	for QR=200, 1 mR at 80 kV_P => 200
Canon	Reached Exposure Value	REX	Unitless	for Brightness=c_1, Contrast=c_2, REX α X (mR)[1]	for Brightness = 16, Contrast = 10, 1 mR ≈ 106[1]
GE	Uncompensated Detector Exposure	UDExp	μGy Air KERMA	UDExp α X (μGy)	80 kV_P, standard filtration, no grid
GE	Compensated Detector Exposure	CDExp	μGy Air KERMA	CDExp α X (μGy)	kV_P, grid, and additional filter compensation
GE	Detector Exposure Index	DEI	Unitless	DEI ≈ 2.4X (mR)[a]	Not available
Swissray	Dose Indicator	DI	Unitless	Not available	Not available
Imaging Dynamics Company	Accutech	f#	Unitless	$2^{f\#}$=X(mR)/X_{tgt}(mR)	80 kV_P + 1 mm Cu
Philips	Exposure Index	EI	Unitless	100/S α X (mR)	RQA5, 70 kV, +21 mm Al, HVL=7.1 mm Al
Siemens Medical Systems	Exposure Index	EXI	μGy Air KERMA	X(μGy)=EI/100	RQA5, 70 kV +0.6 mm Cu, HVL=6.8 mm Al
Alara CR	Exposure Indicator Value	EIV	mbels	EIV + 300 = 2X	1 mR at RQA5, 70 kV, +21 mm Al, HVL=7.1 mm Al => EIV=2000
iCRco	Exposure Index	none	Unitless	Exposure Index log [X (mR)]	1 mR at 80 kV_P + 1.5 mm Cu =>Exp Ind=0

[a] From empirical data

detector. Therefore, EI can be considered to be expressed in units of "mbels." It is important to note that the target EI value differs for general purpose (GP) and detail (HR) cassettes for this manufacturer.

Agfa CR uses an exposure indicator known as "lgM" which represents the logarithm of the median exposure value within a region of interest (ROI). Each image is assigned a user-selected speed class that determines the gain at which the image will be processed. Because of this, the actual radiation exposure required to produce a specific lgM value differs with different speed class setting. When the numerical value of lgM changes by 0.301 (the logarithm of 2), this indicates a factor of 2 difference in the exposure to the detector. Therefore, lgM can be considered to be expressed in units of "bels."

For the most part cassetteless DR manufacturers have been slow in developing exposure indices. Several of the vendors did not originally, and some still do not, incorporate a "true" exposure indicator, i.e., a quantity that reports radiation exposure to the detector. Instead, they relied on dose-area product (DAP), kerma-area product (KAP), or other quantities that represent an estimate of dose to the patient. These values were straightforward for the manufacturers to implement because the integrated systems allowed for knowledge of the generator settings, collimator field size, etc., which were used to calculate the value and are now required by IEC (and, hence, National Electrical Manufacturers Association [NEMA]). While these values may be of some use for calculating patient dose, they provide no useful information to the technologist with respect to the adequacy of radiation exposure to the detector.

Of those cassetteless DR systems utilizing detector exposure indicators, four are presented in the appendices: Imaging Dynamics; Philips; GE Healthcare; and Siemens Medical Systems. Not to be confused with the Kodak "EI," Philips uses an exposure index, "EI," that is inversely proportional to the air kerma, so that it somewhat parallels the S-number described above for Fujifilm. Unlike the Fujifilm approach, Philips conforms to the ISO 9236-1 convention for speed. The Philips EI also differs from Fujifilm S-number in that the scale used for EI is represented in bigger discrete steps (e.g., 100, 125, 160, 200, 250, 320, 400, 500, etc.). The EI steps are such that it takes approximately a 25% change in exposure for a change in EI step to occur, thus smaller changes in technique factor selection go undetected from an EI standpoint.

One of the key steps in calculation of any exposure indicator is the segmentation of anatomy or determination of the ROI. The determination of exposure indicator is oftentimes done with the same segmentation as that used for data scaling and grayscale processing. Many indices are quite sensitive to anatomical menu selection because the segmentation process is dependent on anatomical menu selection. In its more recent versions, Philips has improved upon this by decoupling the EI calculation from segmentation.

Imaging Dynamics has introduced a unique index called "f#." The value of the f# is a dimensionless scalar providing the technologist with an indication of the direction and magnitude of their technique selection versus an established target. Negative values represent underexposure and positive values indicate overexposure. The absolute value represents the deviation from the target exposure by factors of 2.

Canon introduced a cassette-based DR system for retrofitting existing x-ray generators. As such, the detector system had limited knowledge of exposure factors similar to that of cassette-based CR systems. Canon DR provided an exposure indicator called "Reached Exposure Value" or "REX." The numerical value of REX is a function of the "brightness" and "contrast" selected by the operator. For a

brightness setting of 16 and a contrast setting of 10, our empirical data at 70 kV_P plus 1.5 mm Cu indicates a REX of approximately 100 per mR with a correlation coefficient of 0.998. By admonishing the technologists against modifying brightness and contrast, REX has been demonstrated to have utility in oversight of exposure factor.[C6]

GE delayed introduction of a detector exposure indicator, instead using DAP for patient dose estimates as described above. However, on its most recent announced cassetteless DR product, GE incorporates three additional parameters indicating detector exposure, including a "Detector Exposure Index," or "DEI," which is a unitless metric comparing detector exposure to the expected exposure value.

All of these exposure indices share certain limitations. Calibration of the exposure indicator is one of the significant sources of variability among manufacturers. The accuracy of each indicator depends on proper calibration to a specific set of exposure conditions.[C7] The exposure conditions differ drastically among the manufacturers primarily with regard to use of added filtration or its absence. It has been shown that a hardened x-ray beam minimizes the sensitivity of the pixel value, and thus exposure indicator, to kV_P and beam energy variations.[C8] A filtered x-ray beam also gives a better clinical representation in that the energy spectrum is more similar to that exiting a patient and incident on the detector during clinical use.

Another limitation shared by the various exposure indices is sensitivity to the mathematical processes used to identify collimation boundaries and segmentation of the anatomically relevant data. The determination of the VOI is a key step in determining the exposure index. The mathematical algorithms should be robust enough to provide a reasonably accurate and reliable estimate of the exposure indicator regardless of collimation boundaries, anatomical positioning, inclusion of foreign bodies, etc., but this is not always the case. In addition, if these processes are performed in conjunction with the segmentation done for image processing purposes, the exposure indicator will be dependent upon the anatomical menu selection.

Some cassetteless DR vendors have implemented methods to address this issue. These methods, having evolved independently by different groups and based on different technologies and system architectures, vary widely. All methods in use today share a common end result: they all report a value that reflects the system sensitivity for a given exposure. This may be used to determine the exposure incident on the image detector. The value should be accurate, consistent, and reproducible. A system that provides inconsistent feedback may result in inconsistent image quality and causes confusion and frustration for the radiologists and the technical staff. Some systems only indicate the DAP to an ideal patient, which is of no use in managing image quality and only satisfies certain regulatory requirements.

The additional appendices contain more detailed descriptions of some of the approaches that have been developed by the various manufacturers.

References

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C2. Seibert, J.A., D.K. Shelton, and E.H. Moore. (1996). "Computed radiography x-ray exposure trends." *Acad Radiol* 4:313–318.

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C8. Tucker, D.M., and P.S. Rezentes. (1997). "The relationship between pixel value and beam quality in photostimulable phosphor imaging." *Med Phys* 24(6):887–893.

Appendix D

Agfa CR

Agfa CR systems provide exposure feedback for each acquired image in the form of a dose index called lgM. The lgM value indicates the deviation, expressed as the logarithm of the median exposure level in a calculated region of interest from the expected value. Similar to conventional radiography, the user selects this expected exposure value during the image acquisition process by choosing a speed class in the user interface. The speed class defines the operating point of the acquisition system.

For example, according to ISO 9236-1, a 400-speed screen/film system requires an average detector dose of 2.5 μGy to achieve a predefined aim density under specified exposure conditions. A speed class of 400 indicates that the CR system is adjusted to expect about 2.5 μGy detector dose at the center of its much wider (~500:1, or ~2.7 logE) operating range. The relationship between lgM, calculated detector dose, and speed class can be expressed as follows:

$$\mathrm{lgM} = 1.9607 + \log\left(\frac{Dose\left(\mu\mathrm{Gy}\right)}{2.5}\right) + \log\left(\frac{\text{Speed Class}}{400}\right).$$

Thus, if the calculated (median) detector dose for an image taken with speed class = 400 is 2.5 μGy, lgM will have its baseline, or reference value of 1.9607. If the detector dose is twice as high as expected for the selected speed class, lgM will increase by 0.301 (log2). If the detector dose is half as high as expected for the selected speed class, lgM will decrease by 0.301. Note that whenever Dose(μGy)*Speed Class = 1000 (analogous to ISO 9236-1), lgM always takes on its reference value. These relationships assume that the system's signal response (gray level vs. dose) has been calibrated according to Agfa's recommended procedure (which uses 75 kV_P, 1.5 mm Cu).

The calculation of lgM is based on a histogram analysis of the acquired (12-bit) image. The gray levels (called Scanned Average Level, or SAL in Agfa parlance) of this 12-bit image represent the square root of exposure, rather than the more commonly used log. This quantization scheme removes the signal dependence of the (Poisson) noise in the image, producing uniform noise amplitude everywhere. Regardless of the quantization scheme, histograms of radiographic images usually contain several peaks, corresponding, for example, to areas of beam collimation (low exposure), direct x-ray background (high exposure), and the anatomical region of interest between them (see Figure D1). Through spatial image segmentation and histogram analysis, the lgM algorithm first identifies the peak in the histogram (if there is one) corresponding to collimated areas and eliminates it from further consideration. By looking at the shape and amplitude of other peaks in the histogram, it then finds and analyzes the peak corresponding to direct x-ray background (if there is one). The remaining typically broader main peak is assumed to contain the relevant clinical information. This is the region of interest in which lgM is calculated. The algorithm first derives reasonable endpoints for this main histogram lobe, and finds its median value, which defines the lgM value for that image. By comparing the lgM value to the reference lgM value, the deviation from the expected detector exposure can be found. This information is stored in the image header and displayed on the output image.

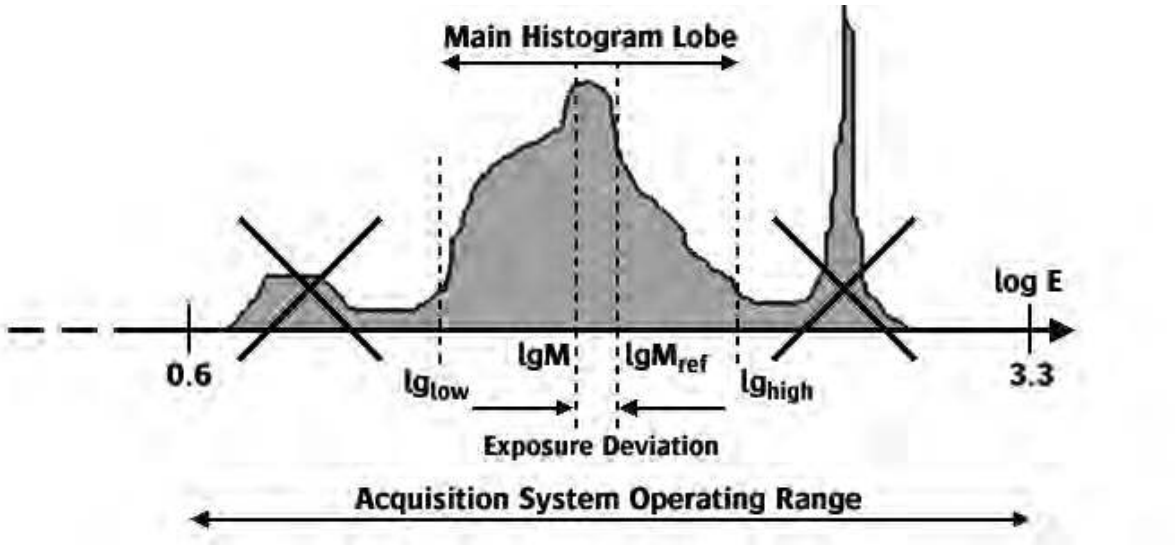

Figure D1. Schematic drawing of the histogram of a typical radiographic image.

In addition to providing per-image dose feedback, Agfa CR systems also provide tools to monitor dose per exam over time. The dose monitoring software enables each facility to set up reference dose (lgM_{ref}) values for up to 200 exam categories. This can be done by simply defining the expected values, or empirically during a learning phase, in which the software compiles statistics for and registers lgM values for 50 consecutive images in each category. When the lgM value of a newly acquired image deviates from the stored reference value for that exam category, the image is flagged, and the output image contains a numerical and visual (bar graph) display of the lgM value relative to the reference value that shows the extent of over- or underexposure. The software also maintains a history file containing dose (lgM) information for the last 50 exposures in each exam category so that radiologists, radiology administrators, or physicists can monitor exposure consistency and investigate/correct any occasional or systematic deviations.

Appendix E

Fujifilm FCR

Note: The following explanation is specific for FCR systems with 10-bit data output. The explanation is applicable in principle to FCR systems with greater than 10-bit output, recognizing that the 10-bit-specific references (e.g., 2^{10} and 1023) would be replaced with values appropriate for the output for the specific FCR system of interest.

Histogram analysis is used to define the wanted versus unwanted signals in a scanned image plate for a particular incident exposure and examination type. As the linear exposure latitude for the imaging plate is very wide, a variable reading sensitivity (sensitivity number, S) is necessary to map the stimulated luminescence of the imaging plate to a range of output digital numbers within a 10-bit range (1024 discrete gray levels).

In the ***Automatic*** mode, the PSP reader determines the latitude as well as the minimum and maximum stimulated luminescence values of the information extracted by the EDR process. The imaging plate is scanned directly using a combination of an analog logarithmic amplifier and a 12-bit (4096 gray levels) ADC encompassing the full dynamic range of the stimulated luminescence intensity at a fixed photomultiplier tube (PMT) sensitivity and gain. In order to normalize the image data and extract the desired range, an "electronic" EDR process is applied to the resultant data. Final image output is described by 10 bits (1024 gray levels). Values identified by the EDR process include the maximum and minimum log photostimulated luminescence (PSL) signals (S_1 and S_2, respectively) in the image histogram as shown in Figure E1. Examination-specific algorithms evaluate the shape of the histogram to determine the "useful" signal range. Within this range, the median input digital value, S_k, is "mapped" to the digital output value 511 in the 10-bit digital range. The *Sensitivity number*, *S*, is calculated as: $\boldsymbol{S = 4 \times 10^{(4-S_k)}}$, and is an index indicating the reading sensitivity and is inversely proportional to the incident exposure on the plate. The approximate relationship to the mean incident exposure is given as: exposure **(mR)** $\cong$ **200/*S*** for standard x-ray beam conditions (80 kV_P, 3 mm Al total filtration equivalent). The latitude number, ***L***, is an index representing the logarithmic range of digitization of the stimulated luminescence signals about the median value, S_k. ***L*** is calculated from the maximum and minimum luminescence values within the defined image area and the corresponding digital output values of the reading unit as:

$$\boldsymbol{L = 1023 \times (S_1 - S_2) / (Q_1 - Q_2)},$$

where Q_1 and Q_2 are the digital values corresponding to the log PSL output signals S_1 and S_2 of the reading unit, respectively. An example of an image histogram with the above-mentioned parameters is illustrated in Figure E1. An S_k value of 2.30 corresponds to an incident exposure of 1.0 mR. The latitude of the image reader and ST image plate usually ranges from a logarithmic PSL intensity of 0.3 (0.01 mR) to 4.3 (100 mR).

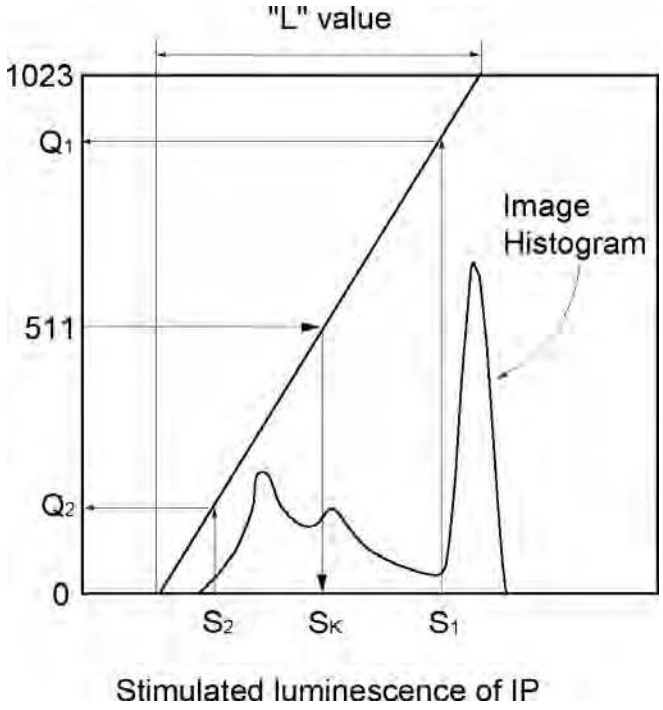

Figure E1. Sensitivity and Latitude numbers defined for the Fujifilm PSP system output parameters as related to the image histogram.

Appendix F

Kodak CR (now Carestream Health, Inc.)

Kodak DirectView DR and Kodak DirectView CR products provide the user with an EXPOSURE INDEX for each clinical image, which is a calibrated measure of the exposure incident on the detector. The following description of the EXPOSURE INDEX applies to CsI-based Kodak DirectView DR systems and Kodak DirectView CR systems used for general radiography applications. The common measure of receptor exposure reflects a highly integrated design philosophy for these products, which extends to the user interface and the underlying image data handling.

For-Processing Image

A FOR-PROCESSING IMAGE is computed from the RAW IMAGE DATA acquired for each image. The details of the computation depend on the technology. It is quite different for storage-phosphor-based CR images than it is for flat-panel DR images. However, in both cases the result is a FOR-PROCESSING IMAGE that is calibrated to an x-ray exposure under a STANDARD CALIBRATION CONDITION and represented on a common logarithmic scale. Kodak CR and DR systems allow users access to the FOR-PROCESSING IMAGE.

System Calibration

It is very useful to have a simple-to-reproduce, scatter-free exposure condition to calibrate digital detectors. Kodak CR and DR systems are calibrated at 80 kV_P with a 0.5 mm Cu and 1.0 mm Al added filtration at the x-ray tube housing. This choice for a STANDARD CALIBRATION CONDITION has been shown to minimize the sensitivity to small errors in kV_P[1] as well as to mitigate the effects of expected differences in inherent tube filtration. Kodak CR and DR systems are calibrated to produce a relationship between the FOR-PROCESSING IMAGE pixel values and the incident x-ray exposure given by

$$P = 1000 \cdot \log_{10}\left(\frac{K}{K_0}\right) + 1059,$$

where P is the pixel value, K is the incident air kerma in μGy, and K_0 is 1.0 μGy. Measurement of the incident exposure excludes the effects of backscatter from the CR or DR detector. CR values are for GP-25 storage phosphor plates and require a 5-minute delay between exposure and processing to be observed.

If measurements are made in milliroentgens, an alternate expression,

$$P = 1000 \cdot \log_{10}\left(\frac{E}{E_0}\right) + 2000,$$

where P is the pixel value, E is the incident exposure in mR, and E_0 is 1.0 mR, can be used.

[1] Samei, E., J.A. Seibert, C.E. Willis, M.J. Flynn, E. Mah, and K.L. Junck. (2001). "Performance evaluation of computed radiography systems." *Med Phys* 28:361-371.

Exposure Index

Image segmentation is a key step in processing the FOR-PROCESSING IMAGE of clinical images to create a FOR-PRESENTATION image that will be sent to a printer or to a Picture Archiving and Communications system (PACS). The purpose of segmentation is to identify an ANATOMICAL REGION OF INTEREST for each image. Proprietary algorithms detect and eliminate the FOREGROUND and BACKGROUND regions from consideration. FOREGROUND is that area of the image that is occluded by collimation. BACKGROUND is the image area that receives the x-ray exposure unattenuated by the patient. The remaining image area is evaluated with pixel-value and texture-sensitive algorithms to derive the unique ANATOMICAL REGION OF INTEREST for that image. Optimal tonal rendering is derived from histogram analysis of pixel values in the ANATOMICAL REGION OF INTEREST. The EXPOSURE INDEX for each image is the average pixel value of the FOR-PROCESSING IMAGE within the ANATOMICAL REGION OF INTEREST .

Exposure Index Reporting and Documentation

The EXPOSURE INDEX for each image is displayed on the graphical user interface of Kodak CR and DR systems. It is also incorporated into the DICOM header created for each image as DICOM tag (0018,1405). Other exposure-relevant information recorded in the DICOM header includes: kV_P (0018,0060), tube current (0018,1151), exposure duration (0018,1150), and the current-time product in mAs (0018,1152). The EXPOSURE INDEX for each image acquired is also entered into a log file on the acquisition system along with other relevant information, including the date, time, patient ID, body part, view, accession number, and image-reject comments (if any). Summary information is accessible to key operators (normally the chief radiographer or department administrator).

X-Ray Spectrum Dependence of Exposure Index

The response of DR systems is characterized by the relationship between incident air kerma dose and the pixel values in original images. Because system responses are x-ray-spectrum dependent, it is instructive to use the ISO 9236-1 standard, which specifies four x-ray beam conditions that span the range of common clinical examinations. These are intended to represent the beam conditions (including scatter) incident upon the detector for projection radiography of the extremities (ISO I), the skull (ISO II), the lumbar spine (ISO III), and the chest (ISO IV). The system response for the four ISO beam conditions, as well as the STANDARD CALIBRATION CONDITION, is given as an algebraic equation, represented in tabular form, and shown graphically below.

Algebraic Representation

The relationship between pixel value of the FOR-PROCESSING IMAGE and incident exposure can be summarized as

$$P = 1000 \cdot \log\left(\frac{K}{K_0}\right) + B,$$

where K is the incident air kerma in μGy, K_0 is 1.0 μGy, and B is a beam quality offset that depends upon the incident x-ray beam condition, or as

$$P = 1000 \cdot \log\left(\frac{E}{E_0}\right) + C,$$

where P is the pixel value, E is the incident exposure in mR, E_0 is 1.0 mR, and C is a beam quality offset that depends upon the incident x-ray beam condition. The constants for each beam condition are given in Table F1.

Table F1. Exposure Response Constants for Kodak's CR and DR Systems

	Kodak CR System		*Kodak DR System*	
X-Ray Beam	***B***	***C***	***B***	***C***
ISO – I	839	1780	648	1589
ISO – II	1059	2000	973	1914
ISO – III	1071	2012	1039	1980
ISO – IV	1059	2000	1025	1966
STD Calibration Condition	1059	2000	1059	2000

Tabulation

The relationship between pixel value in the FOR-PROCESSING IMAGE and the incident exposure is illustrated for Kodak CR and DR systems in Table F2 and Table F3, respectively. The values for the STANDARD CALIBRATION CONDITION (labeled STD) are by design the same for CR and DR systems. However, because of the differences in detector technology, the responses to the ISO beams differ.

Table F2. Kodak CR Systems (GP-25 Cassette): FOR-PROCESSING IMAGE **P**ixel **V**alues vs. *I*ncident **E**xposure

Air Kerma (μGy)	Exposure (mR)	Pixel Value				
		STD	ISO-I	ISO-II	ISO-III	ISO-IV
20.0	2.29	2360	2140	2360	2372	2360
17.8	2.04	2310	2090	2310	2322	2310
15.9	1.82	2260	2040	2260	2272	2260
14.2	1.62	2210	1990	2210	2222	2210
12.6	1.45	2160	1940	2160	2172	2160
11.2	1.29	2110	1890	2110	2122	2110
10.0	1.15	2060	1840	2060	2072	2060
8.93	1.02	2010	1790	2010	2022	2010
7.96	0.912	1960	1740	1960	1972	1960
7.10	0.813	1910	1690	1910	1922	1910
6.32	0.724	1860	1640	1860	1872	1860
5.64	0.646	1810	1590	1810	1822	1810
5.02	0.575	1760	1540	1760	1772	1760
4.48	0.513	1710	1490	1710	1722	1710
3.99	0.457	1660	1440	1660	1672	1660
3.56	0.407	1610	1390	1610	1622	1610
3.17	0.363	1560	1340	1560	1572	1560
2.83	0.324	1510	1290	1510	1522	1510
2.52	0.288	1460	1240	1460	1472	1460
2.24	0.257	1410	1190	1410	1422	1410
2.00	0.229	1360	1140	1360	1372	1360
1.78	0.204	1310	1090	1310	1322	1310
1.59	0.182	1260	1040	1260	1272	1260
1.42	0.162	1210	990	1210	1222	1210
1.26	0.145	1160	940	1160	1172	1160
1.12	0.129	1110	890	1110	1122	1110
1.00	0.115	1060	840	1060	1072	1060

Table F3. Kodak DR Systems: FOR-PROCESSING IMAGE **P**ixel **V**alues vs. **I**ncident **E**xposure

Air Kerma (µGy)	Exposure (mR)	Pixel Value				
		STD	ISO-I	ISO-II	ISO-III	ISO-IV
20.0	2.29	2360	1949	2274	2340	2326
17.8	2.04	2310	1899	2224	2290	2276
15.9	1.82	2260	1849	2174	2240	2226
14.2	1.62	2210	1799	2124	2190	2176
12.6	1.45	2160	1749	2074	2140	2126
11.2	1.29	2110	1699	2024	2090	2076
10.0	1.15	2060	1649	1974	2040	2026
8.93	1.02	2010	1599	1924	1990	1976
7.96	0.912	1960	1549	1874	1940	1926
7.10	0.813	1910	1499	1824	1890	1876
6.32	0.724	1860	1449	1774	1840	1826
5.64	0.646	1810	1399	1724	1790	1776
5.02	0.575	1760	1349	1674	1740	1726
4.48	0.513	1710	1299	1624	1690	1676
3.99	0.457	1660	1249	1574	1640	1626
3.56	0.407	1610	1199	1524	1590	1576
3.17	0.363	1560	1149	1474	1540	1526
2.83	0.324	1510	1099	1424	1490	1476
2.52	0.288	1460	1049	1374	1440	1426
2.24	0.257	1410	999	1324	1390	1376
2.00	0.229	1360	949	1274	1340	1326
1.78	0.204	1310	899	1224	1290	1276
1.59	0.182	1260	849	1174	1240	1226
1.42	0.162	1210	799	1124	1190	1176
1.26	0.145	1160	749	1074	1140	1126
1.12	0.129	1110	699	1024	1090	1076
1.00	0.115	1060	649	974	1040	1026

Graphical Representation

The dependence of pixel value on incident exposure for the STANDARD CALIBRATION CONDITION as well as the four ISO beam conditions is shown graphically in Figure F1 for the Kodak CR systems and in Figure F2 for Kodak DR systems. For the CR systems, only the ISO-I (extremity) condition results in a significantly different response. The other three ISO beam conditions and the calibration beam all result in system responses that are nearly identical and therefore are plotted as a single line.

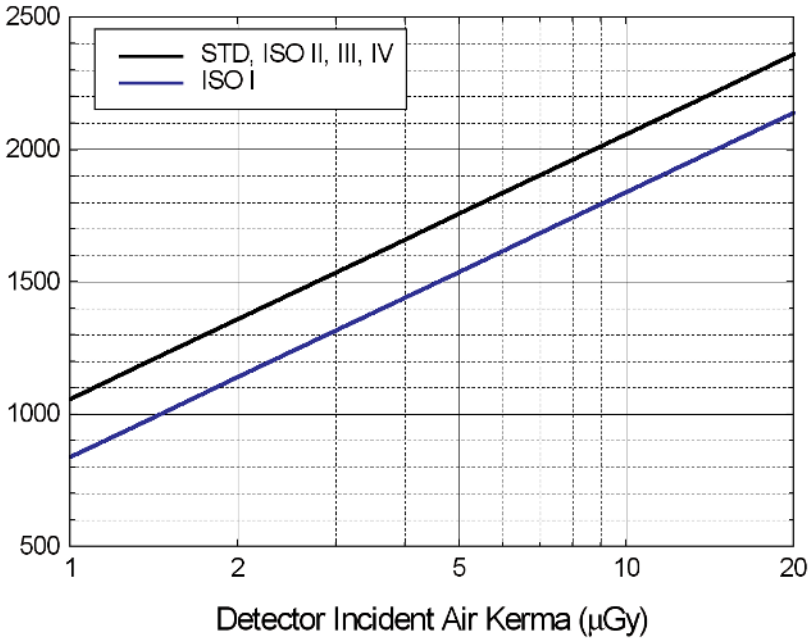

Figure F1. Kodak CR systems (GP-25 cassette): FOR-PROCESSING IMAGE pixel values vs. incident exposure.

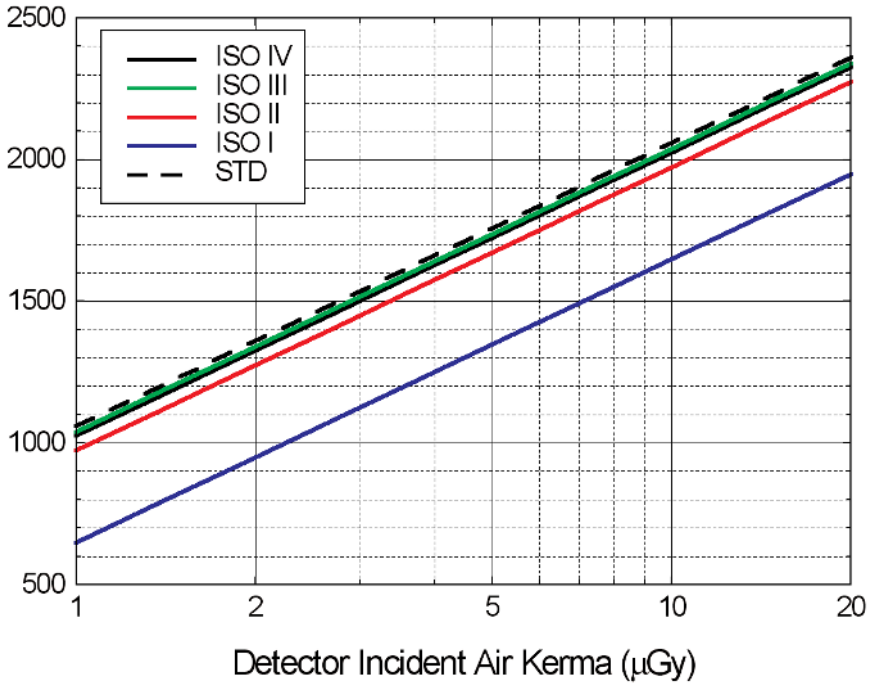

Figure F2. Kodak DR Systems: FOR-PROCESSING IMAGE pixel values vs. incident exposure.

Appendix G

Konica Minolta CR

The amount of photostimulable light emission versus incident x-ray dose exhibits good linearity over a range of over 4 orders of magnitude. The REGIUS 12-bit (4096 level) quantization range (QR) is specified by a QR parameter such the range of quantized exposures X is according to the relationship:

$$200/QR \times 1[mR] \times 10^{-1.5} < X < 200/QR \times 1[mR] \times 10^{+2.5}. \quad (a)$$

For example, if QR=200, the quantized range is $10^{-1.5}$ to $10^{+2.5}$ mR, or 0.0316 to 316 mR. Output signals are proportional to $\log_{10}$ (mR), so an incident detector exposure of 0.0316 mR is mapped to value 0, and of an exposure of 1 mR is mapped to an output signal 1535. (The REGIUS is calibrated using the image from an exposure corresponding to a beam quality of 80 kV, 2.0 m).

The QR parameter is related to an equivalent screen/film system speed (referred to as an S-value) as follows. Let R be the incident x-ray exposure that produces a REGIUS output value of 1535 and a printed film optical density (using a fixed printer mapping) of 1.2 within a specified image area of interest. From expression (a) above, the corresponding exposure is readily determined as:

$$X = R = 200/QR. \quad (b)$$

If read out with QR=200, R equals 1 mR; if read with QR=400, R is 0.5 mR, etc. Suppose it is desired to darken (or lighten) the printed film such that some other pixel value within a different area of interest is printed with an optical density of 1.2. REGIUS gradation processing uses an S-value to adjust output pixel values to achieve the corresponding darker or lighter printed image (using the same printer mapping). Let R` be the actual x-ray dose required to produce a screen/film image optical density of 1.2 in the desired region. This S-value is then defined as:

$$S = QR \times R/R` \quad (c)$$

where R depends on the QR parameter as given in expression (b) above. From its definition and the above discussion, we observe the following properties of the S-value:

(1) S-value is independent of the QR parameter; by substitution of (b) into (c), S = 200/R.
(2) S-value is determined from pixel values obtained following gradation processing. Gradation processing is determined by Konica Minolta's original auto-gradation processing algorithm; however, this can be manually changed by the operator.
(3) S-value is in inverse proportional to x-ray dose; i.e., for exposures of the same object under identical conditions, if the x-ray dose is n-times, the S-value will be 1/n-times.
(4) When S-value of the image is 200, the incident x-ray exposure to the object area (especially the region of interest) output with a printed film density 1.2 is 1 mR.

A consequence of property (2) is that S-value is not uniquely determined by the amount of x-ray exposure. However, for any particular (exam type-specific) suitable gradation processing, properties (1) and (2) allow the S-value to serve as a very useful relative exposure index.

Appendix H

Imaging Dynamics

The goal is to determine a method for providing an index of exposure which was easy for the technologist to understand and which would be traceably related to an existing and well-recognized standard.

An exposure of 1 mR will produce a mid-range optical density on a 200-speed screen/film system. An exposure of 0.5 mR will produce the same optical density on a 400-speed screen/film system. Our exposure index is based on this relationship. We determine for an image the estimated radiation input to produce its mid-range density and relate that to the screen/film system speed that would have responded in a similar fashion.

The first requirement is a calibration of the digital system's response to input radiation. To emulate the typical exit spectra from a patient's body, we harden the beam by adding 1 mm of copper filtration. Measurements are then made at 80 kV_P to determine the system input/output characteristic in terms of milliroentgens per digital number (mR/DN). This input/output ratio, R_{1O}, is stored as a system characteristic and does not need to be recalculated on an image-by-image basis.

To calculate the exposure index, which we refer to as the f# due to its similarity to aperture f-stops on a lens, we segment the image to exclude areas of direct exposure and areas outside the exposed region. The remainder represents the patient anatomy. The median value, I_{Med}, of the anatomy histogram is found. Experimental data has shown that using the median value will in most cases give an accurate representation of the mid-range optical density in the image. It is not unduly skewed by small errors in image segmentation. This is not the case if the mean is used, where a significant shift in value occurs if the segmentation happens to include some areas of direct exposure.

Using the input/output ratio R_{1O} and the median value I_{Med}, we calculate:

$$X_{Med} = I_{Med} \cdot R_{1O},$$

where X_{Med} is the radiation level which resulted in the mid-range density.

Effective speed, S_E is therefore given by:

$$S_E = \frac{200}{X_{Med}}.$$

In clinical practice, different screen/film types are used for different examinations. A 400-speed system is common for chest exams, while 100 speed is common for extremities such as hands. The slower speed gives finer detail.

Our exposure index takes this into account by relating the effective speed S_E to a target speed S_T. The target speed is determined on an institutional basis and is stored in an anatomical parameters database for reference when processing each image.

The radiation technologist should not be required to remember what the correct screen/film speed equivalent is for each anatomy in assessing the exposure index. We therefore calculate the relationship of the effective speed to the target speed. By expressing this as a log base 2, thus:

$$f\# = \log_2\left(\frac{S_T}{S_E}\right).$$

The resultant index is dimensionless and applies equally regardless of the target speed. The radiation technologist knows that they are aiming for an $f\#$ of zero. Underexposure by a factor of 2 will give a value of −1, by a factor of 4 will give −2. Overexposure by two times will give a value of 1; by four times will give a value of 2. In practical terms, the technologist can reasonably expect that if an image with $f\#$ between −1 and +1 is obtained, the exposure will be of reasonable diagnostic quality. It is worth noting that no exposure index derived solely from the properties of the image can be completely reliable in clinical use. There will always be a small number of cases where unusual pathology or implants will alter the overall balance such that the image will give a misleading index. Technologists should be advised to use their own best judgment in conjunction with the exposure index to determine if an image needs to be repeated. It would be inappropriate, for example, to repeat an image based on a high $f\#$ simply because it lies outside the institution's guidelines if the resulting image is of very high quality.

One advantage of this numbering system is that technologists are generally familiar with the range of density settings on the automatic exposure control (AEC), or phototimer system. The *f*#'s described here may be thought of as somewhat analogous, with plus and minus densities. By using a system that echoes an already familiar numbering scheme, the technologists are more likely to be comfortable with it and pay closer attention to the results. By contrast, other exposure indices in use today require the technologist to understand less intuitive numbering schemes. For example, one widely used system displays the log median value and asks the technologist to target a proper exposure around 2.2.

Each 0.3 increment or decrement represents a doubling or halving of exposure, respectively. While this is certainly a viable scheme, it is not intuitive.

Appendix I

Philips DigitalDiagnost Exposure Index

General

The Philips DigitalDiagnost flat-panel DR system calculates an exposure index (EI) for every image. The EI is inversely proportional to the detector air kerma, *K*, and is derived from a "characteristic" pixel value of the image.

The scaling of the EI is defined in a way similar to screen/film speed (ISO 9236-1):

$$\mathrm{EI} = 1000 / K \tag{1}$$

where *K* is the air kerma in μGy at the detector entrance.

The air kerma *K* is obtained from the characteristic pixel value PV_c and the sensitivity *SENS* of the detector, expressed in digital numbers per μGy:

$$K = PV_c / SENS. \tag{2}$$

The sensitivity of the flat-panel detector after applying the standard detector-specific corrections is $SENS = 207\ \mu\mathrm{Gy}^{-1}$, for a beam quality corresponding to RQA5 according to IEC 61267 (70 kV, 21 mm Al added filtration, HVL 7.1 mm Al).

Exposure Index Values

The exposure index for the DigitalDiagnost is intentionally confined to values that follow the ISO R'10 scale, well-known from, e.g., screen/film speeds. The numbers calculated according to equations (1) and (2) are thus rounded to the values …, 100, 125, 160, 200, 250, 320, 400,… (see Table I1). Each step correponds to a factor of $10^{0.1}$ (or an increase by about 25%).

The rationale for this grading is the following:

- under clinical conditions the reproducibility of the EI for fixed detector exposure conditions is approximately of this size, owing mainly to variations in image/histogram evaluation for different patients/examinations;
- one step of the ISO R'10 scale corresponds to one "exposure point," which is a scale well-known to most x-ray techs and is, for example, also used for the grading of the mAs scale on many x-ray generators.

Table I1. Relation between Detector Exposure/Air Kerma, Pixel Value, and Exposure Index(for Beam Quality RQA5)

Detector Exposure Ks [mR]	Air kerma Ks [µGy]	Pixel value PV_c (lin scale)	Pre image pixel value (log scale)	EI
5.10	44.67	9291	23808	20
4.05	35.48	7380	23208	25
3.22	28.18	5862	22608	32
2.56	22.39	4657	22008	40
2.03	17.78	3699	21408	50
1.61	14.13	2938	20808	63
1.28	11.22	2334	20208	80
1.02	8.91	1854	19608	100
0.81	7.08	1473	19008	125
0.64	5.62	1170	18408	160
0.51	4.47	929	17808	200
0.41	3.55	738	17208	250
0.32	2.82	586	16608	320
0.26	2.24	466	16008	400
0.20	1.78	370	15408	500
0.16	1.41	294	14808	630
0.13	1.12	233	14208	800
0.10	0.89	185	13608	1000
0.081	0.71	147	13008	1250
0.064	0.56	117	12408	1600
0.051	0.45	93	11808	2000
0.041	0.35	74	11208	2500
0.03	0.28	59	10608	3200
0.03	0.22	47	10008	4000
0.02	0.18	37	9408	5000
0.02	0.14	29	8808	6300
				8000

Determination of Characteristic Pixel Value

An x-ray image usually contains a wide range of pixel values. An important step in the calculation of the exposure index according to equations (1) and (2) is to determine a characteristic pixel value PV_c, i.e., a pixel value that corresponds to the average detector signal representing the target area of the examination.

This process usually comprises two steps:

- the determination of an subarea (ROI) of the full image, containing the target area;
- the determination of the characteristic pixel value in this ROI. This can be the average or the median pixel in this subarea; however other, more sophisticated algorithms involving the pixel histogram may also be used.

Slightly different approaches are used in different software releases of the DigitalDiagnost. *Up to and including release 1.2* the determination of the characteristic pixel value is coupled to the "ranging" algorithm which detects the exposed area and finds specific pixel values from the (cumulative) histogram, used to adapt the display look-up table (LUT). Since the LUT adaptation may be dependent on the selected image type (examination/anatomy), the characteristic pixel value and such the EI may depend on the type of examination, even for similar histograms.

Starting with release 1.3 the determination of the characteristic pixel value for the EI-determination is decoupled from the determination of ranging parameters and is independent of the type of examination.

For images with automatic exposure control (AEC), PV_c is determined as the median pixel value in the area(s) corresponding to the activated measuring field(s) of the AEC.

For images with manual selection of exposure parameters, the area in which PV_c is determined is defined as follows: the characteristic pixel value is defined as the median pixel value of the center 25% area of the image (called the "quarter field"). Collimated and direct radiation areas are masked out before calculating the median.

kV Correction

As the sensitivity of the detector changes with x-ray photon energy, the relation between pixel value and incident air kerma is not fixed for different beam qualities. Consequently, a given EI will correspond to different exposure values (air kerma values) for different kilovoltages. This effect is most pronounced for low kV_P, where the sensitivity (pixel value/μGy) may be only 30% of that at 70 kV. To mitigate this effect, a kV correction factor is applied in the EI calculation in the DigitalDiagnost (starting with release 1.2), which compensates for changes in the sensitivity (see Figure I1).

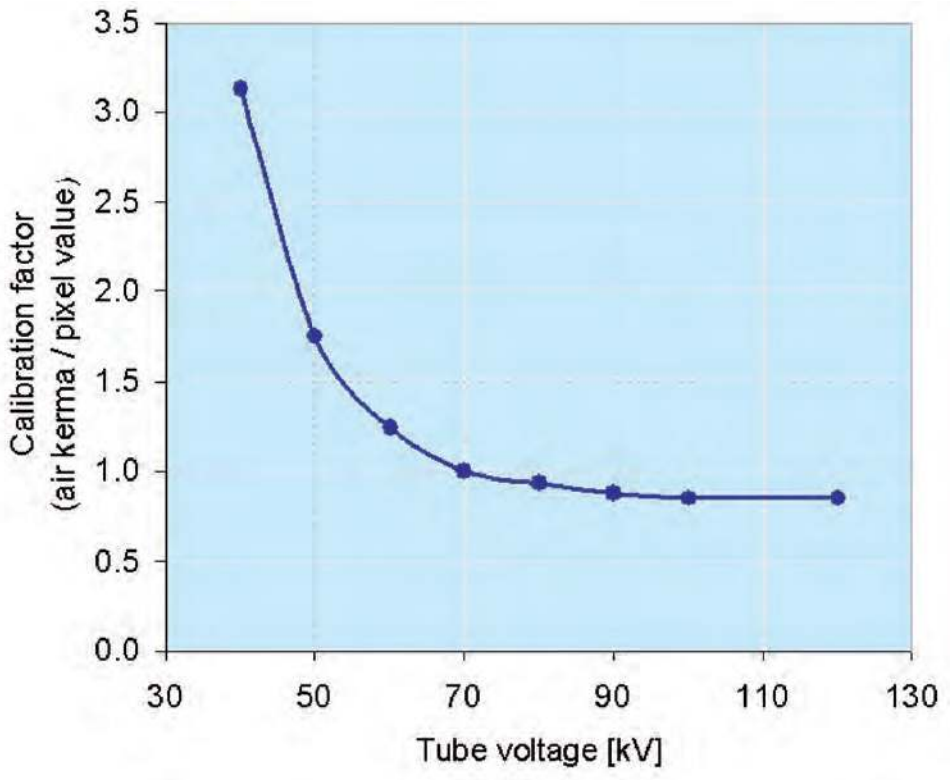

Figure I1. kV correction factor for the Exposure Index as used in the DigitalDiagnost.

Appendix J

GE Healthcare

Definium Systems

GE digital radiography *Definium* systems provide the user with three values related to the incident exposure (air kerma) to the digital detector:

- Uncompensated Detector Exposure (**UDExp**) in μGy
- Compensated Detector Exposure (**CDExp**) in μGy
- Detector Exposure Index (**DEI**), unitless

UDExp, CDExp, and DEI are displayed as (optional) image annotations on the acquisition workstation and are stored in the image DICOM header. DEI is also displayed on the acquisition user interface in order to provide quick feedback to the radiography technologist.

1. **Anatomy Segmentation:** As a preliminary step, the anatomy regions are identified in the *raw* image and a Median Anatomy Value (A_M) in unit of *counts* is determined. (The raw image is the image read from the digital detector after bad pixel, offset, and gain correction.)

2. **Detector Sensitivity:** Two sensitivity values are estimated for the digital detector based on calibrations at the manufacturer and/or by field engineers.
   - Uncompensated Detector Sensitivity, USens, is defined as the conversion efficiency of the detector in units of counts/μGy at 80 kV_P (with a standard filtration and no anti-scatter grid).
   - Compensated Detector Sensitivity, CSens, is equal to USens after compensation for kV_P, presence of anti-scatter grid, and user-selected additional collimator Cu filtration.

3. **Detector Exposure:** Two exposure values are calculated using USens and CSens.
   - Uncompensated Detector Exposure, **UDExp** = A_M/USens
   - Compensated Detector Exposure, **CDExp** = A_M/CSens

4. **Detector Exposure Index:** DEI is a unitless normalized metric relating obtained median anatomy count value (A_M) to *expected* count value for the used technique (kV_P, filtration, and anti-scatter grid). Expected count values are derived using the Automatic Exposure Control (AEC) with standard acrylic phantoms, appropriate over the kV_P range.

   Hence, DEI can be effectively used for indicating
   - Over-/underexposure due to patient mispositioning or incorrect selection of ion chambers with AEC acquisitions, and
   - Over-/underexposure due to inappropriate technique selection with fixed-time acquisitions.
   - Over-/underexposure due to other operator-related or system-related events.

DEI values are displayed on the acquisition user interface after each acquisition, along with a DEI Lower Limit and a DEI Upper limit that are user-configurable for each anatomical view and patient type (Adult vs. Pediatric).

For each acquisition performed on the system, **UDExp**, **CDExp**, and **DEI**, along with other technique and acquisition information, are logged on the system and can be exported to a .csv file on CD at any time.

General Properties

- **UDExp**, **CDExp**, and **DEI** are independent of image processing and reprocessing. The values, once calculated, are stored in the DICOM header and cannot be modified by the user.
- **UDExp**, **CDExp**, and **DEI** are potentially affected by errors in the identification of the anatomical regions in the image, the identification of collimated regions in the image (auto-shuttering), or the presence of large-area prosthesis/shielding. Before closing an exam, the user has the option of modifying/correcting the auto-shuttering and reprocessing the image to get more accurate **UDExp**, **CDExp**, and **DEI** values. (Note: The image processing will NOT affect the DEI metrics. Only the shutter location will.)

Technical Mode

A user with administrative privileges on the system can select the Technical Mode for detector exposure metrics, whereby all metrics are calculated based on a 512×512 pixel square in the center of the image. In this mode, the result of *Anatomy Segmentation* is not used in the calculations. This mode can be useful when imaging non-humanoid phantoms.

XR/d and XQ/i Systems

The GE XQ/i and XR/d digital detector systems do not directly report a detector-exposure or sensitivity index to the operator. A "Sensitivity" value is embedded in the DICOM header. This value is related to detector exposure through the original pixel value.

For images other than those of Chest exams, feedback is given to the operators on adequacy of technique by darkening or lightening the final image depending on what raw mean signal value in the recognized image histogram is present in the image relative to the expected value for the selected anatomical view (this behavior no longer applies in the new *Definium* systems, which provide detector exposure metrics).

Appendix K

Alara CR

In order to optimize radiation dose, image quality, and use of the CR reader's dynamic range for a wide variety of radiographic studies, Alara's CR product provides the capability of changing system gain. Analogous to screen/film radiography, we have called the various gain settings Speed Classes (SC). The nominal or target x-ray exposure for each SC is summarized in Table K1.

Table K1. Target X-Ray Exposures for Various Speed Classes

Speed Class	*Target Exposure (mR)*
50	4.0
75	2.67
100	2.0
200	1.0
300	0.67
400	0.5
800	0.25

For each image, an Exposure Indicator Value (EIV) is computed. For all speed classes, and for Standard and High Resolution modes, the target EIV is 2000. The EIV is logarithmically related to the energy deposited in the plate: changes of 300 in the EIV correspond to changes by factors of 2 in exposure.

The EIV for each Alara CR device is calibrated using an RQA5 x-ray spectrum (70 kV, 21 mm Al added filtration, HVL = 7.1 mm Al; IEC 61267:1994). The gains (PMT voltage settings) required to achieve a target digital count for each of the Target Exposures listed in Table K1 are determined. Thus for a particular device, a table of PMT voltage settings by Speed Class is generated. At system installation, each exam type is assigned a speed class according to site preferences. Subsequent selection of exam type automatically selects the speed class.

Prior to image processing, the EIV is computed from the 16-bit, linear-with-exposure image according to the following basic steps:

1. Using a combination of histogram and morphological analysis, image regions corresponding to overscan, x-ray beam collimation, and direct exposure are identified.

2. The mean pixel value of the remaining region, which corresponds to anatomy, is computed and converted to mR via the RQA5 calibration mentioned previously.

3. EIV is computed according to:

$$\mathrm{EIV} = 1000 \cdot \log_{10}\left[\mathrm{ResScale} \cdot \frac{\mathrm{SC} \cdot \mathrm{mR}}{2}\right]$$

where ResScale accounts for the slightly different system response at Standard and High Resolution. SC and mR refer to the system speed class and the mean anatomy exposure in mR, respectively.

The EIV is displayed as a numerical value on the image on the QC workstation, and is shown graphically on a horizontal scale along the bottom of the thumbnail views of the images. The EIV is also stored in DICOM tag (0018,1405) (Relative X-Ray Exposure). Alara provides tools to analyze EIV trends.

Appendix L

Siemens Medical Systems

Description of the exposure index from Siemens Medical System's digital radiographic systems is summarized in Table L1.

Table L1. Exposure Index Description for Siemen's DR Systems

Nr.	*Question*	*Device*
1.1	Company	Siemens AG Medical Solutions
1.2	X-ray equipment type	AXIOM Aristos (FX, MX, TX, VX) Digital Flat-Detector
2	Name of the exposure index (if more than one, please add an extra sheet for each)	Exposure Index, abbrev.: EXI
3	What is the notation in the DICOM header and what does the number mean?	DICOM Group/Element: 0018,1405 Relative x-ray exposure; Direct declaration of the EXI-value as data type IS, no conversion needed.
4	Where is the exposure index?	It is displayed as image-legend on softcopy and hardcopy devices.
5	Definition of the used exposure index	The exposed field is subdivided in a 3×3 matrix, where the central segment is defined as the region of interest. The exposure index is calculated as the average out of the original pixel values in the central segment. The calculation scheme is independent of the selected organ program, the exposure method, the measuring field (when using AEC), and the image processing parameters.
5.1	Functional relationship between exposure index and dose	The EXI-value is a relative value, directly proportional to the dose. Doubling of the absorbed dose in the image receptor results in a doubling of the EXI-value.
5.2	Calibration conditions	For 70 kV and an added filter of 0.6 mm Cu (following beam quality RQA5 in IEC 61267:1994-09), a calibration factor c is determined and documented in the system manual: Air Kerma [Gy] = c × EXI.
5.3	Dependence (e.g., tube voltage, collimating, organ range, selected organ program,...)	Depends on: collimation, beam quality, examined organ. Depends not on: organ program name, selected exposure method (manually or automatically), selected measuring field.
5.4	Accuracy of the exposure index (accuracy of calibration, relationship to image receptor dose or speed)	Calibration factor c: ±10% (uncertainty of dose measurement) at calibration conditions. Absolute values at identical doses: the tolerance of the conversion factor (Pixel value/dose) for different detectors is ±15%.
5.5	Reproducibility	<5% (limited by reproducibility of generator and automatic exposure control, respectively).
5.6	Using at technical exposures	Currently available test phantoms for acceptance and constancy testing can be used without modification. At similar positioning of the test phantom, collimation, and exposure parameters,a deviation of ±10% indicates a significant change.

(Continued)

Table L1. Exposure Index Description for Siemen's DR Systems (Continued)

Nr.	*Question*	*Device*
5.7	Precision of the exposure index in constancy tests (and at constancy conditions)	Scaling: EXI-values are scaled as the original 14-bit pixel values. (*quod vide* 5.5 and 5.6)
6	What statements are possible with the exposure index in the medical image and what conditons and limits are valid?	The EXI-value is a relative measure for images of the same type, acquired at user-defined standard protocols. The EXI-values can vary for different organs and projections. With the EXI-value it is possible to discover: – changes in the dose-presettings (speed) – changes in the selection of measuring fields – wrong positioning of the measuring field w.r.t. the organ.
7	That does the company think about aims of the statement or usage of the discussed exposure index?	1. Control of the system components "AEC" and "image receptor" for constancy test. 2. Control of the exposure parameters in clinical routine.

Appendix M

iCRco

The exposure index is the key link between the x-ray physics at the site and the specific capture device. It represents an estimate of the radiation a patient receives from an x-ray exposure. In conventional screen/film x-ray, the dose can be estimated by the darkness of the x-ray film itself. For CR systems, the appearance of the digital x-ray image on a computer monitor does not depend on the dose level. However, the actual pixel values recorded by the PMTs correlate well with the actual x-ray dose. Therefore, it is possible to calibrate a CR device to act as a dosimeter: for example, the dose captured by the phosphor plate can be accurately measured using an iCRco CR scanner and translated into an exposure number.

A Generalized and Practical Aproach

The mature status of CR technologies combined with the competitive nature of the marketplace has resulted in vendor-specific methodologies for computing exposure index. This has detracted from the original purpose of the exposure index: to provide a level of confidence of image quality to the technologists while minimizing radiation to the patient. This creates confusion and an unnecessary barrier in the migration to digital.

iCRco has developed an approach that accomplishes two important goals: (1) it facilitates migration towards a generalized standard, and (2) it provides a practical easy-to-use metric for end-users to apply in a clinical environment.

Our method provides a dimensionless number representing a level 0 performance relative to an expected anatomy-specific exposure. This is not too different from the number generated by the method developed by Imaging Dynamics Company (IDC). For example, in IDC's approach, an exposure of –2 is a 4 times underexposure, –1 is 2 times underexposure, 0 is perfectly exposed, +1 is 2 times overexposed, and +2 is 4 times overexposed.

The analytics behind our approach have the fine precision required for an exact continuous numerical indicator if required. However, for practical applications we have simplified the display by grouping "performance ranges" into a discrete number ranging from –2, –1, 0, +1, +2, see Figure M1. The goal for the user is to keep the images in the neutral range (around 0) independent of the anatomical view.

Graphically, we have color coded the number shown on the computer screen, if the level is within the –2 to +2 range, the number is shown in green (as shown in Figure M2). If the level is outside the –2 or +2 range, the number is shown in red.

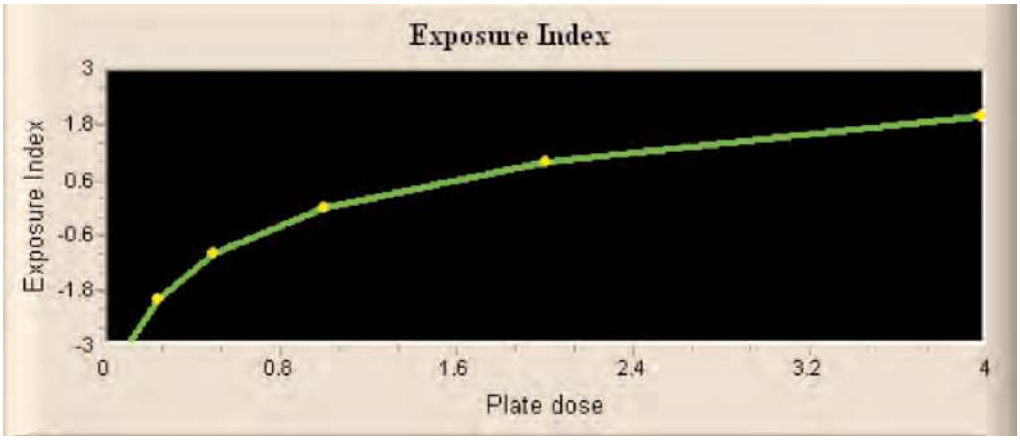

Figure M1. Relative exposure index as a function of plate dose in mR.

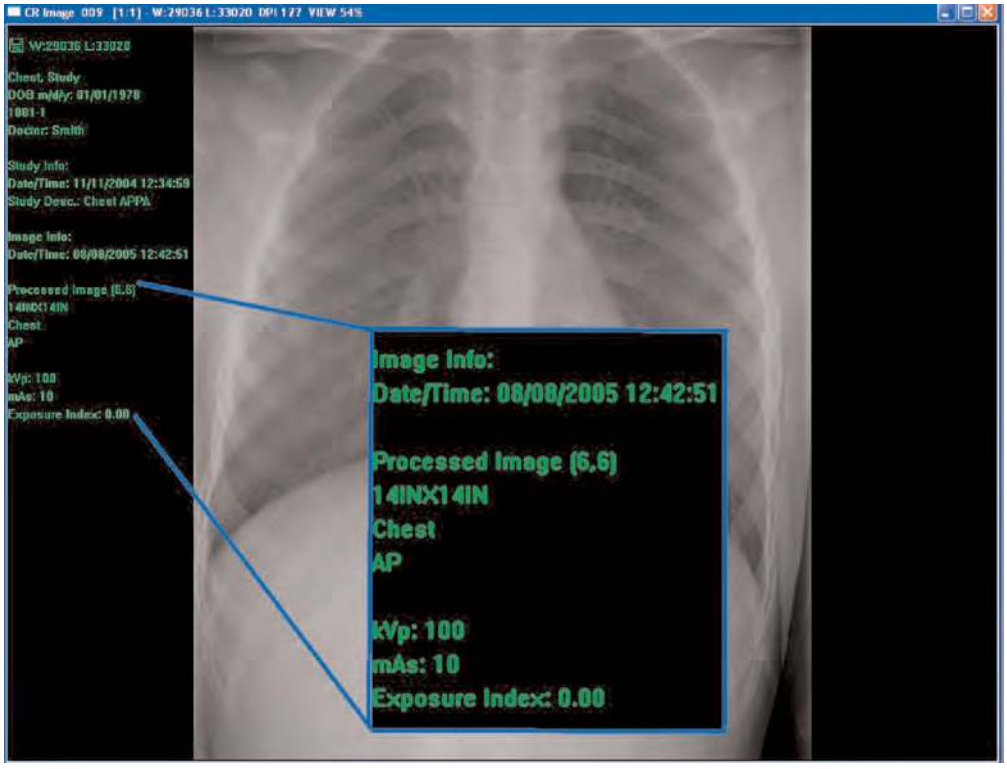

Figure M2. Magnified view of display of exposure index. For values in the –2.0 to +2.0 range the exposure index is displayed in green; if outside the range, the value is displayed in red to signal to the technician that the exposure is out of range.

Exposure Index Calibration Wizard

iCRco's equipment comes with a built-in software module which calibrates the CR device to the x-ray at the site. The software presents an interactive step-by-step procedure that can be performed during installation and maintenance. Reference exposures are taken at progressively increasing levels of dose and measured using a dosimeter. These calibrated exposures serve as the reference levels for the exposure index. The objective is to calibrate the device around an 80 kV_P source, hardened with 1.5 mm Cu at the x-ray source. The pixel analysis is performed on an area corresponding to 80% of a 14 by 17 in. cassette. The software performs a regression by which the linearity of the system is evaluated as a function of dose, as shown in Figure M3.

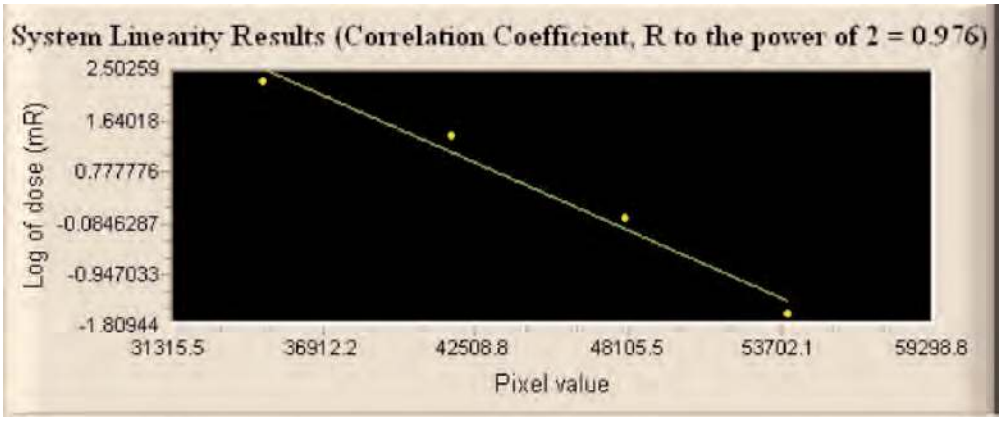

Figure M3. Linearity of the exposure index is shown as a function of log of dose and its corresponding pixel value.

There is a built-in table of anatomy-specific target exposures which serve as reference.

The resulting exposure index is a ratio of the pixel value analysis as described above and the expected reference value for the given anatomy. The exposure index represents a deviation from the reference. Of course, the reference levels can be adjusted based on the policy and tastes of users at the site.

To facilitate and demonstrate the generalized approach, in our system we have implemented translation of our numbers into exposure numbers of other vendors including Fujifilm S Number and Agfa IgM. Translation into other methodologies can be implemented as well.

Appendix N

Canon Medical Systems

REX Value and Exposure

With digital x-ray systems, including the CXDI system, by using the analysis explained in the previous section, the density of the image is always automatically adjusted to a specific level regardless of the dose amount. Even if the density is constant, however, when the dose is high, a clear image with good SNR is produced, and when it is low, noise is conspicuous in the resulting image.

The level of the dose is determined by the level of the image quality allowed by the doctor for examination and diagnosis.

If the image is always output with a constant density, particularly with phototimer radiography, it is not known at what level of dose the image was actually taken.

For this reason, an index is needed as a guide to the level of dose used.

The REX value (Reached EXposure Value) is defined as having the purpose of indicating the dose.

Before explaining the REX value, a typical example of the dose index using the two-point method for analysis is shown in Figure N1.

As shown in the figure, with the two-point method, the largest and smallest pixel values corresponding to the image are obtained from the histogram and the midpoint is taken as the dose index.

With the one-point method, on the other hand, the simplest method is to use the pixel values obtained by analysis directly.

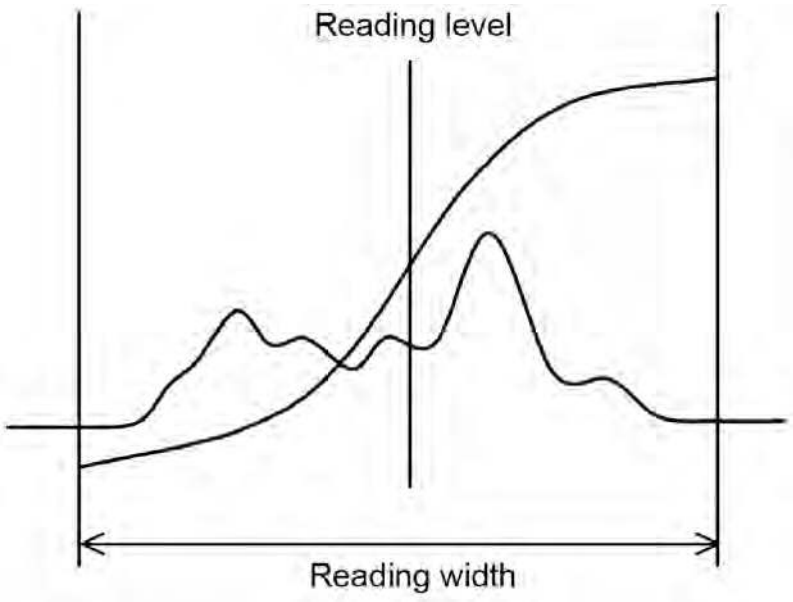

Figure N1.

If this method is used, however, the pixel value obtained by analysis is completely different for each region and the stability of the analysis is directly affected, making it an ambiguous standard for the dose index.

With the CXDI system, therefore, as shown in Figure N2, without directly using the pixel value obtained by analysis, the reference density is specified as 0.75D and the pixel value that gives this density is used as the dose index. This is called the REX value.

However, as a proportional relation with the dose is not established when the pixel value after log conversion is used, the pixel value before log conversion is used as the REX value.

The definition of the REX value is explained here using detailed illustrations. Figure N3 shows the definition of the REX value using a thoracic image as the example.

The 1st-4th quadrants are defined by counterclockwise rotation, where the 1st quadrant is on the top right. The 1st quadrant is the LUT curve showing the relation between system output and film density; the 3rd quadrant is the sensor characteristic curve showing the relation between the dose detected by the sensor and the sensor output value; and the 4th quadrant is the linear-log curve for converting the sensor output value proportional to the dose into the system output proportional to the logarithm of the dose.

The following is a brief explanation using only the 3rd and 4th quadrants in the right half of the figure.

First, to explain the algorithm by which the LUT curve is determined.

The CXDI determines the feature pixel value of the target chest image by automated analysis.

In this figure, it is assumed that the sensor output value corresponding to this pixel value (pixel value before log conversion) is 1220.

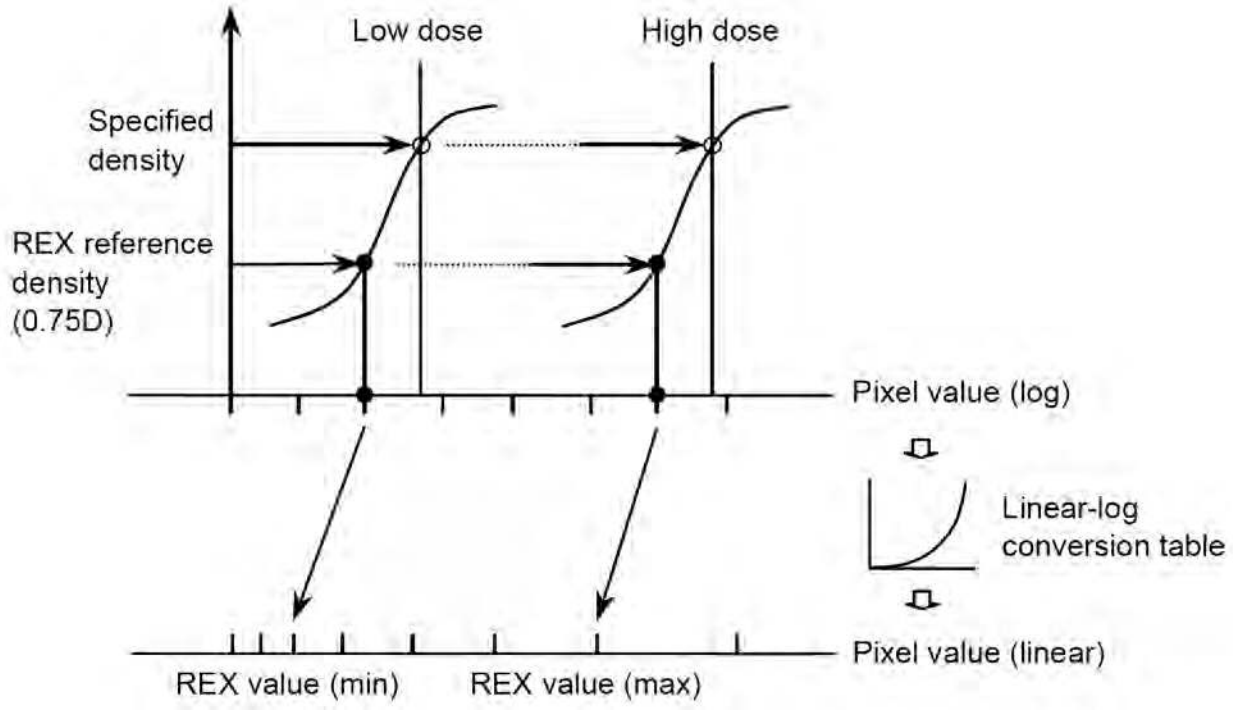

Figure N2.

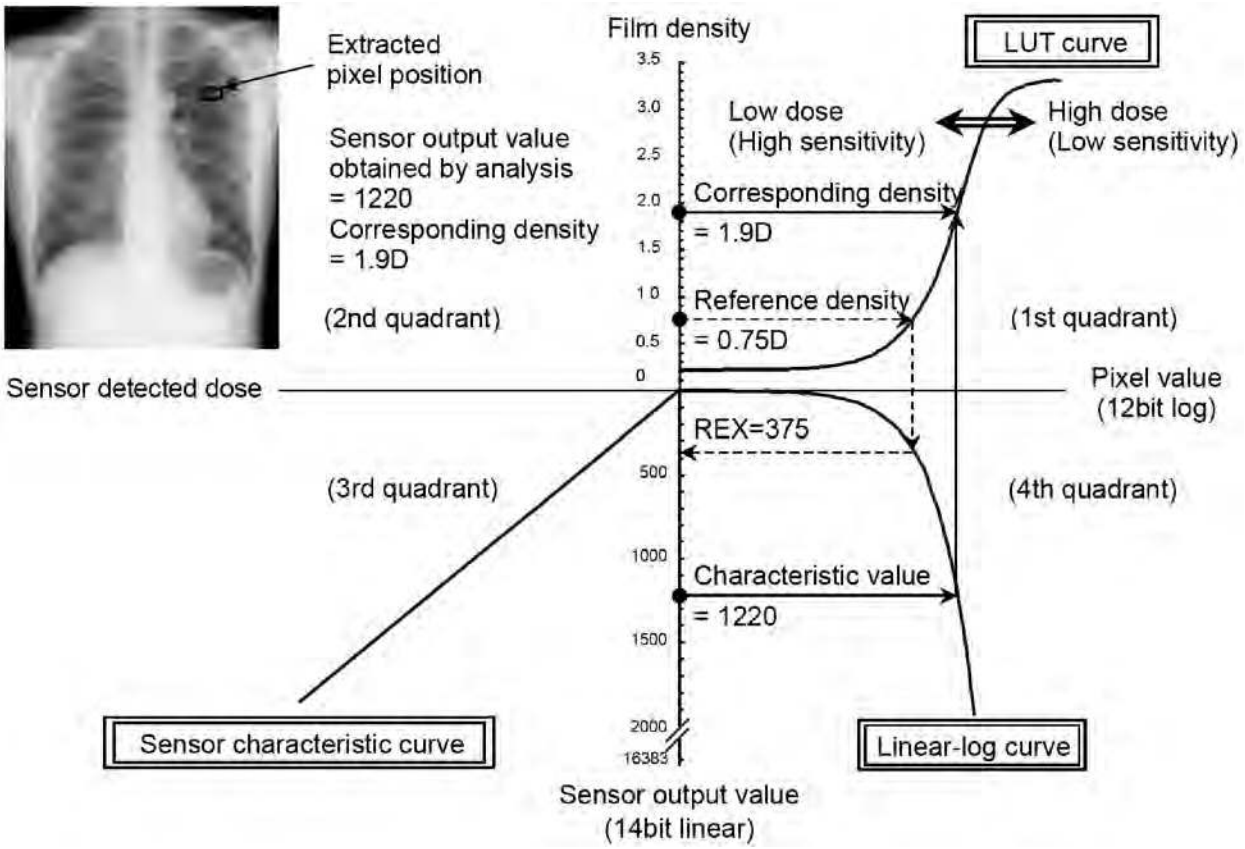

Figure N3.

It is also assumed that the corresponding density showing the film density at the position of this pixel is designated as 1.9D.

According to these assumptions, the LUT curve is shifted horizontally, the amount of shift being determined at the position where the arrow from the feature pixel value and the arrow from the corresponding density intersect.

Next is an explanation of the algorithm by which the REX value is determined.

The REX value is determined at the position where the arrow from the reference density 0.75D (fixed) meets the sensor output axis.

In other words, the REX value denotes sensor output at the position expressed by density 0.75D on the film output by the CXDI.

Next is an explanation of the REX value in proportion to the x-ray dose. Figure N4 shows the change in the REX value when the dose is low.

In Figure N4, it is assumed that the sensor output corresponding to the feature pixel value determined by analysis is 650, the patient exposure dose is 650/1220=0.53 times, and the corresponding density is 1.9D, the same as in Figure N3.

As the dose is small, in this figure the characteristic value is 650.

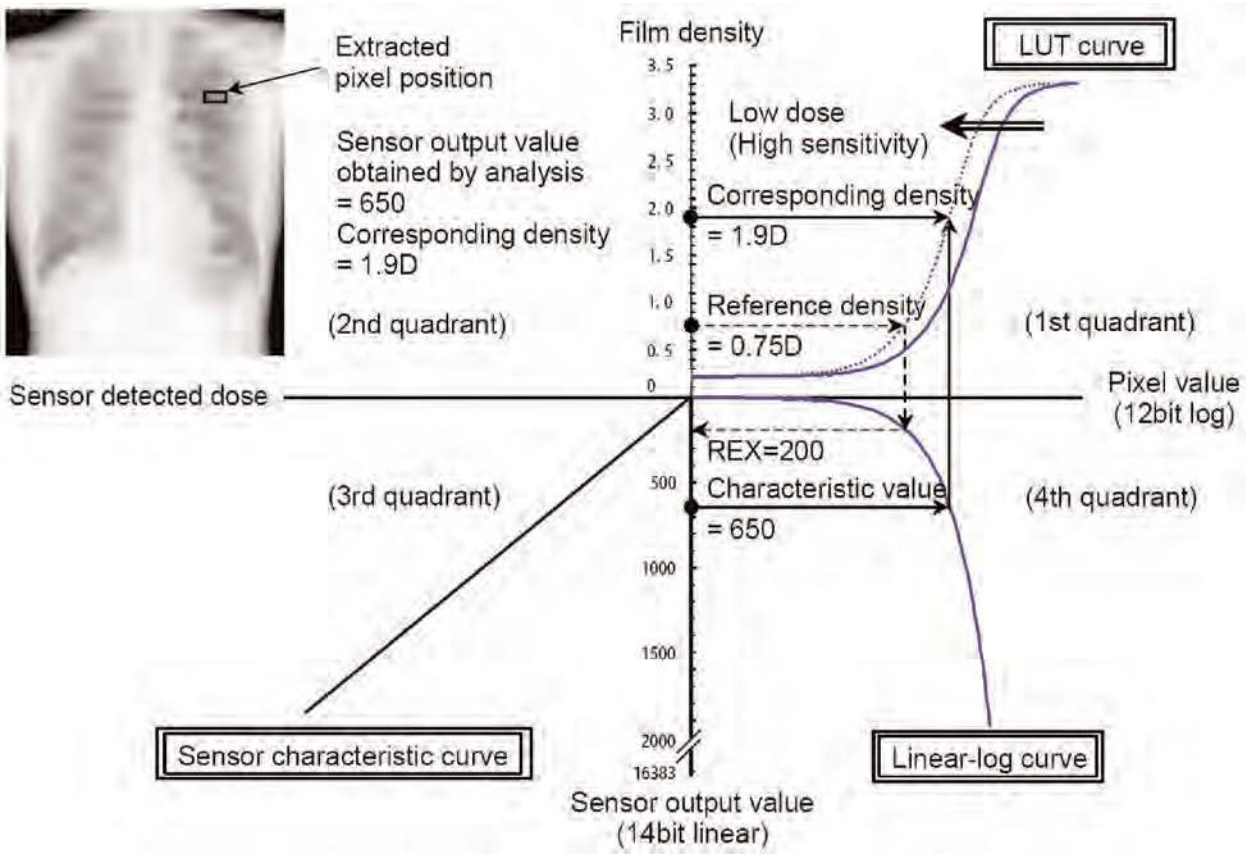

Figure N4.

As the density is constant, the LUT curve moves to the position where the arrow from the sensor output value 650 and the arrow from the corresponding density 1.9D intersect, and is shifted to the position indicated by the dotted line.

Corresponding to this shift in the LUT curve, the REX value decreases to 200.

At this time the relation 1220/650=375/200 is maintained.

As a result, if the shape of the LUT curve is identical and it is only a case of a shift in position, the REX value is in proportion to the radiographic dose.

On the other hand, it is possible that the REX value denotes (the reciprocal of) the system sensitivity.

The REX value is reduced to the extent that the LUT curve shifts to the left of the figure, and the high sensitivity radiography could be said to correspond to the low dose.

In the case of screen/film systems, as the LUT curve does not shift, when the dose is low, the density is simply low.

However, with the CXDI, the system sensitivity changes and the density is always constant.

Consequently, the REX value could be said to be (the reciprocal of) the system sensitivity.

As shown above, the relation between the REX value and the x-ray dose is proportional.

However, although the REX value is the x-ray dose index, it is not the absolute index.

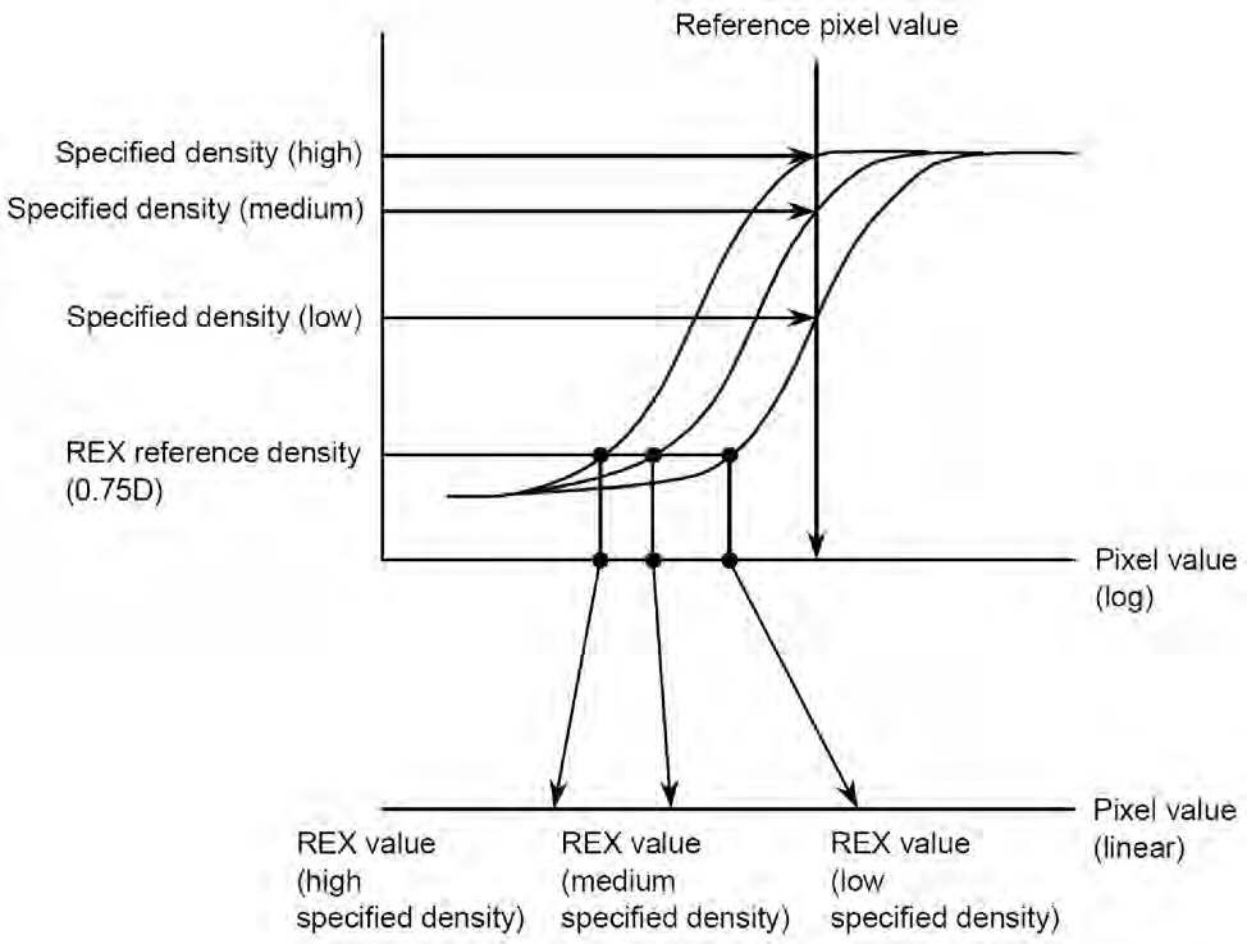

Figure N5.

The REX value assumes that adjustment, including interactive adjustment, is finished and the QA image has been defined.

If the QA image has been defined and the LUT is the same as always, the proportional relation with the x-ray dose will be established, but if the density and contrast change, the REX value also fluctuates.

Figure N5 shows this fluctuation of the REX value due to changes in density.

The REX value decreases as the specified density and the density increase, and the REX value increases as they decrease.

Figure N6 shows the fluctuations in the REX value due to changes in the contrast.

If the contrast is increased when the specified density is higher than the REX reference density, the REX value will increase, and if the contrast is reduced, the REX value will also be reduced.

For the above reason, the REX value fluctuates when the specified density and contrast change, which means that the REX value is determined when all image processing parameters are fixed.

The REX value is only meaningful when adjustment is finished and the QA image is defined.

Furthermore, even if the REX value is moved when the LUT is varied by the doctor or x-ray technician, or according to the x-rayed body part, the actual dose amount cannot be judged simply by comparing the REX value.

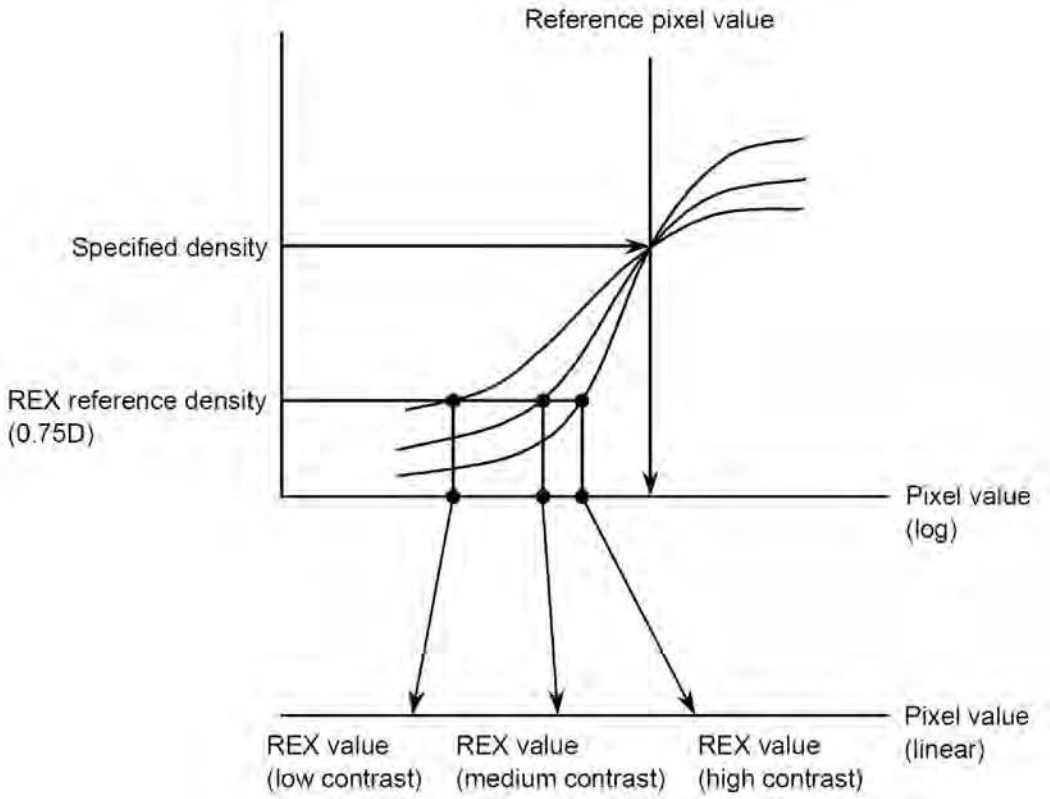

Figure N6.

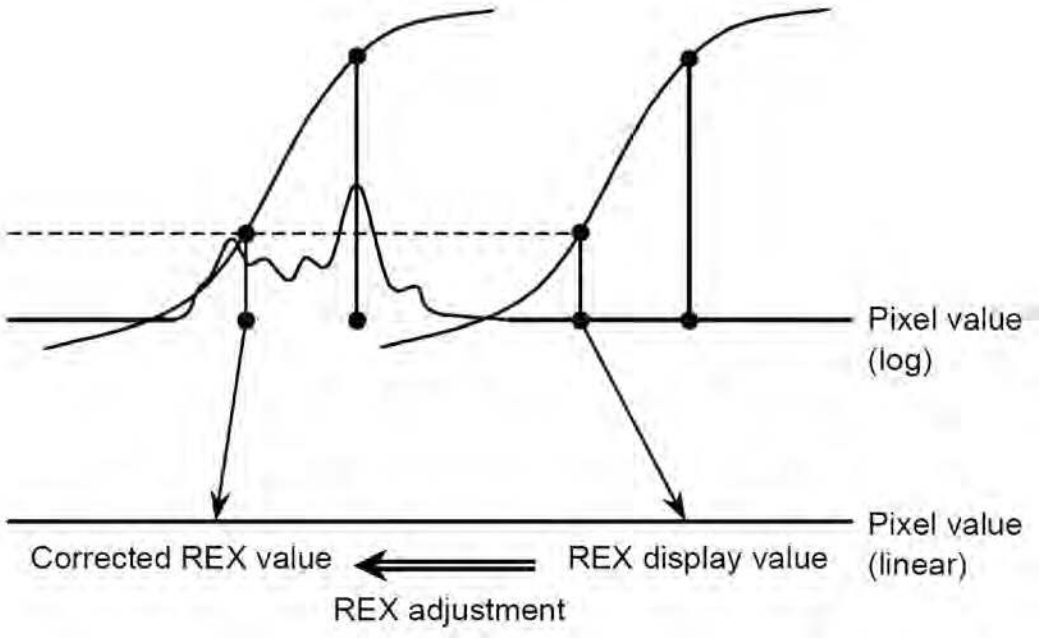

Figure N7.

Finally, the following explains the correction method using the REX value (REX adjustment) in the event of analysis failure.

Figure N7 shows a bright image due to analysis failure.

If the image is white due to analysis failure, the LUT curve has been shifted to the right (to a higher pixel value) in relation to the range in which the image exists.

As for the REX value at this time, large values are displayed.

Consequently, if the REX value is corrected to a smaller value, the LUT curve is shifted to the left to the range of the image and the correct density is achieved.

Alternatively, if the image is black, it can be corrected to the appropriate density by correcting the REX value to a higher value.

Appendix O

SwissVision Dose Indicator

The dose indicator in the SwissVision implementation will be calculated, based on the window center value, provided by the MUSICA processing. This window center value describes the center of area of the logarithmically scaled histogram. The conversion from the linear 12-bit and 16-bit range to the logarithmic scale is calculated as follows:

$$log12bit = 3.2767 + log10(lin12bit/4095) \qquad valid\ range: 0.0 - 3.28$$
$$log16bit = 3.2767 + log10(lin16bit/65535) \qquad valid\ range: 0.0 - 3.28$$

Swissray formula for DI:

$$PixelValuecenter = 10^{(WindowCenter - 3.2767)} * 2^{12}$$

$$DI = round\,(30*(log10(PixelValuecenter\,/2^{16})+3.2767)+1) \qquad if\ PixelValuecenter > 34.3$$

$$= 1 \qquad if\ PixelValuecenter < 34.3$$

Legend:

"WindowCenter"	MUSICA output
DI	Dose Indicator