Recommended Exposure Indicator for Digital Radiography

Report of AAPM Task Group #116

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1) Purpose and Scope

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

For more than a decade, the phenomenon of "Exposure creep" in photostimulable storage phosphor imaging has been reported. (Freedman 1993, Gur 1993, Seibert 1996) 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. Excessive detector exposures produce high quality images with improved noise characteristics but at the expense of increased patient dose. As a result, radiologists tend to complain about under-exposed images but remain silent when images are acquired at higher dose levels. Therefore, 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 or poor images. Consequently, average exposure levels tend to creep up over time if a clear indicator of exposure is not provided.

Techniques required to achieve optimal radiographic imaging in Digital Radiography (DR) are often different than those used for film/screen. In addition, different DR detectors may require different technique factors due to differences in the energy dependence of the detector materials in use. This inconsistency among DR systems may cause confusion and sub-optimal 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 image receptor exposure provided and actively monitored by the imaging system.

A standardized indicator of the exposure incident on a DR receptor that is consistent from manufacturer to manufacturer and model to model is needed. This could be used to monitor differences in exposure between rooms 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.

The purpose of this report is to recommend a standard indicator which reflects the radiation exposure that is incident on an image receptor after every exposure event. 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. This indicator does not represent exposure to a patient.

2) Definition of Terms Used

Digital radiography 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. This section defines terms used in this document that relate to digital radiography processes.

Digital Radiography (DR)

Radiographic imaging technology producing digital projection images such as those using photostimulable storage phosphor (Computed Radiography or CR), amorphous Selenium, amorphous Silicon, CCD, and 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 that results in a specific beam HVL (see section 4 Standardized Radiation Exposure Conditions).

For-processing pixel values (Q)

The image pixel values produced by a DR system after necessary corrections have been applied to the initially recorded raw data [see IEC62220-1 ed. 1 for a complete description of appropriate correction methods]. 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. Geometrical distortion.

The relationship between Q and K_{STD} may vary for different DR systems. Manufacturers are expected 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)

For-processing pixel values, Q, that have been converted to have a specific relation to a standardized radiation exposure (K_{STD}). Using the DR systems relationship between Q and K_{STD}, Q values are converted to Q_K values such that the converted values that have a specific relation to air kerma, $Q_K = 1{,}000*\log_{10}(K_{STD}/K_o)$ when K_{STD} is in microgray units, $K_o = 0.001$ μGy, and $K_{STD} \geq K_o$.

For-presentation image values (Q_P)

For-processing detector values are typically modified by image processing to produce an image with values suitable for display. 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. Detector values suitable for presentation (Q_P) are typically sent to display devices (printers or

workstations) or image archives. NEMA standards, including DICOM Part 14, define these as presentation values, or P-values.

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 detector values in defined regions of an exposure to the detector, in which case, the median value, either $<Q>$ or $<Q_K>$, is converted to the air kerma from a standardized radiation exposure, K_{STD}, that would produce the same detector response. The regions where the median is determined may be defined in different ways (Section 5 Assessment of Detector Response, K_{IND}). The value should be reported in microgray units with 3 significant figures.

Image Values of Interest (VOI)

Pixel values in the original image (Q) that correspond to the primary anatomic region in the recorded image area for a particular body part and anatomical view from which K_{IND} is calculated.

Target Equivalent Air Kerma Value (K_{TGT})

The optimum K_{IND} value that should result from any properly exposed image. 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.

Relative Exposure Factor (f_{REL})

An indicator as to whether the detector response for a specific image, K_{IND}, agrees with $K_{TGT}(b.v)$. Relative exposures are to be reported as $f_{REL}= \log_2(K_{IND}/K_{TGT}(b,v)$ with one significant decimal of precision (i.e. 0.0, 0.6, -1.3 etc.). f_{REL} is intended as an indicator for radiographers and radiologists as to whether the technique used to acquire a radiograph was correct.

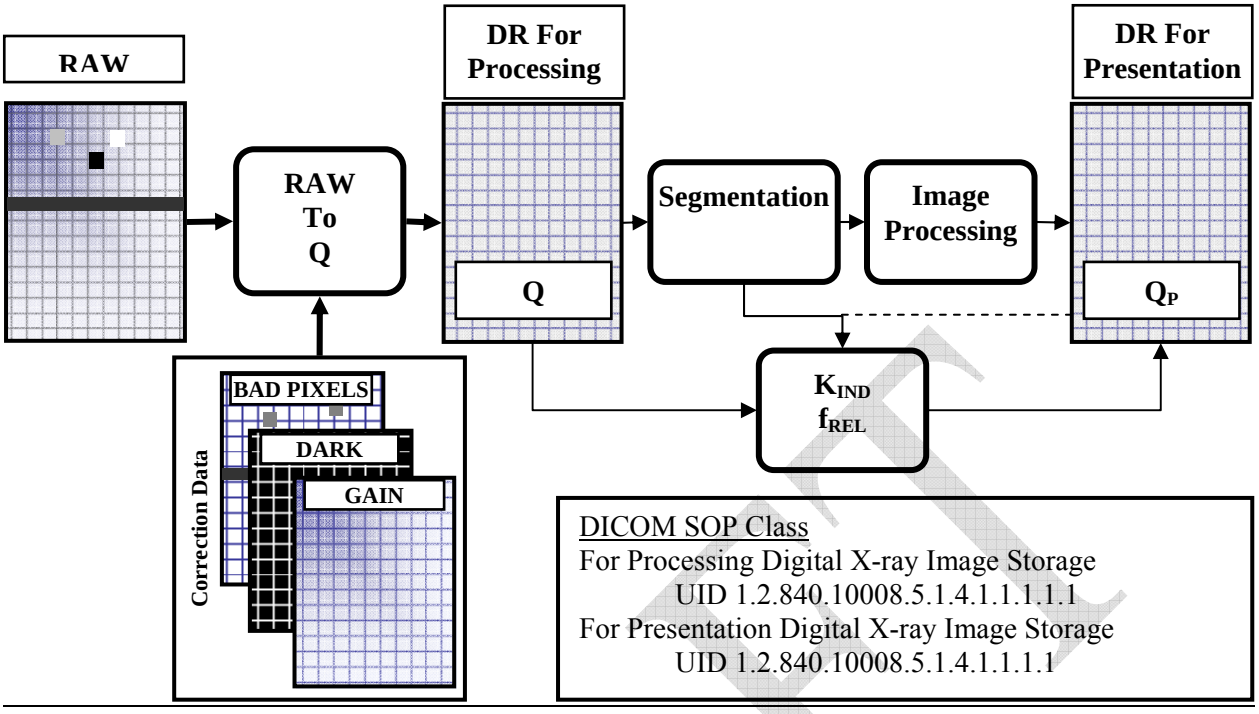

Figure 1: Essential processes in the acquisition of a digital radiograph. K_{IND} and f_{REL} are computed from Q values using segmentation information.

3) Recommendations

This report makes the following specific recommendations regarding indicators of exposure for digital radiography systems:

a) 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 NEMA incorporate a new element for digital radiography 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 3 significant figures.

b) 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 indicator, termed the Relative Exposure Factor (f_{REL}), is to be displayed to the operator of the system and included in the DICOM header. The Relative Exposure Factor should be prominently displayed to the operator of the digital radiography system immediately after every exposure and immediately after any modification of the detected image values of interest, and should be included in the DICOM header of every image in a new element to be added by DICOM which will be a signed decimal value between -9.9 and +9.9 with one significant digit after the decimal.

c) The Indicated Equivalent Air Kerma, K_{IND}, and the Relative Exposure Factor, f_{REL}, are determined from the VOI (see section 5). It is recommended that systems provide display functions to optionally delineate the defined VOI 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.

d) For tests of system performance, 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 part 10 format on a media storage device.

e) The relationship between Q_K values and the standardized radiation exposure incident to the DR receptor 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.

f) 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 and standard deviation of values within graphically defined regions. Small interactively drawn circular or rectangular regions are appropriate for this purpose.

g) For testing of systems, manufacturers should provide methods to remove the anti-scatter grid without otherwise changing the detectors response or provide grid attenuation factors to be used in calibration.

4) Standardized Radiation Exposure Conditions

A uniform field radiation exposure made to the detector of a DR system is used to assess the relation between corrected image values recorded by the detector (Q) and the quantity of radiation incident on the detector. 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 human tissues. This section recommends standardized radiation conditions to be used for this purpose (Table 1). Since DR system response is energy-dependant, it is recommended that two standard beam conditions be defined, one for imaging of the chest at tube potential settings above 100 kV_P and one for all other radiographic images. The conditions for general radiographic systems differ significantly from those for mammography systems. This report addresses only general radiographic and dedicated chest systems.

The IEC has previously made recommendations for standard radiation conditions for use in testing medical diagnostic x-ray systems (IEC 61267). 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 which is impractical for field measurements. For the first edition of IEC 61267, kVp was to be adjusted to achieve a desired beam half value layer (HVL). However, for the second edition, more stringent contraints were place on the beam quality before added filtration rather than allowing kVp adjustments. As a consequence, the conditions recommended in the second edition are applicable only to laboratory facilities.

Instead, TG116 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 (Table 1). The target HVL is intended to be reasonably

close to RQA5 for general radiography and RQA9 for chest radiography. Minor adjustments in indicated kVp and added filtration are permitted to achieve the target beam quality.

Applications	kV_P	Added Filtration	Target HVL	IEC Surrogate
General Radiography	66.5 – 73.5	0.5 mm Cu + (0 - 3.0) mm Al*	6.8 ± 0.2 mm Al*	RQA-5
Dedicated Chest	114 - 126	1.0 mm Cu + (0 - 4.0) mm Al*	11.6 ± 0.3 mm Al*	RQA-9

*** Type 1100**

Table 1

The use of copper as a component of the added filtration is recommended in order to reduce the overall thickness of added material. In a prior publication, 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% (Samei 2001). The additional Al material achieves a HVL near the desired nominal while keeping the copper thickness at a value that is readily available from metal foil suppliers. It is acceptable to substitute brass made from copper and zinc with minimal other impurities. The added Al material should be on the beam exit surface of the Cu so that Cu characteristic radiation is absorbed. 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.

Added filtration with copper as indicated in Table 1 plus 3-4 mm of Aluminum are suitable for x-ray tubes with modest intrinsic filtration. For an x-ray tube spectra with HVL of 2.58 at 70 kVp (RQR5), computational simulations 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 for RQA5, 0.5 mm Cu plus 3 mm Al (type 1100), or 24 cm of muscle. For a tube with HVL of 5.00 at 120 kVp (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 of Al (type 1100).

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 aluminum. The kV_P may also be adjusted, if necessary. Similarly for RQA9, 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 this.

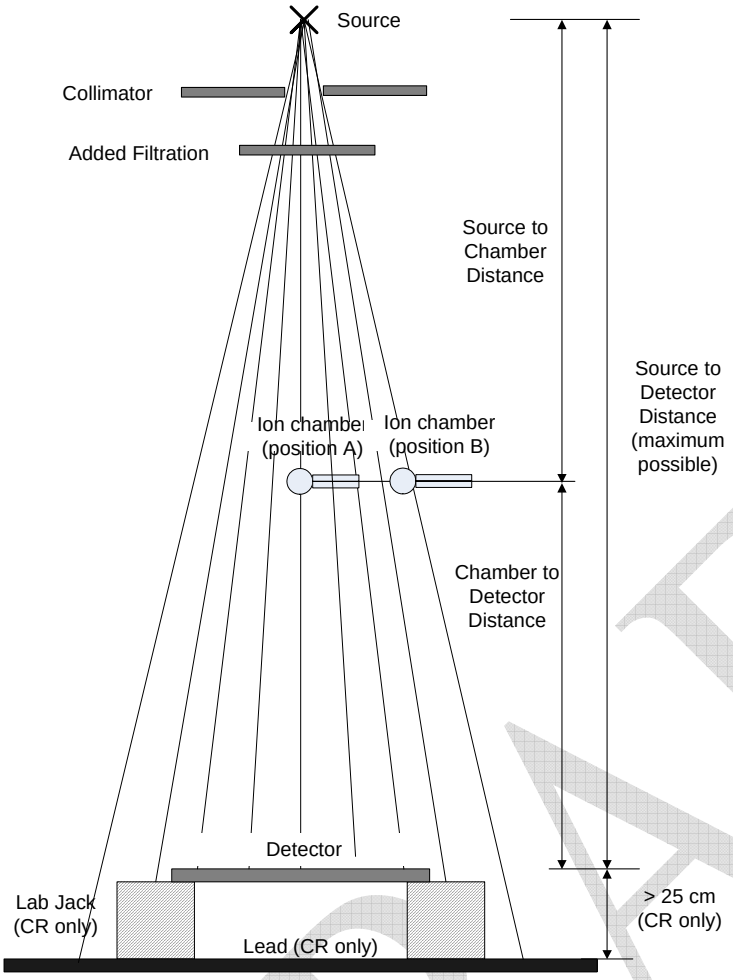

Figure 2

The remainder of this section describes the measurement geometry to be used to determine K_{STD} under the standard radiation exposure conditions which is shown in Figure 2. 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 condition.
2. Add the specified filtration at the face of the collimator (center of range listed in Table 1)
3. The detector should be placed as far from the x-ray source as possible.
    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 2)
    b. If present, remove the anti-scatter grid without otherwise modifying the response of the detector. If the grid cannot be removed, obtain the grid attenuation factor from the DR system or grid vendor.

c. If the detector is not square, the long axis of the detector should be perpendicular to the x-ray tube A-C axis.

4. Place a calibrated ion chamber at the center of the beam approximately midway between the source and detector (see Position A in Figure 2). The distances should be measured to the center of the chamber and to the surface to the detector. The distance to the x-ray source, to the center of the ion chamber and the surface of the detector must be accurately known. If the distance from the detector housing surface to the detector is not labeled consult the manufacturer for this measurement.
5. Collimate the x-ray beam to only cover the ion chamber with no more than 1 inch margins.
6. If desired, the HVL of the beam can be measured at this point and the kVp or added filtration adjusted to obtain a value based on the specifications in Table 1. The detector should be covered with a lead apron or similar barrier when making the exposures for HVL determination and adjustment.
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, changing the mAs setting to obtain the desired air kerma at 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 2).
9. 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.
10. Make an exposure using the mAs found in step 6 above and 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.

5) 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}), ie. 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 4 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>). Rectangular or circular regions having an area 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.

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. A median operator is specified so that the median of image values can be computed and transformed using the known relationship between for-processing image pixel values (Q) and exposure. 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 easily computed 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 to the VOI in the recorded image. The VOI 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 sub-diaphragmatic portions of the image would be secondary regions. However, the mediastinum would be a primary region for a thoracic spine radiograph. Hence the VOI's for the AP 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 VOI is 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 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 phototimer cell.

More expert scene recognition algorithms may be used to identify the VOI. 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 VOI. Additionally, methods to manually adjust the VOI after the automated VOI recognition 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 a recent report (Van Metter, 2006), it has been suggested that the 'for-presentation' image pixel values (Q_P) be used to define the region for computation of K_{IND}. 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}. 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 to region used to compute K_{IND}, its value should reflect any changes to the VOI that are made by the operator.

This report does not make recommendations as to how the VOI is 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. It is recognized that the defined region from which K_{IND} is computed has strong influence on the result. With further effort, it is hoped that a consistent method can be recommended in the future.

6) Reporting Relative Exposure Factor (f_{REL})

The Indicated Equivalent Air Kerma, K_{IND}, is an indicator of the receptor response in regions where anatomically important tissues have been recorded by a DR receptor. 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 produce the same receptor response. For this reason, it is referred to as an 'equivalent' air kerma. Generally, the actual incident exposure required to produce the same receptor response will vary if kV_P is varied when a radiograph is made of a specific object. For the general radiography standard beam conditions, the incident exposure required for the same receptor response in a typical DR receptor varies modestly for kV_P values in the range from 55 to 90.

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 receptor response as for high energy x-rays. Thus the variation in signal-to-noise ratio for KVp values between 55 and 90 is sufficiently small to make K_{IND} an effective indicator of image quality with respect to the recorded signal-to-noise ratio. Above 90 kV_P, the K_{IND} should be determined relative to a standard beam with higher average energy to maintain a consistent relationship between SNR and the indicator.

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 indicated K_{IND} in relation to that targeted for the body part and view of the radiograph.

As defined in section 2, f_{REL} is to be expressed as:

$$f_{REL} = \log_2(K_{IND}/K_{TGT}(b,v)),$$

where $K_{TGT}(b,v)$ is the targeted value for body part b and view v.

f_{REL} is intended to be an indication to persons performing or interpreting radiographic examinations whether the signal-to-noise ratio in the VOI is considered acceptable. 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 TG 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 1 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 2 would provide appropriate information in terms of both direction (over- or under-exposure indicated by a positive or negative value, respectively) and magnitude (+1 is double the intended exposure, -1 is half the intended exposure) on needed technique corrections.

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 a higher detective quantum efficiency. 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.

a) "Speed"

The definition of radiographic speed according to ISO 9236-1 is the radiation exposure required to achieve a net optical density of 1.0 on the developed film. With digital radiography there is no fixed relationship between the radiation exposure and the resultant density in the image. With film-screen receptors a change in speed affects the spatial resolution properties of the receptor. This same relationship does not hold true with digital image receptors since sharpness is independent of the amount of exposure used to acquire the digital image.

Several manufacturers currently use an exposure indicator which parallels the concept of "speed" or "speed class" used by film manufacturers (See Appendix). In addition, many manufacturers and users have become accustomed to referencing their systems as functioning within a given speed class. This has created some misunderstandings and scientific inaccuracies which have been discussed in the literature (Huda, 2005). TG116 recommends avoiding the concept of "speed class" when referring to DR system performance. K_{TGT} values should be used to describe how one system may vary from another with respect to radiographs of a particular body part and view.

The characterization of a digital radiographic system as being a given speed class may give the false indication that it should always be operated at a specific exposure level. 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 ALARA concept should prevail in that the minimum amount of exposure should be used to achieve the necessary diagnostic information (Willis and Slovis, 2004). Using the speed class characterization for given digital imaging systems may increase the possibility that ALARA is violated for some imaging tasks.

For DR systems, the appropriate incident exposure is a variable based on the desired signal-to-noise ratio rather than on the resulting optical density of a radiograph. To emphasize this important difference, it is recommended that speed not be used to describe the recordings from a DR system. Rather, the K_{TGT} values should be used to describe how one system may vary from another with respect to radiographs of a particular body part and view.

7) Clinical Use of the Relative Exposure Factor (F_{Rel})

The clinical use of the Relative Exposure indicator is essentially the same as that of film optical density: it serves as an indicator of proper radiographic exposure technique. For film/screen 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 significantly lower resolution (1024x1024 or less), lower brightness and capable of rendering fewer grey levels than those to be used for diagnostic reading. These workstations are also rarely calibrated to DICOM PS3.14. As 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 reader. The Relative Exposure indicator can be used clinically to ensure that the amount of radiation delivered to the detector is appropriate for a given imaging task.

a) Exposure Indicator and Radiographic Techniques

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

f_{REL} 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. Once $K_{TGT}(b,v)$ levels are set, it is useful to identify several types of "control limits" on f_{REL}: 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 the primary determinant of when a repeated film is needed, the reason for repeating a digital image is primarily noise-related. What would be a significantly underexposed film image may be of adequate diagnostic value in digital form. Since this judgment depends upon the diagnostic task, it is appropriate to seek consultation with a radiologist for certain ranges of f_{REL}-indicated under- and over-exposure prior to repeating. 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. Any significant deviation for the established target range should require management oversight to determine the cause for the deviation and implement appropriate corrective action such as re-training, re-calibration of the equipment, or re-assessment of the target value.

To be effective, care must be taken assure that appropriate targets and limits are posted and the radiographers are educated and periodically re-educated as to their meaning.

f_{REL}	Range Action
> +1.0	Excessive patient radiation exposure: repeat only if relevant anatomy is "burned out", require immediate management follow-up
+0.5 to + 1.0	Overexposure: repeat only if "burnout"
-0.5 to +0.5	Target range
Less than -0.5	Underexposed: consult radiologist for repeat
Less than -1.0	Repeat

Table 2: Exposure Indicator f_{REL} Control Limits for Clinical Images

Note that the example f_{REL} "control limits" for DR repeats in Table 2 may be considerably broader than those tolerable for some film-based exams such as chests. For example, consider a film with an average gradient of about 2.5 and a tolerable density range of $\pm$ 0.3 OD. Using the relationship:

$$\Delta OD = \gamma \, LOG_{10}(E_2/E_1) = 2.5 \, LOG_{10}(E_2/E_1),$$

a $\pm$ 0.4 f_{REL} target range would correspond to an exposure range of

$$\Delta f_{REL} = 0.8 = Log_2(K_{IND,max}/K_{TGT}(b,v)) - Log_2(K_{IND,min}/K_{TGT}(b,v))$$

$$= Log_2(K_{IND,max}/K_{IND,min})$$

$$(K_{IND,max}/K_{IND,min}) = E_2/E_1 = 1.7 \; (+/- \; 16\%)$$

and an optical density range of

$$\Delta OD = 2.5 \, LOG_{10}(1.7)$$

$$= 0.3 \text{ OD}$$

which could easily push parts of a image into the toe or shoulder of the film's response.

This has also been investigated by VanMetter and Yorkston (1996) for chest and abdominal imaging for a wide range of patient thicknesses under controlled experimental conditions. Their data shows that for a very limited data set taken under highly controlled conditions, most (but not all) AEC controlled images for chest and abdomen are expected to fall within the range of $f_{REL} = \pm 0.4$.

Operators should be instructed that high f_{REL} values are associated with excessive radiation dose but have good image quality with respect to noise. Tighter limits on f_{REL} 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).

b) 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 film/screen imaging, despite the wide dynamic range of DR. Regardless of receptor 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 receptor. Like film/screen systems, digital receptors have significant energy dependence, which in general differs from that of the AEC sensors. Depending on design and calibration of the AEC, the result can be digital image levels that vary substantially from the desired level.

A well-designed AEC should be capable of modifying required receptor 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 signal-to-noise ratio (Christodoulou et al, 2000). Assuming that AEC performance is evaluated under clinically relevant conditions which can be simulated by various thicknesses of acrylic and kV_P's ranging from 60 to 120 (Hendee and Rossi, 1979). The K_{IND} can serve as the indicator of image signal level for this purpose, just as optical density did for film.

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 than 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 film/screen 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 film/screen systems and may allow for energy compensation appropriate for film/screen. 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 be higher for AEC-based exposures at lower kV_P's, because the AEC compensation intended for rare-earth film/screen systems significantly overcorrects for lower kV_P's (Goldman, 2004). If this is the case, $K_{TGT}(b,v)$ values for f_{REL} may need to be adjusted upward to appropriately reflect this energy dependence.

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, +/-0.20 optical density units, Wilkinson and Heggie, 1997). Certainly, the much narrower latitude of film/screen 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.

c) Inappropriate clinical use of f_{REL}

A final note regarding f_{REL}'s and clinical techniques: even if images being produced clinically have corresponding f_{REL}'s well with the target range, the clinical techniques used may still not be appropriate. One can just as readily achieve an acceptable f_{REL} for an AP L-spine view with 65

kV_P as with 85 kV_P; evidence of under-penetration 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 may tend to raise or lower f_{REL}'s (depending on the exam and projection) and perhaps hide inappropriate technique. It is essential that all aspects of good clinical technique be adhered to with digital imaging, and an appropriate f_{REL} should not be interpreted as proof of good work.

Overexposed images should not be repeated unless parts of the anatomy of interest are "burned out" or "clipped" (i.e., exposure levels saturated the dynamic range of the digital detector system).

8) Recommend 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 improve the overall quality of the imaging operation.

For instance, section 3c calls for an overlay that graphically illustrates the pixels in a given image which have been used to calculate $\bar{Q}_{FP}$. This would provide a very quick method of determining that the automated VOI-recognition and 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 VOI min and max overlaid on it showing the minimum and maximum Q values used for $\bar{Q}_{FP}$ determination. Finally, there are many clever ways to indicate the f_{REL} for every image using a sliding bar or color coded tool with position and or color linked to the magnitude of f_{REL}.

Other highly desirable features are logs of the f_{REL} values and reasons for rejected and repeated films stored on the system along with anatomical view selection and technique factor information for every image. Software to analyze this log to assist with process improvement by identifying potential problem exams, problems with equipment, and technologists in need of continuing education is also invaluable to the user community.

As already mentioned in Section 5, systems could 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.

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.

9) Application to Mammography, Veterinary and Dental Radiography

Digital mammography, veterinary and dental radiography can all potentially benefit from a universal exposure indicator for the same reasons one is needed for DR applications. Digital radiography 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. 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: Current Status of Exposure Indices

A variety of exposure indicators have been provided by manufacturers of digital radiography systems. Some of these are summarized in Table 3, which illustrates the wide variation in terms, units, mathematical form, and calibration conditions of exposure indicators. Inconsistency among manufacturers is presently the primary drawback for clinical use of exposure indices. 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.

Tabs 1-11 to this appendix presents detailed descriptions of exposure indices provided by some of the digital radiography vendors.

The use of exposure indicators began with the cassette-based 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 indictors 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 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 image receptor and, indirectly, the appropriateness of selected technique factors. Not only was this useful to the technologist when setting technique factors but, from a more global perspective, it allowed hospitals to analyze overall exposure trends (Willis et al., SPIE). QC programs based on exposure indicator monitoring have been shown to moderate exposure in actual clinical practice (Seibert, Academic Radiology). 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 six manufacturers are presented in the Tabs: 1. Agfa; 2. Fuji; 3. Kodak; 4. Konica 8. Alara and 12. ICRco.

Fuji's "Sensitivity" or "S-number" is the oldest exposure indicator. This index 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 image receptor 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.(Huda, Radiology 2005) Accurate interpretation of the S-number is limited without knowledge of the value of "Latitude" or "L-number" for the particular image (Chotas and Ravin, Investigative Radiology. 1992) . Approximately two-and-one-half 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 two in exposure to the receptor. 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. Each image is assigned a user-selected “speed class” which 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 two difference in the exposure to the receptor. Therefore, lgM can be considered to be expressed in units of "bels".

For the most part cassette-less 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 image receptor. 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, 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 image receptor.

Of those cassette-less DR systems utilizing detector exposure indicators, 4 are presented in Tabs: 5. Imaging Dynamics; 6. Philips; 7. GE Healthcare and 10. 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 Fuji. Unlike the Fuji approach, Philips conforms to the ISO-9236-1 convention for speed. The Philips EI also differs from Fuji 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 region-of-interest (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 under-exposure and positive values indicate over-exposure. The absolute value represents the deviation from the target exposure by factors of two.

Canon introduced a cassette-based DR system for retrofitting existing x-ray generators. As such, the receptor 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 roughly 100 per mR, but the value is a function of the "brightness" and "contrast" selected by the operator. By admonishing the technologists against

modifying brightness and contrast, REX has been demonstrated to have utility in oversight of exposure factor (Arreola and Rill, 2004).

GE delayed introduction of a detector exposure indicator, instead using DAP for patient dose estimates as described above. However, on its most recent announced cassette-less DR product, GE incorporates three additional parameters indicating receptor 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. (Goldman 2004) The exposure conditions differ drastically among the manufacturers primarily with regards 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 kVP and beam energy variations (Tucker and Rezentes, 1997). 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 receptor during clinical use.

Other limitation shared by the various exposure indices is that of sensitivity to the mathematical processes used to identify collimation boundaries and segmentation of the anatomically relevant data. The determination of the ROI 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 cassette-less 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 dose-area product to an ideal patient, which is of no use in managing image quality and only satisfies certain regulatory requirements.

The remainder of this section contains a more detailed description of some of the approaches that have been developed by the various manufacturers.

Table 3. DR Exposure Indicators, Units, and Calibration Conditions (adapted from Willis CE. Strategies for dose reduction in ordinary radiographic examinations using CR and DR. Pediatric Radiology 34(Suppl 3):S196-S200, 2004)

Manufacturer	***Indicator Name***	***Symbol***	***Units***	***Exposure Dependence***	***Calibration Conditions***
Fuji	Sensitivity Number	S number	Unitless	$200/S \propto X$ (mR)	1 mR at 80 kVP 3mm Al HVL => S=200
Kodak	Exposure Index	EI	Mbels	EI + 300 = 2X	1 mR at 80 kVP + 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 kVP + 1.5 mm Cu =>lgM=2.96
Konica	Sensitivity Number	S value	Unitless	for QR=k, $200/S \propto X$ (mR)	for QR=200, 1 mR at 80 kVP => 200
Canon	Reached Exposure Value	REX	Unitless	for Brightness=c_1, Contrast = c_2, $REX \propto X$ (mR)	for Brightness = 16, Contrast = 10, 1 mR => 106 (?)
GE	Dose Area Product	DAP	$dGy\text{-}cm^2$		
GE	Entrance Skin Exposure	ESE	mGy		
GE	Detector Exposure Index	DEI	Unitless		
Hologic	Exam Factor, Center of Mass of log E histogram				
Hologic	Dose Area Product	DAP			
Hologic	Accumulated Dose				
SwissRay	none				
Imaging Dynamics Corporation	log of Median of histogram				
Imaging Dynamics Corporation	f#			$2^{f\#} = X_{relative}$	
Philips	Kerma Area Product	KAP			
Philips	Exposure Index	EI	unitless	$100/S \propto X$ (mR)	
Siemens	Exposure Indext	EI	µ Gy Air KERMA	X(µ Gy)=EI/100	RQA5, 70kV, 21 mmAl, HVL=6.8 mm Al

Tab 1: 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 S/F system requires an average detector dose of 2.5μGy 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:

$$\lg M = 1.9607 + \log\left(\frac{Dose(\mu Gy)}{2.5}\right) + \log\left(\frac{SpeedClass}{400}\right).$$

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 kVP, 1.5 mm Cu).

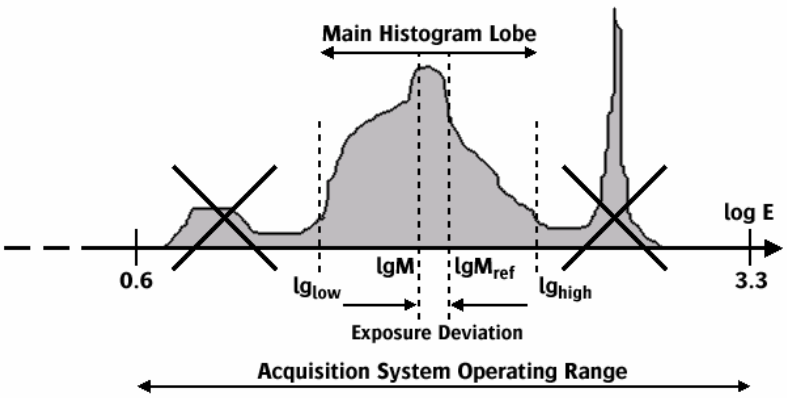

Figure 3 Schematic drawing of the histogram of a typical radiographic image

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 a 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 3). 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.

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 fifty exposures in each exam category so that radiologists, radiology administrators or physicists can monitor exposure consistency and investigate/correct any occasional or systematic deviations.

Tab 2: Fuji FCR

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 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 1. 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: $\mathbf{S = 4 \times 10^{(4-S_k)}}$, 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) ≅ 200 / *S*** for standard x-ray beam conditions (80 kVP, ~3.0 mm Al HVL). 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 image histogram with the above-mentioned parameters is illustrated in Figure 4. A 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).

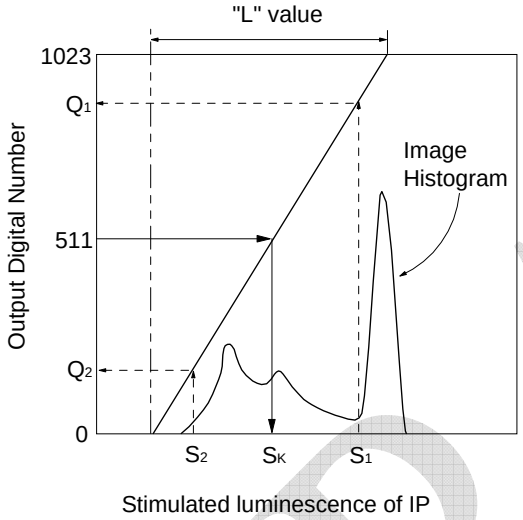

Figure 4. Sensitivity and Latitude numbers defined for the Fuji PSP system output parameters as related to the image histogram

Tab 3: Kodak CR

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 image receptor. 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 kVp with a 0.5 mm copper and 1.0 mm aluminum 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 kVp[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

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

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 milli-Roentgens an alternate expression

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

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

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 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: kVp (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).

[1] Ehsan Samei, J. Anthony Seibert, Charles E. Willis, Michael J. Flynn, Eugene Mah and Kevin L. Junck, "Performance evaluation of computed radiography systems," Med. Phys. **28**, 361-71 (2001).

X-ray Spectrum Dependence of Exposure Index

The response of digital radiography 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

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

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

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

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 1.

Table 1. 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 2** and **Table 3**, 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 2. Kodak CR systems (GP-25 cassette) - FOR-PROCESSING IMAGE pixel values versus incident exposure.

Air Kerma (microGy)	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 3. Kodak DR systems - FOR-PROCESSING IMAGE pixel values versus incident exposure.

Air Kerma (microGy)	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 1 for the Kodak CR systems and in Figure 2 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.

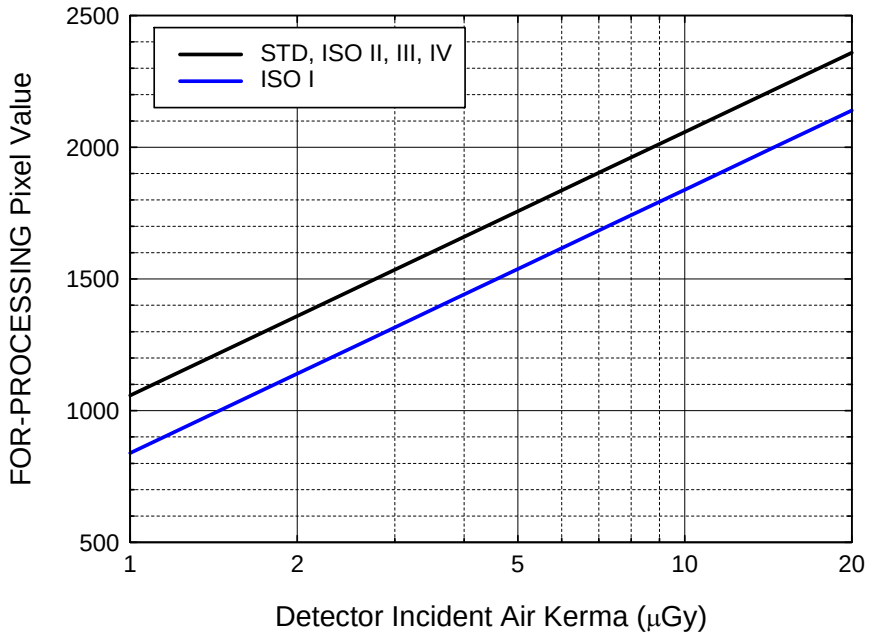

Figure 1. Kodak CR systems (GP-25 cassette) - FOR-PROCESSING IMAGE pixel values versus incident exposure.

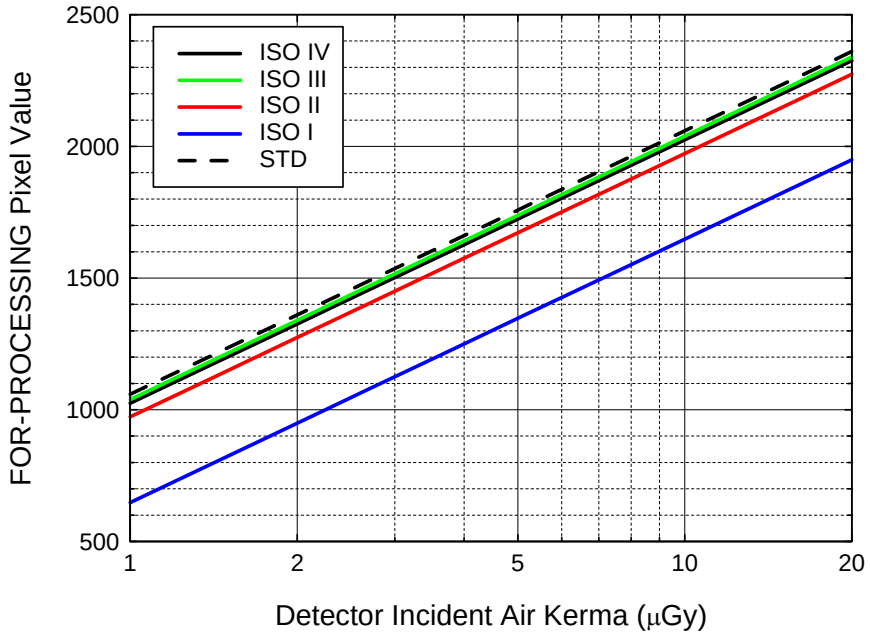

Figure 2. Kodak DR systems - FOR-PROCESSING IMAGE pixel values versus incident exposure.

Tab 4: Konica Minolta CR

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

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

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 80kV, 2.0 m).

The QR parameter is related to an equivalent film/screen 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:

$$X = R = 200/QR \quad \text{(b)}$$

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 film/screen image optical density of 1.2 in the desired region This S-Value is then defined as:

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

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 1mR.

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.

Tab 5: 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 1mR will produce a mid range optical density on a 200 speed film/screen system. An exposure of 0.5mR will produce the same optical density on a 400-speed film/screen 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 film/screen system speed that would have responded in a similar fashion.

The first requirement is a calibration of the digital systems response to input radiation. To emulate the typical exit spectra from a patient's body, we harden the beam by adding one millimeter of copper filtration. Measurements are then made at 80kVP to determine the system input /output characteristic in terms of milliroentgens per digital number (mR/DN). This input / output ratio, RIO, 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, IMed, 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_{IO} and the median value $I_{Med,}$ we calculate:

$$X_{med} = I_{med} \cdot R_{10}$$

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 film /screen 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 SE to a target speed ST. 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 film / screen 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 two 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 two will give a value of –1, by a factor of four 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 judgement 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, we believe that 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.

Tab 6: Philips Digital Diagnost

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

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

$$RD = 1000 / K \qquad (1)$$

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:

$$K = PV_c / SENS \qquad (2)$$

The sensitivity of the flat-panel detector after applying the standard detector-specific corrections is SENS = 207 μGy^{-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 Eq. (1) and (2) are thus rounded to the values …,100, 125, 160, 200, 250, 320, 400, …(see Table 4). Each step corresponds 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 RD 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, e.g., also used for the grading of the mAs scale on many X-ray generators.

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 Eq. (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:

1. The determination of an sub-area (ROI) of the full image, containing the target area;
2. The determination of the characteristic pixel value in this ROI. This can be the average or the median pixel in this sub-area; 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 LUT. Since the LUT adaptation may be dependent on the selected image type (examination/anatomy), the characteristic pixel value and such the RD 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 RD-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 PVc 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 RD will correspond to different exposure values (air kerma values) for different kilovoltages. This effect is most pronounced for low kVP, 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 RD calculation in the DigitalDiagnost (starting with release 1.2), which compensates for changes in the sensitivity (see Fig. 5).

Table 4: 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
				20
5.10	44.67	9291	23808	
				25
4.05	35.48	7380	23208	
				32
3.22	28.18	5862	22608	
				40
2.56	22.39	4657	22008	
				50
2.03	17.78	3699	21408	
				63
1.61	14.13	2938	20808	
				80
1.28	11.22	2334	20208	
				100
1.02	8.91	1854	19608	
				125
0.81	7.08	1473	19008	
				160
0.64	5.62	1170	18408	
				200
0.51	4.47	929	17808	
				250
0.41	3.55	738	17208	
				320
0.32	2.82	586	16608	
				400
0.26	2.24	466	16008	
				500
0.20	1.78	370	15408	
				630
0.16	1.41	294	14808	
				800
0.13	1.12	233	14208	
				1000
0.10	0.89	185	13608	
				1250
0.081	0.71	147	13008	
				1600
0.064	0.56	117	12408	
				2000
0.051	0.45	93	11808	
				2500
0.041	0.35	74	11208	
				3200
0.03	0.28	59	10608	
				4000
0.03	0.22	47	10008	
				5000
0.02	0.18	37	9408	
				6300
0.02	0.14	29	8808	
				8000

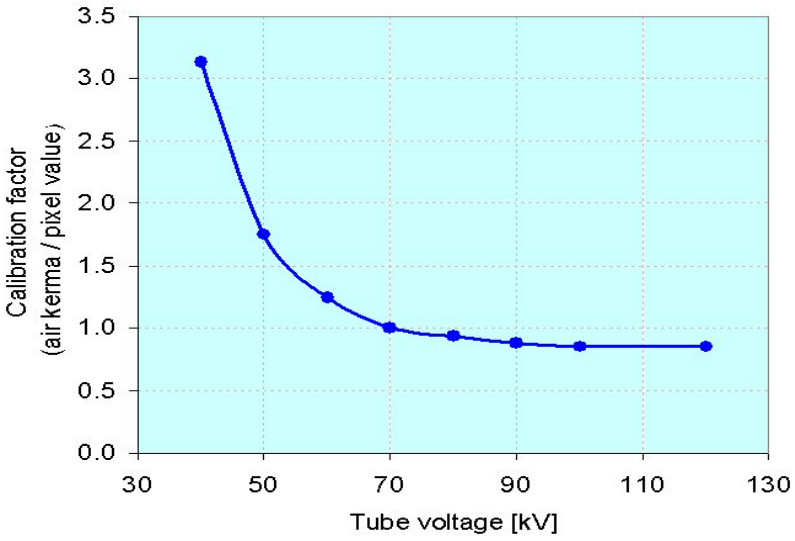

Figure 5: kV correction factor for the Exposure Index as used in the DigitalDiagnost.

Tab 7: GE Healthcare

The GE flat-panel digital detector systems do not report an exposure or sensitivity index to the operator today. A "Sensitivity" value is embedded in the DICOM header which is related to exposure at the detector through the original pixel value. For images other than PA and Lateral Chest, 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. Future systems (Definium) will include an index which is described here.

Detector Exposure Metrics

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**)

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.

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)

1) 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 kVP (with a standard filtration and no anti-scatter grid).
- Compensated Detector Sensitivity, CSens, is equal to USens after compensation for kVP, presence of anti-scatter grid, and user-selected additional collimator Cu filtration.

2) 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

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

Hence, DEI can be effectively used for indicating

(a) Over/under-exposure due to patient mis-positioning or incorrect selection of ion chambers with AEC acquisitions, and
(b) Over/under-exposure due to inappropriate technique selection with fixed-time acquisitions.
(c) Over/under-exposure due to other operator-related or system-related events.

Over a reasonable clinical technique range, DEI values should be more consistent than CDExp values for a given anatomical view.

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.

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 re-processing. 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.

Tab 8. Alara CR

Table 5 DR Exposure Indicators, Units, and Calibration Conditions

Manufacturer	*Indicator Name*	*Symbol*	*Units*	*Exposure Depend.*	*Calibration Conditions*
Alara	Exposure Indicator Value	EIV	mbels	EIV+300=2X	1 mR @ SC200 (RQA5) =>2000

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 film-screen 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 6.

Table 6. 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 1 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 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:

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

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.

Tab 9: Canon

No official response from the vendor at the time of this writing.

Tab 10: Siemens Medical Systems

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

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	The exposed field is subdivided in a 3 x 3 matrix, where

	index	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 of 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 [μGy] = c x 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.
5.7	Precision of the exposure index	Scaling: EXI-values are scaled as the original 14bit

	in constancy tests (and at constancy conditions)	pixel values. (quod vide 5.5 and 5.6)
6	What statements are possible with the exposure index in the medical image and what conditions 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	What 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.

Tab 11: SwissRay

No official response from the vendor at the time of this writing

Tab 12: 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 film/screen 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 photomultiplier tubes 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 approach

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 which 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 o performance relative to an expected anatomy-specific exposure. This is not too different from the number generated by the method developed by Imaging Dynamics Corporation (IDC). For example, in IDC's approach, an exposure of -2 is a 4 times under exposure, -1 is 2 times under exposure, 0 is perfectly exposed, +1 is 2 times over exposed, and +2 is 4 times over exposed.

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 discrete number ranging from -2, -1, 0, +1, +2, see Figure 1. The goal for the user is to keep the images in the neutral range (around 0) independent of the anatomical view.

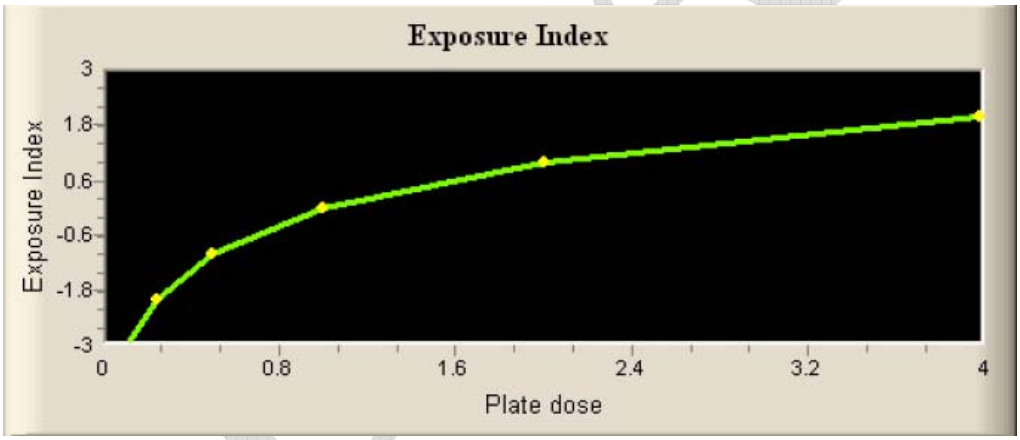

Figure 6. Relative exposure index as a function of plate dose in mR.

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 7 below). If the level is outside the -2 or +2 range the number is shown in red.

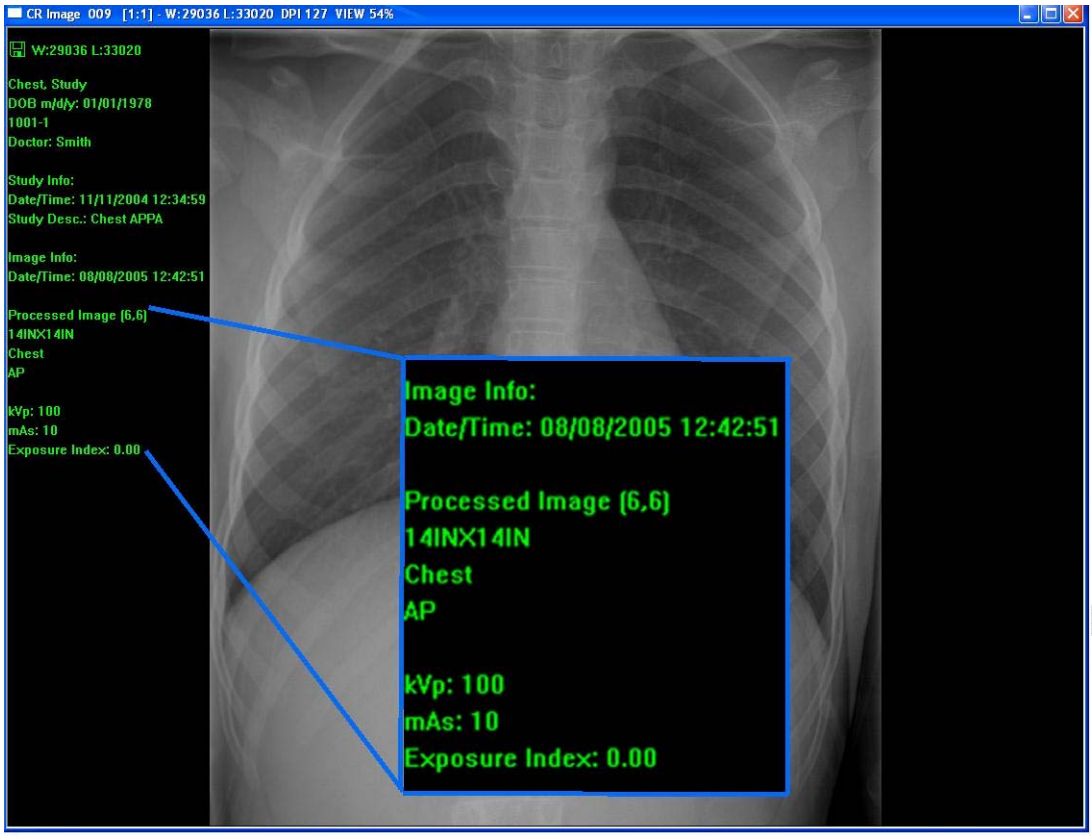

Figure 7. Magnified view of display of exposure index. For values in the -2.0 to +2.0 range the Exposure Index is displayed in green; if outside the range, the value is displayed in red to signal to the technician that the exposure is out of range.

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 which 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 kVp source, hardened with 1.5mm of Copper at the x-ray source. The pixel analysis is performed on an area corresponding to 80% of a 14" by 17" cassette. The software performs a regression by which the linearity of the system is evaluated as a function of dose, as shown in Figure 8 below.

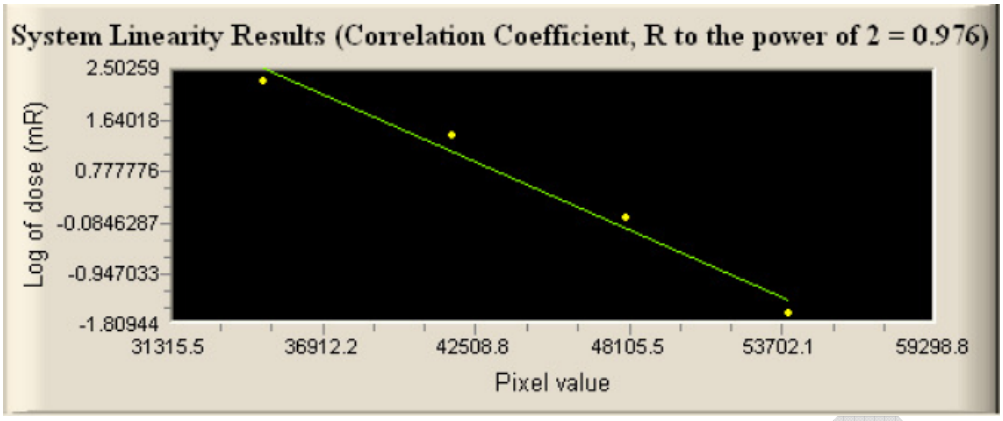

Figure 8. Linearity of the exposure index is shown as a function of log of dose and its corresponding pixel value.

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 Fuji S Number and Agfa IgM. Translation into other methodologies can be implemented as well.

Other References

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