Digital Radiography Image Quality: Image Acquisition

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This article on digital radiography image acquisition is the first of two articles written as part of an intersociety effort to establish image quality standards for digital and computed radiography. The topic of the other paper is digital radiography image processing and display. The articles were developed collaboratively by the ACR, the American Association of Physicists in Medicine, and the Society for Imaging Informatics in Medicine. Increasingly, medical imaging and patient information are being managed using digital data during acquisition, transmission, storage, display, interpretation, and consultation. Data management during each of these operations has a direct impact on the quality of patient care. These articles describe what is known to improve image quality for digital and computed radiography and make recommendations on optimal acquisition, processing, and display. The practice of digital radiography is a rapidly evolving technology that will require the timely revision of any guidelines and standards. This document provides a basis for the technologies available today in clinical practice and may be useful in guiding the future clinical practice of digital radiography.

Key Words: Digital radiography, image quality, image acquisition

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INTRODUCTION AND DEFINITIONS

Computed radiography (CR) and *digital radiography* (DR) are the commonly used terms for digital radiographic detectors. Computed radiography uses a photostimulable storage phosphor that stores the latent image with subsequent processing using a stimulating laser beam and can be easily adapted to a cassette-based system analogous to that used in screen-film (SF) radiography. Historically, DR has been used to describe a digital x-ray imaging system that reads the transmitted x-ray signal immediately after exposure with the detector in place. There are several types of detectors that can be classified as DR systems, including automated (cassetteless) CR systems. In addition, some DR systems are adapted to a cassette-based x-ray system. Thus, the historical nomenclature becomes less accurate as technology advances because distinct classification into the two broad categories of CR and DR is no longer possible. More appropriate is the distinction based on cassette vs cassetteless operation. In this article, digital radiography is used to refer to all types of digital radiographic systems, including both those historically termed CR and those historically termed DR.

There are many ways to categorize the current state-of-the-art digital radiography technology. One categorization considers (1) form factor, (2) image acquisition time, and (3) x-ray signal conversion methodology. The concept of cassette-based vs cassetteless operation is defined using the term *form factor*. A cassette-based digital detector uses the SF paradigm, which allows the use of existing imaging modality infrastructure and provides excellent positioning flexibility. On the other hand, the labor-intensive handling of cassettes and the need to wait for the image, often with batch-mode processing, leads to a loss of time efficiency. Cassetteless operation indicates the ability to acquire the x-ray signal and, without subsequent user intervention, produce an image at a local

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workstation for review and manipulation. By *image acquisition time*, a distinction is made regarding the ability to acquire an image nearly instantaneously over the full field of view for a large area detector vs the sequential, longer scan time acquisition of a slot-scan device. *X-ray signal conversion* refers to the method by which the x-rays are converted to the output signal of interest and is usually described as being either an indirect or direct radiation detection scheme. All digital detectors produce an output signal, usually in the form of electrons, which represent a quantity of charge that corresponds to the number of x-rays absorbed in a given detector element (del) in the detector. The charge is then converted to a digital value for storage in the image matrix. *Indirect* refers to the conversion of x-rays into secondary information carriers, such as light via a scintillator, before conversion to charge. *Direct* refers to the conversion of x-rays into electron-hole pairs and the direct collection of charge using semiconductors as the x-ray converter, without a secondary conversion event. There are advantages and disadvantages that can be ascribed to each of the digital detectors in these broad categories, and they determine the best solution for a given clinical task or imaging application.

This article is applicable to the practice of cassette-based and cassetteless digital radiography. It defines motivations, equipment guidelines, specifications of data manipulation and management, and quality control and quality improvement procedures for the use of digital radiography that should result in high quality radiologic patient care. It also includes a brief review of the differences between digital radiography and SF radiography in terms of imaging characteristics such as sensitivity, dynamic range, noise, and spatial resolution. A brief overview of current receptor technologies available for the clinical practice of digital radiography also is included. A glossary of commonly used terminology is included in the companion paper on image processing and display.

Motivations for using digital radiography in clinical practice include, but are not limited to:

1. the significantly larger range of x-ray intensities that can be imaged by digital receptors compared with analog systems;
2. the independence of displayed contrast from the x-ray tube potential setting (kVp) through adjustment of the display window width;
3. the independence of displayed brightness from milliampere-second (mAs) setting through adjustment of the display window level;
4. the availability of image processing and computer-aided diagnostic algorithms for image enhancement and analysis;
5. the easier and more reliable generation of accurately labeled and identified image data;
6. the ability to electronically transmit data to an appropriate storage medium from which it can be electronically retrieved for display for formal interpretation, review, and consultation; and
7. the ability to transmit data to and from remote sites for consultation, review, or formal interpretation.

Components of the performance of high-quality digital radiography include, but are not limited to:

1. the development of validated imaging protocols so that the consistency of image quality and radiation dose can be established and maintained between rooms and between sites;
2. the use of appropriate compression of image data to facilitate transmission or storage, without loss of clinically significant information;
3. the archiving of data to maintain accurate patient medical records in a form that may be retrieved in a timely fashion;
4. the ability to retrieve data from available prior imaging studies to be displayed for comparison with a current study;
5. the ability to apply image processing for better display of acquired information;
6. adherence to applicable facility, state, and federal regulations;
7. maintenance of patient confidentiality;
8. minimization of the occurrence of poor image quality;
9. minimization of the delivery of inappropriate ionizing radiation dose to patients; and
10. the promotion of clinical efficiency and continuous quality improvement.

GENERAL REQUIREMENTS

Image acquisition should be performed in accordance with the appropriate ACR modality or examination guideline or standard. At the time of acquisition, the system must have capabilities for capturing demographic as well as imaging information such as accession number, patient name, identification number, date and time of examination, name of facility or institution of acquisition, type of examination, patient or anatomic part orientation (eg, right, left, superior, inferior), amount and method of data compression, and display of the total number of images acquired in the study. This information (except for the number of images acquired) must be associated with the images when transmitted. These fields should be formatted according to the Digital Imaging and Communications in Medicine (DICOM) standard.

The ability to record patient date of birth and gender and a brief patient history is recommended. The ability to record peak kilovoltage (kVp), tube current (mA), exposure time, beam filtration, and radiation exposure indicator is imperative, particularly for cassette and cassetteless digital radiography systems that have a direct interface to the x-ray generator information. The use of the DICOM DX object is recommended instead of the more limited CR object for digital radiography. It is also recommended that gray-scale presentation state objects be used to transmit annotations, shutter, and display lookup tables (LUTs).

COMPARISON OF SCREEN-FILM AND DIGITAL RECEPTORS

There are several fundamental differences between SF systems and digital (CR or DR) systems in terms of the physical processes involved in image acquisition. The different processes introduce different constraints on the factors determining image quality, such as spatial resolution, contrast, and noise. Here we briefly describe the basic operation of each system and then discuss the most important acquisition-related factors affecting image quality for each.

Screen-Film

In SF radiography, an x-ray photon absorbed in the screen deposits energy, which is converted into the production of multiple visible light photons that travel through the screen and in turn deposit energy in the film grains in the emulsion layers of the film. The image information is stored in the film emulsion during an exposure through the accumulation of exposed grains. Regions of the film with a greater density of exposed grains have higher optical density and appear dark after film processing and thus correspond to regions of greater x-ray absorption in the screen. However, the relationship between optical density and the number of absorbed x-rays is nonlinear. For analog SF radiography, the acquisition, display, and storage of the image are intrinsically nonseparable. This means that acquisition techniques must be used that result in both acceptable film darkening while maximizing image contrast. Because the range of acceptable optical density is limited by view box luminance and the response of the human visual system, maximization of image contrast occurs at the expense of exposure dynamic range. This is referred to as the exposure latitude, and it is quite narrow in SF radiography compared with digital radiography.

Limitations in SF radiography include narrow exposure latitude, the need for chemical processing, inefficient mechanical handling, incompatibility with electronic transmission and image enhancement, and higher costs for film materials and labor. The chief advantages are high spatial resolution, consistency of image appearance, and the accumulated experience of radiologists, medical physicists, and technologists in utilizing SF radiography in an optimal manner.

Screen-film images can be converted to a digital form using a film digitizer to produce an equivalent digital image. The characteristics of the output digitized image are chiefly those of the SF image, with its inherent advantages and disadvantages as listed above. Two major digitizer types are based on laser point scanning and optical line transmission scanning, using a light collection guide or an array of charge-coupled device (CCD) photosensors, respectively [1]. In general, these digitizers provide good gray-scale rendition, with the laser systems typically having more accurate conversion of high optical density values and the CCD array scanners able to have a more compact, higher resolution scanner capability. Film digitizers should output the DICOM radiographic imaging information object descriptor with patient and acquisition method demographic data and must meet the minimum spatial frequency resolution requirements of 2.5 lp/mm at the film plane for conventional digital radiography images. In general, film digitizers introduce additional noise in the digital image, especially in regions of high optical density.

Computed Radiography

In photostimulable storage phosphor (CR) systems, x-ray photons are absorbed in the storage phosphor, also known as the imaging plate (IP). Unlike conventional screens, the deposition of x-ray energy in storage phosphors results in the storage of a portion of the energy in highly localized, metastable areas called f-centers, which serve as energy wells. During an x-ray exposure, the image is built up in the phosphor through the accumulation of f-centers. After the exposure, plate readers scan a (red) laser beam over the surface of the IP. In one implementation the laser beam sweeps across the IP (scan direction) and in combination with IP translation (subscan direction) the beam extracts the latent image from the IP. The plate reading time depends on the size of the detector and the scan speed of the reader. Some newer readers use a line-scan approach (the laser illumination on the plate is a line) rather than a point-scan approach to increase readout speed, but the principle is the same in both approaches. At each location, the energy of the incident laser photons triggers a deexcitation of the f-centers, resulting in the prompt emission of (indigo) light photons. A portion of these photons is detected by a photosensitive device (usually a photomultiplier tube for point-scan readers or a linear solid-state photodiode array for line-scan readers), whose output is then digitized and stored. Although the amount of photostimulated light detected

is proportional to the number of f-centers and thus to the number of absorbed x-rays at that location (ie, the relationship between the photomultiplier tube signal and the number of absorbed x-rays is linear), nonlinear amplification of the photomultiplier tube signal before digitization is often used. The resulting raw pixel values are subsequently processed for display using a combination of segmentation, rescaling, and filtering algorithms. The size of the image matrix is a function of the IP dimensions (18 × 24 cm, 24 × 30 cm, 35 × 35 cm, 35 × 43 cm, and custom sizes) and the pixel sampling pitch (typically 100 to 200 μm in general radiography). Output image sizes vary between 8 and 16 MB for radiography with a typical matrix size of approximately 2,000 × 2,500 pixels. A detailed description of technical issues associated with CR imaging is given in the report of American Association of Physicists in Medicine Task Group 10 [2].

DR Systems

DR systems encompass a number of different technologies that are rapidly evolving. At this time, the majority of DR systems use thin-film transistor (TFT) arrays, commonly known as flat-panel arrays. Systems based on CCDs are available for general-purpose and chest radiography, using scanned, slot-shaped detectors [3-5]; a single CCD; or tiled arrays of CCDs optically coupled to larger area x-ray-to-light converters. Thin-film transistor arrays are composed of a matrix of discrete dels, each of which contains a transistor. The transistors operate as gates, permitting an electric charge to flow through them only when they are turned on. During an x-ray exposure, the gates are turned off, and the image is built up in the dels in the form of an electric charge, with the amount of charge in each del proportional to the number of x-rays absorbed in that region of the detector (again, a linear relationship). The means by which x-ray energy is converted to a stored electrical charge varies somewhat between manufacturers but can be broadly classified according to whether it involves an intermediate conversion of x-ray energy to visible photons (indirect flat-panel devices) or not (direct flat-panel devices). In an indirect device, an x-ray-to-light converter, similar to that used in SF imaging, is placed in contact with the TFT array. Each del of the TFT array contains a light sensor (a photodiode) to convert the fluorescent light to a stored electric charge. In a direct device, a layer of material is deposited directly onto the TFT array. When an x-ray photon is absorbed in this photoconductor material, an electric charge is generated and collected in the dels of the TFT array. For both indirect and direct flat-panel detectors, after the x-ray exposure, the TFT array gates are turned on one row at a time, and the amount of charge stored in each del of that row is transferred through drain lines to a row of charge amplifiers at the edge of the array for digitization and storage. The entire detector is read out by successively turning on all of the rows in a sequential fashion and storing the digital data in corresponding locations in the output digital image matrix.

Advantages and Limitations of Digital Radiography

The main limitations of CR and DR are higher initial cost (especially for DR), a lack of familiarity on the part of both radiologists and technologists with electronic image display and with online soft copy reading (compared with alternator-based batch-mode reading), and the lack of consistent feedback to technologists concerning the use of optimal acquisition techniques. The latter problem, along with the much larger dynamic range of digital systems, has led to a gradual increase in patient radiation dose, an issue discussed in more detail below [6-8]. The advantages of CR and DR include the separation of acquisition, display, and archiving, allowing tremendous flexibility using image-processing functions such as those that adjust the level (analogous to the brightness) and window width (analogous to the contrast) of the image gray-scale presentation. However, display contrast is limited by the inherent image signal-to-noise ratio (SNR), because as the signal contrast is increased, so is the visibility of noise. Other advantages include: anatomy-specific presentation and disease-specific algorithms; in most cases better x-ray detection efficiency and higher detective quantum efficiency (DQE), permitting lower doses to patients; the ability to use a second computer reader to assist the radiologist; a reduction in the number of image retakes due to underexposure or overexposure; and the elimination of labor-intensive handling and distribution of images during the acquisition process.

FACTORS AFFECTING IMAGE QUALITY

Just as with SF radiography, a number of factors affect the quality of the image in digital radiography. Contrast, detail, and noise are the primary factors associated with image quality, and they play a major role in CR and DR. Some additional factors are the result of the digital nature of the process, and they are discussed below.

Detection Efficiency

For SF, CR, and DR, the efficiency with which incident x-rays are absorbed is determined by the absorber thickness, density, and composition. Efficiency can be increased by increasing material density or absorber thickness (usually at the expense of spatial resolution) or through the incorporation of materials having atomic numbers that provide a good match between the x-ray

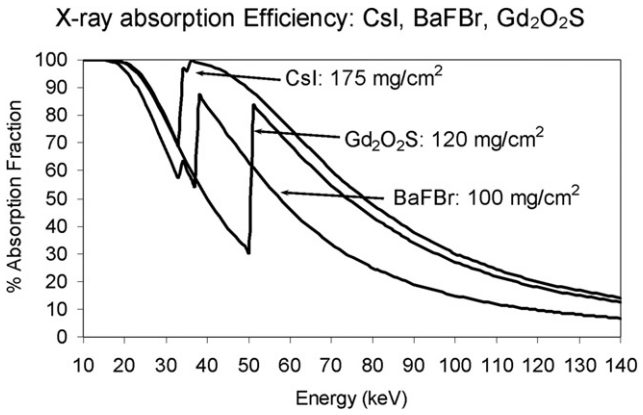

Fig 1. The photon absorption fractions for computed radiography and rare-earth x-ray phosphors are plotted as a function of energy. Phosphor thicknesses are representative of a standard 400-speed conventional screen, a standard resolution computed radiography phosphor detector (100 mg/cm^2), and a cesium iodide phosphor commonly used in indirect thin-film transistor array and optically coupled charge-coupled device camera digital radiography systems [2].

spectrum exiting the patient and the absorption characteristics of the material. Figure 1 shows the absorption efficiency of 3 materials commonly used as x-ray scintillators, plotted against incident x-ray energy [2].

Dynamic Range

In digital chest radiography, the ratio of the maximum to minimum x-ray exposures incident on the detector surface can be greater than 100:1 [9]. For high-quality digital radiography, the receptor must be able to maintain good contrast resolution over this wide dynamic range. The dynamic range of an x-ray imaging system is the ratio of the largest and smallest input x-ray intensities that can be imaged. For all systems, the smallest useful intensity is determined by the intrinsic system noise. The signal must be large enough to exceed this noise, combined with the x-ray quantum noise. The largest intensity is determined by receptor saturation. The dynamic range of SF systems, often referred to as the latitude, is determined by the loss of contrast at low and high exposure levels (the toe and shoulder, respectively, of the H&D curve). For SF systems, the useful dynamic range is dependent on the screen and film characteristics, ranging from approximately 10:1 to 100:1 [10]. By comparison, the response curve of CR is linear (the photostimulated light output is directly proportional to x-ray exposure) over approximately 4 orders of magnitude (a ratio of 10,000:1) [2]. Figure 2 compares the response curves for a 400-speed SF system with that of a typical CR system [2]. The dynamic range of DR systems is determined at the low end by the system noise and at the high end by the charge-holding capacity of the dels. When this capacity is filled, the detector must be read out to remove the charge before any subsequent exposure. The dynamic range of typical TFT arrays used in general radiography, with del sizes of 150 to 200 μm, is approximately 3 to 4 orders of magnitude (a ratio of 1,000:1 to 10,000:1) [11].

Spatial Sampling

All digital detectors sample the continuously varying x-ray fluence at their input at discrete locations separated by some interval called the sampling pitch. In CR systems, the sampling pitch is the distance between adjacent laser beam positions during plate readout. In DR systems, the pitch is the center-to-center spacing between dels. The spatial frequency at which sampling occurs determines how well high-frequency fluctuations in x-ray fluence (those that occur over short distances in the imaging plane) are imaged by the digital system. The sampling frequency of a system with sampling pitch p is $1/p$. For example, if the pitch is 100 μm (0.1 mm), the sampling frequency is 10 per millimeter (10 mm^{-1}). The highest spatial frequency in the incident x-ray fluence that can be reliably imaged (the Nyquist frequency) is half the sampling frequency, or $1/(2p)$. Thus, a system with 100-μm pitch can image frequencies up to 5 cycles per millimeter. If the x-ray fluence incident on the receptor contains information at frequencies higher than the Nyquist frequency, and the presampling modulation

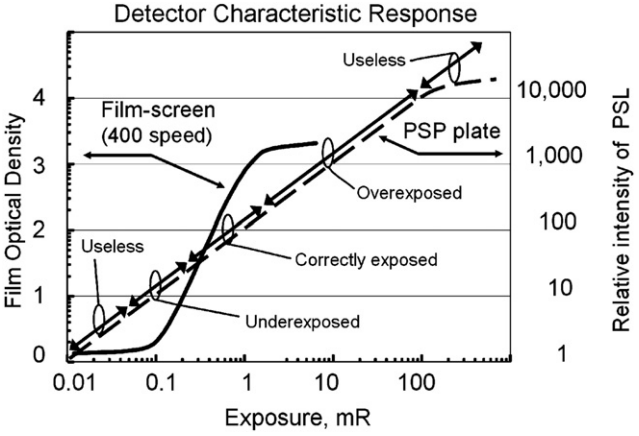

Fig 2. The characteristic response of a 400-speed rare-earth screen-film (solid S-shaped curve) and the computed radiography photostimulable phosphor (PSP) detector (dashed curve) are compared. Exposure ranges indicated by double arrows roughly indicate the exposure ranges characterized as underexposed, correct, or overexposed. Useless areas depict system responses that do not contain information useful for diagnosis because of either excessive quantum noise at the low-dose end of the scale or saturation of the PSP detector [2].

transfer function (MTF) is nonvanishing at these frequencies, the result is false image signals occurring at low frequencies. This phenomenon, called aliasing, produces large-scale artifacts in images and reduces the ability to see fine details in the images.

Spatial Resolution

Spatial resolution is the ability of an imaging system to allow two adjacent structures to be visualized as being separate, or the distinctness of an edge in the image (ie, sharpness). Spatial resolution losses occur because of blurring caused by geometric factors (eg, the size of the x-ray tube focal spot, light diffusion in the receptor), del effective aperture size, and motion of the patient relative to the x-ray source and image receptor. The measurement of spatial resolution is performed by subjective or objective methods. Subjective measurement is achieved with a bar pattern of alternating radio-opaque bars and radiolucent spaces of equal width, imaged to determine the limiting resolution in line pairs per unit of distance, usually expressed in units of line pairs per millimeter. Intrinsic detector resolution measurements are performed by fixing the bar pattern to the receptor surface to eliminate focal spot blurring. System resolution (including the effects of the focal spot blur) uses a bar pattern placed at a clinically relevant distance above the receptor. For digital systems, the resolution may be different in the row and column directions, often requiring separate evaluations. The limiting resolution is the frequency at which the bars and spaces can no longer be visualized. Objective measurements by the MTF are obtained by measuring the transfer of signal amplitude (contrast) of sinusoidal patterns (of various frequencies) from incident x-rays to the output. The system MTF is determined by the product of the individual MTF components along the signal chain. The system MTF can be measured by imaging a test object containing a narrow slit or a sharp edge. For most digital radiography examinations (except possibly digital fluoroscopy), a limiting system spatial resolution of at least 2.5 mm^{-1} is essential and higher resolution (e.g. 5.0 mm^{-1}) is desirable for specialized applications.

In SF imaging, spatial resolution is determined primarily by the thickness of the screen, because the light produced when an x-ray photon is absorbed is emitted in all directions from that point. Thicker screens provide more distance for the light to spread before reaching the film and have poorer spatial resolution. The film has very high resolution and is not responsible for spatial resolution degradation. A conventional 400-speed SF system for general radiography has a spatial resolution limit (the smallest visible set of bars in a bar pattern image) of approximately 7 lp/mm (7 mm^{-1}). In mammography, in which very thin screens are used, the limiting resolution is often greater than 15 mm^{-1}.

In CR imaging, the primary source of spatial resolution loss is scattering of the laser light beam during image plate readout. Laser light scatter results in the deexcitation of locations in the phosphor that are somewhat larger than the size of the laser beam and potentially larger than the separation between laser positions. As a result, the actual blur extends beyond the pixel size. Because thicker phosphors result in a greater scattering depth and more blur, CR systems have been introduced that permit the plate to be read from both sides [12]. Also, CR systems with a structured crystalline geometry allow a greater thickness and enhanced detection efficiency without the corresponding loss of spatial resolution [13].

Spatial resolution in DR systems depends primarily on two factors. The first factor for indirect systems is the spread of light photons in the x-ray-to-light conversion process, and this can be important. This results in blurring similar to that described above because of the spread of photons within the screen of an SF system. To minimize visible photon spread, several manufacturers of indirect DR systems use structured converters, in which the converter material (usually cesium iodide) is formed into narrow, parallel columnar structures, oriented so that the x-ray photons are incident along the long dimension of the columns. The advantage of this structure is that the majority of the light exiting the converter onto the TFT array has been internally reflected along the length of the columns. The internal reflection process thus confines the light photons to a region close to the actual location of the x-ray absorption. This approach permits improved absorption efficiency through thicker absorbers (longer columns) without as much spatial resolution degradation as seen in unstructured converters constructed using the same thickness. Direct-conversion DR systems do not suffer from this effect, because the spread of the electrons within the photoconductor material as they are accelerated toward the TFT array is minimal.

The second factor affecting spatial resolution in DR is the size of the del itself. Because all x-rays absorbed within a single del during an exposure contribute to a single quantity (the summed charge read from that del), there is no way to distinguish different absorption locations within a del. Therefore, structures in the patient smaller than the del size are smeared out, and their contrast is reduced (this is known as the partial volume effect). Such structures may thus be undetectable unless they are inherently high contrast objects (eg, a microcalcification smaller than a del may be recognized as a calcification because its attenuation properties are so different from the other tissue in the del).

Noise

In radiography, noise can be defined as any fluctuations in an image that do not correspond to variations in the

x-ray attenuation of the object being imaged. Image noise is typically measured by illuminating the receptor with a uniform x-ray fluence, then measuring the variance (the square of the standard deviation) in selected regions of the resulting image. A more informative measure of noise can be obtained by estimating the noise power spectrum (NPS), which characterizes the spatial frequency dependence of the noise [14-16]. Knowledge of the frequency response of noise in an imaging system is important because there are a number of additional noise sources in digital radiography, such as aliasing and electronic noise, that are not present in SF systems.

Ideally, image noise is dominated by the x-ray quantum noise. However, all image receptors contain some internal sources of noise. Random noise (noise not correlated with particular locations on the receptor) typically cannot be corrected and adds to the random noise of the x-ray quanta. Examples of random noise include film granularity noise in SF and electronic noise in CR and DR. Internal noise that has a fixed correlation to locations on the receptor is called fixed-pattern noise. Sources of fixed-pattern noise include spatial variation in screen thickness in SF, position-dependent light collection efficiency in CR plate readers, and variations among preamplifier gains in DR. One advantage of digital imaging is that fixed-pattern noise can be largely eliminated through digital postprocessing. This is typically performed by obtaining a set of nominally identical images using a uniform x-ray flood field. Averaging the flood images results in a single image in which the magnitude of the random noise is small compared with the magnitude of the fixed-pattern noise. This image is stored and used as a map to remove the fixed-pattern noise from all subsequent images in a process called flat fielding. Detector artifacts are common in digital systems as a result of spatial variations in the sensitivity of the receptor, causing the image to have structure that is unrelated to the tissue being imaged. Bad dels, typically aligned in columns and rows, may occur during the manufacturing process. The locations of such point and line defects are typically identified and stored, and adjacent neighbor response values are averaged and used as substitute values for the missing data. The number and proximity of detector defects that is allowable without affecting image quality remains to be specified.

Digitization of the analog detector output voltage to form discrete pixel values introduces a type of noise called quantization noise. Digital imaging systems use an analog-to-digital converter to sample and quantize the image data into discrete integer (gray-scale) values, with the range of values determined by the number of electronic binary (on/off) channels within the ADC, also known as bits. The maximum number of gray-scale values that can be encoded is equal to $2^{\#}$, where # is the number of bits. For example, an 8-bit analog-to-digital converter encodes 2^8 or 256 discrete digital values (0-255) over the signal amplitude range (these values are often called analog-to-digital units). An insufficient number of quantization steps will introduce signal-encoding errors that increase the quantization noise. Digital detectors for projection radiography typically use from 10 to 14 bits (1,024 to 16,384 unique analog-to-digital units) in the output image to keep quantization noise small. Some systems use nonlinear (eg, logarithmic) amplification before digitization, which can further reduce the quantization error, particularly in the low output signal response range of the detector [17].

Scatter radiation is another source of image degradation in radiography, and it acts in the same manner as other noise sources. Scatter reduces the available dynamic range of x-ray intensities at the exit side of the patient. The primary effect of scatter is the reduction of subject contrast. In addition to reducing subject contrast, scatter decreases the SNR because it contains no signal but does contain Poisson quantum noise. Scanned slot DR detectors possess inherent scatter rejection capability and do not require the use of a grid. Area detectors used in CR, DR, and SF radiography should be used with an antiscatter grid in clinical imaging situations in which scatter dominates (usually patient thicknesses in excess of 10 cm). They are often not used with pediatric and extremity imaging. The use of grids is especially important in CR because of the increased scatter sensitivity of barium halide (k-edge approximately 35 keV) compared with SF screens such as gadolinium oxysulfide (k-edge approximately 50 keV) (see Figure 1). When using stationary grids (eg, snap-on grids used in portable x-ray and for lateral tabletop exposures), consider the use of high-frequency grid strips of greater than 60 lines/cm (140 lines/in) to avoid grid aliasing patterns caused by insufficient sampling. Consultation with the vendor for an appropriate grid is recommended.

When using scatter grids with digital detectors, any adjustment in exposure technique (mAs) should be made in the interest of maintaining SNR. Insertion of the grid reduces both the signal (incomplete transmission of the primary radiation by the grid) and noise (removal of scattered radiation), with the relative amount of reduction dependent on both the scatter-to-primary ratio in the beam (higher for larger patients) and the design of the grid. This is different than in SF radiography, in which upward mAs compensation is required both for the removal of primary and scattered radiation to achieve the correct film optical density range. For this reason, the adjustment of radiographic technique upward according to the Bucky factor (defined as the increase in exposure required to obtain the same optical density as with no

grid) overestimates the appropriate technique increase for digital receptors.

Contrast

Contrast (radiographic contrast) is proportional to the magnitude of the signal difference between the structure of interest and its surroundings in the displayed image, which is expressed in terms of the optical density difference between two adjacent areas on the SF film or as the relative brightness difference between the corresponding areas in a digital image displayed on a monitor. For both SF and digital imaging, radiographic contrast is influenced by subject contrast and receptor sensitivity. However, in digital imaging, contrast in the displayed image can also be altered by the adjustment of display parameters independent of the acquisition parameters. Subject contrast is proportional to the relative difference in x-ray exposure on the exit side of the patient and is the result of the attenuating properties of the tissues under study. Attenuation is strongly dependent on the x-ray energy spectrum and is determined by the target material, kilovoltage, and total beam filtration. Subject contrast is further reduced by the presence of scatter. Receptor sensitivity is defined as the amount by which the output (optical density for SF or analog-to-digital unit value for CR and DR) changes per unit change in exposure to the receptor. In digital detector design, sensitivity can, to a certain extent, be selected through electronic gain settings, but this must be done with an awareness of the range of exposures that must be simultaneously measured, taking into account the detector response energy dependence. With a properly designed image acquisition system, the dynamic range should be adequate to measure the entire range of intensities from that of the unattenuated beam outside the patient to that through the densest, thickest part of the anatomy. The signal stored in digital form is directly (or logarithmically) proportional to the amount of radiation transmitted through the patient. For this reason, the stored for-processing image reflects the inherent subject contrast very faithfully, provided the image receptor is operated in the linear (or log-linear) portion of its response curve.

For CR and DR, image processing (usually proprietary to the device's manufacturer and not under control of the radiologist) is used to determine display contrast by establishing the relationship between raw pixel values and gray-scale levels. This produces the final for-presentation image for interpretation either on hard-copy film or soft-copy displays. Discussion of image processing, including contrast enhancement and spatial frequency processing, is presented in the companion paper on image processing and display.

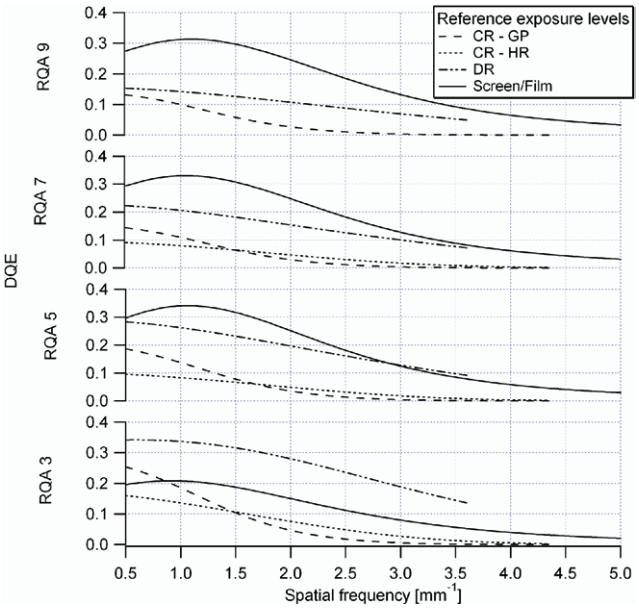

Fig 3. Detective quantum efficiencies of screen-film, general-purpose computed radiography (GP-CR), high-resolution computed radiography (HR-CR), and selenium-based digital radiography (DR) systems. The detective quantum efficiencies (DQEs) were measured using 4 different standardized radiation quality beam qualities, with higher radiation quality (RQA) number corresponding to higher average beam energy [23].

Detective Quantum Efficiency

The DQE is a useful descriptor of image receptor performance because it takes into account detection efficiency, spatial resolution, and noise. The DQE describes the relative efficiency in maintaining the signal-to-noise level available in the imaging process and is defined as SNR^2_{out}/SNR^2_{in}, where SNR^2_{in} is the SNR of the exposure incident on the receptor, numerically equal to the input fluence. SNR_{out} is the ratio of the relative magnitudes of the image signal to the noise. Thus, DQE can be considered the system's SNR transfer efficiency. Efficient SNR transfer is important because the detection performance of both human and ideal observers improves with increasing SNR [18-22].

The DQE is generally plotted against spatial frequency, DQE(f). A perfect system has a DQE(f) of unity at all spatial frequencies. Real systems lose efficiency at higher spatial frequencies because of decreasing spatial resolution and increasing noise sources. DQE(f) is determined by quantitative measurements of the modulation transfer function, MTF(f), to determine the spatial resolution and the noise power spectrum, NPS(f), to determine the noise characteristics, both as a function of spatial frequency, using an accurately known

input exposure to the detector. The exposure required to achieve a given SNR in the image is inversely related to the system DQE. Therefore, systems with higher DQE require less exposure for a given image quality (SNR) than systems with lower DQE. The DQE is important in CR and DR because these technologies can offer significantly improved DQE compared with SF systems. However, it is important to remember that the DQE is an energy-dependent quantity. Figure 3 compares the DQE of a general-purpose radiographic SF system with those of a general-purpose CR system, a high-resolution CR system, and a DR system. Plots are shown for 4 different standardized radiation quality x-ray beams, with the higher radiation quality number corresponding to higher beam energy. Figure 3 demonstrates that the DQEs of the CR and DR systems decrease compared with that of SF with increasing beam energy. As pointed out by the study authors, this suggests that automatically increasing the beam energy when switching from SF to digital radiography may not always be the most appropriate course of action [23].

DOSE CREEP IN DIGITAL RADIOGRAPHY AND ITS PREVENTION

Film-based imaging provides immediate feedback to technologists and radiologists concerning the delivery of the proper radiation exposure to patients. If the optical densities in the film image are too great, the patient received too much radiation, whereas reduced optical densities indicated a reduced radiation exposure relative to the proper value. When digital image receptors replace SF image receptors, the brightness and contrast in the displayed image on the monitor are independent of the exposure used during acquisition. Only the noise in the image changes with the exposure. Radiologists react negatively to digital images with excessive noise (low patient exposure) but rarely complain about images with reduced noise due to excessive patient exposures. Technologists quickly recognize this and gradually increase patient doses by adjusting radiographic techniques upward. This leads to the potential for dose creep in CR and DR [6-8]. The medical physicist and/or imaging specialist should monitor for dose creep on a consistent, ongoing basis, because it is a recurring phenomenon. One effective way to eliminate dose creep is to develop validated radiographic technique charts for all performed examinations as a function of patient size. If these factors are entered into the anatomic programming of state-of-the-art x-ray generator controls, the technologists simply select the radiologic examination and patient size at the beginning of the case and are assured of the use of standard radiographic technique factors and standard radiation exposure regardless of whether the examination is conducted with the image receptor on the table top with manual techniques or in the Bucky tray using the automatic exposure control (AEC) [24].

EXPOSURE INDICATORS FOR DIGITAL RADIOGRAPHY

Currently, most CR manufacturers provide exposure indicators on images displayed on the image reader workstation. However, this information may not be transferred over to the picture archiving and communication system (PACS) or may appear buried in the DICOM information page of the patient. Most PACS have the ability to display the exposure indicator on the image. Check with your PACS vendor and insist that this information be displayed on each image. Today, the various numerical exposure indicators used by different CR manufacturers are not easily comparable. To compound the problem further, some DR manufacturers do not provide an exposure indicator on their processed images, or the information is not transferred to the PACS and is not recoverable.

The American Association of Physicists in Medicine formed Task Group 116 to address the lack of a uniform exposure indicator on digital radiography images. Their report, "Recommended Exposure Indicator for Digital Radiography," will propose standardized radiation exposure conditions from which a relative exposure indicator can be defined. Because the major manufacturers of digital radiographic image receptors participated in the task group, this standardized exposure indicator should migrate into their future products. See *exposure value estimate* in the glossary in the companion paper on image processing and display, which summarizes several of the exposure indices currently used by different manufacturers.

PEDIATRIC IMAGING ISSUES

As with all pediatric imaging examinations involving ionizing radiation, CR and DR should be performed using the lowest possible radiation exposure to the patient (as low as reasonably achievable [ALARA]), because pediatric patients are believed to be up to 10 times more sensitive to ionizing radiation than adults [25]. This suggests that the exposure delivered to the digital image receptor should be approximately 0.4 to 0.8 mR for chest and abdominal examinations [26]. However, it should be noted that adopting a policy of reduced exposure levels to the image receptor to achieve ALARA in pediatric imaging increases the possibility of unacceptably high quantum noise and thus reduces the margin for error in the selection of the parameters that affect the beam quality. Also, the achievable reduction in exposure at the plate is dependent on how well the digital system is optimized for pediatric imaging and on the nature of the particular disease being diagnosed [27].

Commitment and effort on the part of staff members is required to achieve ALARA in pediatric imaging. First, the multiple x-ray machines of a given type in a department require validation, calibration, and matching to

ensure the same exposure for similar radiographic technique parameters. The actual exposure required to produce clinically acceptable images for a given digital receptor is a function of the digital receptor's DQE. The goal is to produce similar image quality in all images regardless of the device chosen and regardless of the size of the pediatric patient. Second, all CR readers or DR image receptors should be adjusted to provide a uniform response to the recorded x-ray image pattern in space. Third, all monitors on all diagnostic display workstations should be tested and calibrated to provide a uniform appearance of displayed images. These topics are addressed in more detail elsewhere [28].

Pediatric patients provide unique challenges when imaged with digital radiography. Uncooperative pediatric patients may require sedation or anesthesia. The range in size from the neonate to the young adult patient requires a wide range in radiographic technique factors and a digital image receptor that properly processes an x-ray pattern in space that may or may not cover the entire image receptor. The decision to use an antiscatter grid must be carefully considered as a function of patient size. Standard positioning aids used to immobilize pediatric patients may generate unacceptable artifacts when used with digital image receptors. Because the x-ray pattern from a pediatric chest compared with an adult chest, for example, has a different dynamic range and other characteristics [24], display parameters used by the digital acquisition device to properly display the digital image may need to be unique as a function of patient size. Display parameters require further alteration if gonadal shielding or orthopedic implants appear in the displayed image. Finally, because scoliosis examinations are common in children, the digital image receptor must provide an efficient method to generate images up to 36 inches in length, preferably without doubly exposing some sections of the patient's anatomy.

PROPER EXPOSURE FACTORS FOR STANDARD EXAMINATIONS

Exposure (technique) charts are part of the standard of care expected by the Joint Commission on Accreditation of Healthcare Organizations and are required in many states. For example, in Ohio, exposure charts are required and must include specification of the body part, thickness, and image receptor components to be used for the procedure. Internationally, most countries require regular use of exposure charts. It is necessary to check your county, state, and local regulations for any specific requirements. You may also be required to compute estimates for entrance skin exposures for these charts.

Exposure charts are an essential quality assurance component of any diagnostic x-ray imaging department. These exposure guides are especially important in CR and DR, in which the wide exposure latitude makes it possible to obtain clinical images that are severely underexposed, resulting in noisy images, or grossly overexposed, resulting in possible data saturation as well as high patient doses. Improper exposures in digital radiography are more likely to go unnoticed because image processing can correct these deficiencies to a much greater degree than SF systems because of the wide latitude of the image receptor. Exposure charts are critical in the portable x-ray environment with CR, because manual technique factors are normally required. An underexposed radiograph will have unacceptable noise, while an overexposed radiograph may appear to be of exceptional quality at the expense of patient dose.

The first challenge is to understand what constitutes a properly exposed radiograph [29-31]. In SF systems, the limited dynamic range of the film determines the exposure latitude. In the digital world, wider exposure latitude can be used to take into account the differing energy responses of the systems and the variability introduced in the dose estimation scheme. Appendix A provides a suggested range of exposures to facilitate identifying whether a proper radiograph has been obtained. The log of median exposure, sensitivity number, exposure index, and sensitivity value are representative exposure indicators used by several CR vendors. The detector exposure column lists an approximate range of corresponding incident exposure values to the CR imaging plate. Note that repeating overexposures is not recommended unless information has been lost. At the low dose end of the scale, it is important for the radiologist and technologist to review underexposed radiographs to develop a comfort level for which radiographs need to be repeated. It is useful to provide some guidance to the technologist with respect to how to correct an improperly exposed radiograph (see Appendix B). Exposure charts must be tailored for each digital radiography x-ray system and detector combination [32]. There is considerable variability in image receptor response because of varying scatter sensitivity, the use of grids with different grid ratios, collimation, beam filtration, the choice of kilovoltage, source to image distance, and image receptor size. Many CR and DR systems also permit the choice of several exposure (speed) classes. Technique factors for each clinically used exposure class should be constructed. In addition, exposure charts should be designed to function over the wide range of adult patient sizes seen in our clinics today and over the even wider range of sizes found in pediatric radiography.

Given a table of body-part thicknesses with suggested technique factors, one can derive, using a spreadsheet program, a full table of values using knowledge of the effect of thickness and kVp on the penetration of the beam. There are also several commercial programs that can be tailored to digital radiography. Typically, the entire exposure chart is calibrated to a single set of exposures based on a common body part, such as the abdomen. A water tank or anthropo-

morphic phantom can be used to develop appropriate techniques for the abdominal radiograph. Water works better than acrylic or aluminum because the scatter characteristics are more comparable with the clinical situation. As discussed previously, scatter characteristics can be quite different for digital radiography than for SF. It is also necessary to measure Bucky factors for the undertable and wall Bucky assemblies as well as for any snap-on grids (also known as grid caps, portable grids) that may be used with the x-ray machine. Once exposure factors have been determined that result in an appropriate radiograph for the abdomen with the various image receptors, all other values for mAs are interpolated using corrections for distance, Bucky factors, exposure class, patient thickness, and machine output. The exposure chart is then cleaned up by rounding values of mAs to match available mAs stations on the machine. An example of a technique chart developed for CR is shown in Appendix C.

Similar exposure charts should be constructed for the use of portable x-ray equipment. The x-ray output of a portable radiography unit is generally substantially lower than that of a 3-phase or high-frequency stationary x-ray machine. This reduced output capability results in increased exposure times compared with conventional radiography. As a result, patient motion is a common image degradation factor. This factor, coupled with the difficulty in positioning grids correctly in a portable setting, results in frequent exposures being taken without the use of a grid, also degrading overall image quality. Manual techniques are used because phototimers are usually not available in the portable setting. An example of an exposure chart suitable for portable radiography in an emergency department trauma setting is shown in Appendix D.

Another issue to consider, particularly for portable radiography, is achieving reproducible image quality for repeated studies such as portable chest images that are used for radiographic detection and evidence of incremental changes. Important for the radiologist is consistent acquisition techniques by the technologist, including kVp, mAs, geometry (source to image distance, patient positioning), the orientation of imaging receptor cassette, the use of a grid, technologist positioning markers, and the consistent use of image-processing parameters that are optimized for the specific study. Acquisition protocols and technique charts with minimal parameters and variables (eg, using crosswise image acquisition for all adult chests with decubitus grids, using a fixed source-to-image receptor distance (SID) of 127 cm [50 in], 110 kVp, and adjusting only mAs as a function of estimated patient girth) are key to establishing the desired reproducibility of image presentation (orientation, magnification, image enhancement) on the radiologist display workstation. An example of an exposure chart specifying these requirements for adult chest portable imaging and other adult examinations is shown in Appendix E.

AUTOMATIC EXPOSURE CONTROL

In CR, phototimers are frequently used with wall and undertable Buckys in emergency, inpatient, and outpatient radiology department settings. These AEC units are designed to turn off the x-ray generator when an appropriate exposure level has been received at the image receptor. Automatic exposure controls work well if properly calibrated and properly positioned over clinically important areas of the patient. They fail if placed incorrectly or if pathology in the AEC region of interest causes a significantly different density to be present than expected. Automatic exposure control devices are energy dependent and may require calibration at multiple kVps to function properly over a wide range of patient sizes. Some x-ray generators have preprogrammed energy response curves, and these units may not work as well with CR as with conventional SF receptors because they were probably manufactured to be matched to typical SF energy and scatter characteristics. In this situation, calibrate the wall Bucky for chest work and the undertable Bucky for abdominal procedures, or whatever is the most common procedure performed on that image receptor at your institution.

In DR, the AEC is normally built into the image receptor assembly, and it works in a manner similar to the phototimers used in SF radiography. However, the number or area of the photosensitive elements may differ from the configurations normally used with conventional SF systems depending on vendor implementation.

Most x-ray generator phototimers are designed to be calibrated for multiple image receptor speeds, usually designated by names such as *regular*, *chest*, and *detail cassettes*. AECs can be calibrated to multiple exposure classes such as 400, 200, and 100 to cover a variety of different CR and DR receptors or specific clinical situations such as pediatrics [33,34]. Some CR vendors permit a variety of exposure classes in their software, so the exposure class on the x-ray console must be manually selected to match the exposure class appropriate for that procedure. This can lead to errors if technologists are not vigilant in matching the x-ray generator exposure class to the CR reader exposure class. With some implementations of CR, the AEC unit must be calibrated on the basis of cassette size instead of exposure class. Check with the vendor so that you know which method is most appropriate for you. In DR, the exposure class may also be user selectable, and vigilance is necessary to ensure that technologists are using the appropriate exposure classes, especially for pediatrics. It is usually set automatically in digital radiography systems with an integral x-ray tube assembly.

SUMMARY: OBSERVATIONS AND CHALLENGES

The rapid movement in radiography away from SF technology to both CR and DR has created both improved efficiency and more options in radiology. Yet at the same time, the increased complexity and lack of transfer of skills learned in the SF era by both radiologists and technologists has produced a gap in the ability to perform quality assurance and quality improvement. It is imperative to understand these digital radiography systems to provide quality radiology services and to protect patients from unnecessary radiation dose.

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Appendix A

On the basis of the vendor's recommendation for what constitutes an appropriately exposed radiograph, one can use Table A1 or a similar table to provide guidance to the technologist regarding proper exposure levels. Several CR exposure indices are illustrated in this example.

With too low an incident exposure, the image is dominated by noise, the corresponding SNR is very poor, and the ability to detect subtle differences in x-ray attenuation is compromised. With too high an incident exposure, highly penetrated areas in the image may suffer from saturation effects and the loss of contrast that cannot be compensated for with image processing. These recommendations depend on several issues, including the type of CR reader (newer systems, such as the dual-light collection systems, have a higher DQE and will consequently have a lower exposure recommendation for all indications), the type of image plate (later-generation image plates may have better exposure performance), the quality and calibration settings of the CR reader and image plate, and the tolerance of the radiologist to image noise, among other considerations. Continuous analysis and feedback are required to get the optimal range of exposures and indications for a given examination at a given site. Similar guidance can be provided for digital radiography for those systems that provide an indication of the exposure used to obtain a radiograph. Use caution in applying this chart to any specific CR implementation since x-ray generator output characteristics and CR plate energy sensitivities vary widely.

Appendix B

Table B1 is a conversion factor table that provides a rapid means for the technologist to understand how to correct for an improperly exposed radiograph. This example is for one CR vendor implementation (Agfa), using log of median exposure as the exposure indicator. Similar tables can be constructed using the information in Table A1 as a guideline for other vendor exposure settings. Technologists are very familiar with the change required to obtain an appropriate SF radiograph, and the same concept is used here to suggest improvements in technique factors with CR, as well as in deciding when to repeat a study. Use caution in applying this chart to any specific CR implementation since x-ray generator output characteristics and CR plate energy sensitivities vary widely.

Appendix C

Table C1 is an example of an exposure chart for CR for a 3-phase, 12-pulse generator. Distances are in inches. When constructing a clinical exposure chart, the mAs values should be adjusted to match the nearest mAs station available on the generator.

Appendix D

Table D1 is a typical mobile radiography exposure chart for CR used in an emergency department trauma setting. This example includes the use of 3 exposure classes, depending on the clinical situation. Note the use of lower kVps and longer exposure times and the minimal use of grids in this portable trauma setting compared with the previous fixed 3-phase generator exposure chart (Table C1). Use caution in applying this chart to any specific CR implementation since x-ray generator output characteristics and CR plate energy sensitivities vary widely.

Appendix E

Table E1 is an example of an exposure chart developed for bedside CR with the standardization of parameters. In this example, all AP chest examinations are performed crosswise (landscape orientation). All adult chest procedures are performed using a decubitus grid. If their conditions allow, patients are inclined to get as close to upright as is possible. Use caution in applying this chart to any specific CR implementation since x-ray generator output characteristics and CR plate energy sensitivities vary widely.

Table A1. Suggested guidelines for QC action based on estimated incident exposure (Conventional CR system for an "equivalent" speed class = 200)

Fuji (S Number)	Agfa (lgM Value)	Kodak (Exposure Index)	Detector Exposure (mR)	Indication
>1,000	<1.45	<1,250	<0.2	Underexposed: repeat if necessary
1000-601	1.45-1.74	1,250-1,549	0.2-0.3	Underexposed: QC exception
600-301	1.75-2.04	1,550-1,849	0.3-0.7	Underexposed: QC review
300-150	2.05-2.35	1,850-2,150	0.7-1.3	Acceptable range
149-75	2.36-2.65	2,151-2,450	1.3-2.7	Overexposed: QC review
74-50	2.66-2.95	2,451-2,750	2.7-4.0	Overexposed: QC exception
<50	>2.95	>2,750	>4.0	Overexposed: QC repeat if necessary

Note: lgM = log of median exposure; QC = quality control; S number = sensitivity number.

Table B1. Required change in milliampere-seconds (mAs) to achieve a target log of median exposure (lgM) value of 2.20

	Underexposed: Increase mAs				Proper Clinical Techniques				Overexposed: Reduce mAs			
	LgM Value	Difference From Target LgM	Required mAs Scale Factor		LgM Value	Difference From Target LgM	Required mAs Scale Factor		LgM Value	Difference From Target LgM	Required mAs Scale Factor	
Underexposed, repeat	1.40	−0.80	6.31		2.00	−0.20	1.58		2.42	0.22	0.60	Overexposed
	1.42	−0.78	0.03		2.02	−0.18	1.51		2.44	0.24	0.58	
	1.44	−0.76	5.75		2.04	−0.16	1.45		2.46	0.26	0.55	
	1.46	−0.74	5.50		2.06	−0.14	1.38		2.48	0.28	0.52	
	1.48	−0.72	5.25		2.08	−0.12	1.32		2.50	0.30	0.50	
	1.50	−0.70	5.01		2.10	−0.10	1.26		2.52	0.32	0.48	
	1.52	−0.68	4.79		2.12	−0.08	1.20		2.54	0.34	0.46	
	1.54	−0.66	4.57		2.14	−0.06	1.15		2.56	0.36	0.44	
	1.56	−0.64	4.37		2.16	−0.04	1.10		2.58	0.38	0.42	
	1.58	−0.62	4.17		2.18	−0.02	1.05		2.60	0.40	0.40	
	1.60	−0.60	3.98	>>>	2.20	0.00	1.00	<<<	2.62	0.42	0.38	Overexposed, check for burnout, repeat if necessary
	1.62	−0.58	3.80		2.22	0.02	0.95		2.64	0.44	0.36	
	1.64	−0.56	3.63		2.24	0.04	0.91		2.66	0.46	0.35	
	1.66	−0.54	3.47		2.26	0.06	0.87		2.68	0.48	0.33	
	1.68	−0.52	3.31		2.28	0.08	0.83		2.70	0.50	0.32	
	1.70	−0.50	3.16		2.30	0.10	0.79		2.72	0.52	0.30	
	1.72	−0.48	3.02		2.32	0.12	0.76		2.74	0.54	0.29	
	1.74	−0.46	2.88		2.34	0.14	0.72		2.76	0.56	0.28	
	1.76	−0.44	2.75		2.36	0.16	0.69		2.78	0.58	0.26	
	1.78	−0.42	2.63		2.38	0.18	0.68		2.80	0.60	0.25	
Underexposed, review with radiologist	1.80	−0.40	2.51		2.40	0.20	0.63		2.82	0.62	0.24	
	1.82	−0.38	2.40						2.84	0.64	0.23	
	1.84	−0.36	2.29						2.86	0.68	0.22	
	1.86	−0.34	2.19						2.88	0.68	0.21	
	1.88	−0.32	2.09						2.90	0.70	0.20	
	1.90	−0.30	2.00						2.92	0.72	0.19	
	1.92	−0.28	1.91						2.94	0.74	0.18	
	1.94	−0.26	1.82						2.96	0.76	0.17	
	1.96	−0.24	1.74						2.98	0.78	0.17	
	1.98	−0.22	1.68						3.00	0.80	0.15	

Example 1: LgM value was 1.80. To get a proper exposure of 2.2, increase the mAs by a factor of 2.5. If the mAs was 20, change it to 2.5 × 20 = 50 mAs. You could also increase the kilovolt peak (kVp). Remember, a 15% increase in kVp doubles the tube output.
Example 2: LgM value was 3.02. To get a proper exposure of 2.2, decrease the mAs by a factor of 0.16. If the mAs was 20, change it to 0.16 × 20 = 32 mAs. You could also decrease the kVp to achieve the same effect.

Table C1. Computed radiography exposure chart

Procedure	View	Distance	Receptor	Speed class	kVp	Total mAs for Different Body Part Thicknesses								
						Average cm	−6 cm	−4 cm	−2 cm	Average	+2 cm	+4 cm	+6 cm	+8 cm
Chest	PA/AP	72	W	200	106	72	1.25	1.67	2.50	3.33	5.00	6.67	10.00	13.33
	LAT	72	W	200	116	30	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	AP	72	T	200	80	22	1.25	1.67	2.50	3.33	5.00	6.67	10.00	13.33
	LAT	72	T	200	90	30	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
Supine chest	AP	40	T	200	80	22	0.90	1.20	1.80	2.40	3.60	4.80	7.20	9.60
Ribs/sternum	AP Above Diaph	72	W	400	60	22	10.00	13.33	20.00	26.67	40.00	53.33	80.00	106.67
	OBL Above Diaph	72	W	400	64	24	15.00	20.00	30.00	40.00	60.00	80.00	120.00	160.00
	AP Below Diaph	72	W	400	76	22	30.00	40.00	60.00	80.00	120.00	160.00	240.00	320.00
Abdomen	AP/PA	40	B	400	80	22	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	30° OBL	40	B	400	60	24	11.00	14.67	22.00	29.33	44.00	58.67	88.00	117.33
Pelvis/hip	AP	40	B	400	80	22	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	LAT	40	B	400	80	22	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
Sacrum	AP	40	B	400	76	20	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	LAT	40	B	400	80	28	15.00	20.00	30.00	40.00	60.00	80.00	120.00	160.00
Lumbar spine	AP	40	B	400	80	22	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	45° OBL	40	B	400	80	26	15.00	20.00	30.00	40.00	60.00	80.00	120.00	160.00
	LAT	40	B	400	80	30	30.00	40.00	60.00	80.00	120.00	160.00	240.00	320.00
	L5/S1	40	B	400	90	30	30.00	40.00	60.00	80.00	120.00	180.00	240.00	320.00
Thoracic spine	AP	40	B	400	76	22	6.50	8.87	13.00	17.33	26.00	34.67	52.00	69.33
	LAT	40	B	400	76	30	15.00	20.00	30.00	40.00	60.00	80.00	120.00	160.00
Cervical spine	AP/Odon	40	B	400	76	13	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	AP/Odon	72	W	400	76	13	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	OBL/LAT	72	W	400	76	13	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
	OBL/LAT	72	T	400	76	13	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
Skull	PA/Caldwell	40	B	400	80	19	5.00	6.67	10.00	13.33	20.00	26.67	40.00	53.33
	LAT	40	B	400	76	15	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	Townes	40	B	400	80	22	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
Sinuses/facial bones	PA/Caldwell	40	B	400	76	19	5.00	6.67	10.00	13.33	20.00	26.67	40.00	53.33
	Waters	40	B	400	80	20	5.00	6.67	10.00	13.33	20.00	26.67	40.00	53.33
	LAT	40	B	400	76	15	1.88	2.50	3.75	5.00	7.50	10.00	15.00	20.00
Hand	PA	40	T	100	54	4	1.10	1.47	2.20	2.93	4.40	5.87	8.80	11.73
	OBL	40	T	100	54	6	1.25	1.67	2.50	3.33	5.00	6.67	10.00	13.33
	Fanned LAT	40	T	100	54	8	2.25	3.00	4.50	6.00	8.00	12.00	16.00	24.00
Wrist	PA	40	T	100	60	5	1.10	1.47	2.20	2.93	4.40	5.67	6.80	11.73
	OBL	40	T	100	60	6	1.85	2.20	3.30	4.40	6.60	8.80	13.20	17.60
	LAT	40	T	100	60	7	2.25	3.00	4.50	6.00	8.00	12.00	18.00	24.00
Forearm	AP	40	T	100	64	7	1.10	1.47	2.20	2.93	4.40	5.87	8.80	11.73
	LAT	40	T	100	64	8	2.25	3.00	4.50	6.00	8.00	12.00	18.00	24.00

AP = anterior-posterior; Diaph = diaphragm; L5/S1 = fifth lumbar vertebrae/first sacra vertebrae; LAT = Lateral; OBL = oblique; Ondo = ondotoid; PA = posterior-anterior.

Table C1. Continued

Procedure	View	Distance	Receptor	Speed class	kVp	Total mAs for Different Body Part Thicknesses								
						Average cm	−6 cm	−4 cm	−2 cm	Average	+2 cm	+4 cm	+6 cm	+8 cm
Elbow	AP	40	T	100	64	7	1.50	2.00	3.00	4.00	8.00	8.00	12.00	16.00
	OBL	40	T	100	64	8	2.00	2.67	4.00	5.33	8.00	10.87	16.00	21.33
	LAT	40	T	100	64	8	3.00	4.00	6.00	8.00	12.00	16.00	24.00	32.00
Humerus	AP	40	T	100	72	8	1.10	1.47	2.20	2.93	4.40	5.67	8.80	11.73
	LAT	40	T	100	72	9	1.80	2.40	3.60	4.80	7.20	9.60	14.40	19.20
Toes	All	40	T	100	60	2	1.10	1.47	2.20	2.93	4.40	5.67	8.80	11.73
Foot	AP	40	T	100	64	7	1.25	1.67	2.60	3.33	5.00	6.67	10.00	13.33
	OBL	40	T	100	64	8	1.65	2.20	3.30	4.40	6.60	8.80	13.20	17.50
	LAT	40	T	100	64	8	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
Calcanus	PA	40	T	100	68	9	3.50	4.67	7.00	9.33	14.00	18.67	26.00	37.33
	LAT	40	T	100	68	8	3.50	4.67	7.00	9.33	14.00	18.67	28.00	37.33
Ankle	AP/OBL	40	T	100	64	9	1.80	2.40	3.60	4.80	7.20	9.60	14.40	19.20
	LAT	40	T	100	64	8	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
Tibia/fibula	AP	40	T	100	68	11	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	LAT	40	T	100	68	10	2.50	3.33	5.00	5.57	10.00	13.33	20.00	26.67
Knee	AP	40	T	100	70	12	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	LAT	40	T	100	70	11	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	AP/OBL	40	B	100	70	12	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
	LAT	40	B	100	70	11	2.50	3.33	5.00	6.67	10.00	13.33	20.00	26.67
Femur	AP/LAT	40	B	400	76	16	6.00	8.00	12.00	16.00	24.00	32.00	48.00	64.00
Shoulder	AP/Transax	40	B	400	76	18	3.00	4.00	6.00	8.00	12.00	16.00	24.00	32.00
	Transthoracic	40	B	400	90	40	7.50	10.00	15.00	20.00	30.00	40.00	60.00	80.00
Clavicle	AP/PA	40	B	400	74	16	3.00	4.00	6.00	8.00	12.00	16.00	24.00	32.00
Scapula	AP	40	B	400	74	18	3.00	4.00	6.00	8.00	12.00	16.00	24.00	32.00
	LAT	40	B	400	74	24	5.00	6.67	10.00	13.33	20.00	26.67	40.00	53.33

Note: AP = anterior-posterior; B = under table Bucky with grid; LAT = lateral; mAs = milliampere-seconds; OBL = oblique; PA = posterior-anterior; T = tabletop nongrid; Transax = trans-axial; W = wall Bucky with grid.

Table D1. Mobile computed radiography exposure chart for trauma radiology

Body Part	Position	Patient Size	kVp	mAs	Distance	Speed Class	Grid
Chest	PA/AP	Small	80	5	72″	200	No
	Upright 72″	Average	85	10	72″	200	No
		Large	90	16	72″	200	No
	PA/AP	Small	70	5	40″	200	No
		Average	75	8	40″	200	No
		Large	85	10	40″	200	No
	Lateral-72″	Small	85	10	72″	200	No
		Average	90	20	72″	200	No
		Large	95	32	72″	200	No
	Lateral	Small	75	10	40″	200	No
		Average	80	16	40″	200	No
		Large	90	20	40″	200	No
	AP	Newborn	65	0.5	40″	400	No
	Lateral	Newborn	70	1	40″	400	No
Abdomen	AP-Grid	Average	70	48	40″	400	Yes
Ankle	AP	Average	55	12.55	40″	100	No
	Lateral	Average	55	10	40″	100	No
Cervical spine	AP	Average	60	7.5	40″	400	No
	Lateral-72″	Average	65	15	72″	400	No
Elbow	AP	Average	55	12.5	40″	100	No
	Lateral	Average	55	10	40″	100	No
Femur	AP/Lateral	Average	55	10	40″	400	No
Foot	AP	Average	52	8	40″	100	No
Hand/Wrist	PA	Average	50	5	40″	100	No
	Lateral	Average	55	8	40″	100	No
Humerus	AP/Lateral	Average	55	2.5	40″	100	No
Knee	AP	Average	55	25	40″	100	No
	Lateral	Average	55	16	40″	100	No
Lumbar spine	AP-Grid	Average	75	48	40″	400	Yes
	Lateral-Grid	Average	85	150	40″	400	Yes
Pelvis/Hip	AP	Average	55	24	40″	400	No
	AP-Grid	Average	70	50	40″	400	Yes
	Lateral-Grid	Average	80	75	40″	400	Yes
Shoulder/Clavicle	AP	Average	55	5	40″	400	No
Skull	AP	Average	60	16	40″	400	No
	Lateral	Average	55	10	40″	400	No
	AP-Grid	Average	75	20	40″	400	Yes
	Lateral-Grid	Average	65	10	40″	400	Yes
Thoracic spine	AP	Average	65	16	40″	400	No

Note: AP = anterior-posterior; kVp = kilovolt peak; mAs = milliampere-seconds.

Table E1. Exposure chart: example of computed radiography exposure chart using grids and standardized parameters

Anatomic Region	Examination	Measurement (cm)	SID	Grid	kVp	Small (mAs)	Medium (mAs)	Large (mAs)
Chest[a]	Chest portable AP	20-30	50″	Yes[a]	110	6	10	12
Skull	Skull PA/AP	18-21	40″	Yes	80	12	20	30
	Skull lateral	14-17	40″	Yes	80	6	12	18
Spine	Cervical AP	11-14	40″	Yes	80	6	12	18
	Thoracic AP	20-24	40″	Yes	80	18	26	40
	Lumbar AP	18-22	40″	Yes	80	24	40	64
Abdomen	Abdomen (KUB)	18-22	40″	Yes	80	24	40	64
Pelvis	Pelvis AP	19-23	40″	Yes	80	24	40	64
	Hip AP	17-21	40″	Yes	80	20	35	50
	Cross table lateral hip	25-31	40″	Yes	85	30	42	55
Upper extremities	Finger	1.5-4	40″	No	60	1	1	2
	Hand AP/OBL	3-5	40″	No	60	1	2	4
	Hand lateral	3-5	40″	No	60	2	3	4
	Wrist AP/OBL	3-6	40″	No	60	1.5	2	3
	Wrist lateral	3-6	40″	No	60	1.5	3	4
	Forearm AP	6-8	40″	No	60	2	3	4
	Forearm lateral	6-8	40″	No	60	2	3	4
	Elbow	6-8	40″	No	60	2	3	4
	Humerus	7-10	40″	Yes	80	6	10	13
	Shoulder	12-16	40″	Yes	80	7	12	15
Lower extremities	Toes	1.5-4	40″	No	60	1.5	2.5	4
	Foot AP/OBL	6-8	40″	No	60	2	3	5
	Foot lateral	6-8	40″	No	60	2.5	4	6
	Ankle AP	8-10	40″	No	60	3	5	7
	Ankle lateral	8-10	40″	No	60	2	3	6
	Tib/Fib AP	10-12	40″	No	60	2	4	6
	Tib/Fib lateral	10-12	40″	No	60	2	4	6
	Knee AP/lateral	10-13	40″	No	60	6	12	20
	Knee AP/lateral	10-13	40″	Yes	60	3	5	8
	Femur AP/lateral	14-17	40″	Yes	80	15	25	40

Note: AP = anterior-posterior; Fib = fibula; KUB = Kidneys, Ureter, Bladder; kVp = kilovolt peak; large = >30 cm; mAs = milliampere-seconds; medium = 25 to 30 cm; OBL = oblique; PA = posterior-anterior; SID = Source to Image receptor Distance; small = 20 to 25 cm; Tib = tibia.

[a]Use short dimension (decubitus) grid, position cassette always cross-wise (landscape).